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Control and Trajectory Generation of a Wearable Mobility Exoskeleton for Spinal Cord Injury Patients

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#### Control and Trajectory Generation of a Wearable Mobility Exoskeleton for Spinal Cord Injury Patients

By

Timothy Alan Swift

A dissertation submitted in partial satisfaction of the

requirements for the degree of

Doctor of Philosophy

in

Engineering – Mechanical Engineering

in the

Graduate Division

of the

University of California, Berkeley

Committee in charge:

Professor Homayoon Kazerooni, Chair Professor Kameshwar Poolla Professor Pieter Abbeel

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#### ABSTRACT

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By

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#### Doctor of Philosophy in Engineering – Mechanical Engineering

University of California, Berkeley

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There are currently nearly 1.3 million people in the United States who have some form of lower extremity paralysis due to a spinal cord injury (SCI). For many of these subjects, a wheelchair is their primary means of mobility which brings along with it a collection of health complications stemming from extended periods of time in a seated position such as osteoporosis, and impaired bowel and circulatory function. To provide these SCI patients with a viable upright mobility option, this work presents the control structure and trajectory generation method for a mobility exoskeleton that allows them to ambulate while reliably generating a gait similar to that of an unimpaired subject. This work is developed using the exoskeleton lower extremity gait system (eLEGS) that is developed by Berkeley Bionics. This device has actuation in the sagittal plane at the hip and knee joints with a passively sprung ankle joint and relies on the user input through lofstrand crutches for overall balance.

The control structure is developed using three independent control levels. The first is the human machine interface (HMI) which is a discrete level that determines the user's intent. Second is a trajectory generation level which is a continuous controller that translates the user intent into desired joint responses based on the system state information. Finally, a low level controller that computes the continuous closed loop control output required for the individual motors to produce the desired joint response. The trajectory generation method uses a finite state machine based on the primary phases of the gait where within the phases the trajectory is generated to allow the gait to progress. A method of swing phase trajectory generation is presented using a finite state machine that develops joint trajectories to actively produce a desired foot motion. Also presented are two theoretical extensions to the trajectory generation method; one of which presents a nonlinear optimization for a near-optimal trajectory using a cost function that produces a gait similar to that of an unaffected subject and the other is a model free technique for planning step parameters that control to a stable gait at a desired velocity.

The controller was shown to be intuitive to new users in a multi-subject pilot study that resulted in 6 of the 7 subjects walking unassisted using lofstrand crutches in a single two hour study. Testing results also shows initial information regarding the potential learning curve for subjects showing that they are capable of developing a mastery of the device. Initial results of energy expenditure of this device compared to conventional ambulation are presented that indicate a 50% reduction when using eLEGS. Based on these results, this robust control structure and novel trajectory generation method lead to the first comprehensive and realistic mobility option for SCI patients.

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#### **1** INTRODUCTION

Currently 2% of the United States population, which is 5.5 million people, lives with some sort of paresis [1]. There are many different causes of paresis within this population including stroke, multiple sclerosis and spinal cord injuries. The patients with a spinal cord injury as their cause of paralysis is estimated to be nearly 1.3 million people, second only to stroke at 1.6 million [1]. Of those, an estimated 250 thousand patients are recovering from a traumatic spinal cord injury (SCI) with nearly 12,000 new cases per year [2]. Similarly, significant instances of traumatic SCI are present in countries other than the United States as well [3].

The quality of life for SCI patients is significantly affected as a result of their injury. Mobility is typically impaired as a result of the injury, and these patients may require extended gait rehabilitation or their injuries may lead to complete loss of lower extremity function depending on the severity of the injury. Many of these patients are left with a wheelchair as their primary mode of mobility [4]. As a result of the limited mobility and prolonged wheel chair use, these subjects are predisposed to pressure sores, osteoporosis, deteriorating urinary tract and bowel function as well poor circulation. To combat these issues, SCI patients use more prescription drugs than the unaffected population leading to a predisposition towards renal failure [5] which is a leading cause of death among this patient population [2].

Standing is one alternative used by patients to address these issues. Studies indicate that standing can have significant benefits in the overall quality of life for these patients [6]. Studies have also demonstrated the benefits provided by regular standing exercise such as increased bone health [7, 8], bowel health [6, 9], improved circulation [6] and improved renal function [10]. Despite these benefits, the majority of SCI patients who use a wheelchair for their primary mode of mobility do not stand regularly. One analysis of the patient population found that 70% of the surveyed SCI subjects did not stand for a number of different reasons including time restraints and the need for assistance [6]. The conventional method for stationary mechanical assisted standing is a standing frame.

To provide the benefits of being upright and load bearing through the skeleton in a more functional manner, other mechanical assistive devices emerged that allowed users to be mobile. The conventional devices used for mobility assistance are a reciprocating gait orthosis (RGO) which mechanically coordinates lower limb motion to torso motion and a knee ankle foot orthosis (KAFO) which holds the legs straight for load bearing as shown in Figure 1.1. Typically orthotic devices such as this are not readily accepted by SCI patients for anything more than occasional standing exercise due to the high energy expenditure requirements they present [11].



**Figure 1.1 – Conventional Mechanical Walking Orthoses** (A) Reciprocating Gait Orthosis (RGO), (B) Knee Ankle Foot Orthosis (KAFO)

Because so few SCI patients are regularly upright, the need exists for a device that makes vertical ambulation a more realistic option for practical use. This dissertation presents just such a device that applies the recent advances in exoskeleton technology to meet the practical needs of the mobile SCI patient. More specifically, this dissertation focuses on the development of a control strategy centered on desired lower extremity positions and methods for developing those desired trajectories in order to develop the intended motion.

### 1.1 PRIOR ART

The research for active exoskeletons in medical applications has a long history. Some of the initial research into legged locomotion conducted in Yugoslavia was completed on an active medical exoskeleton for rehabilitation of patients with lower extremity paresis [12]. While the research led to significant strides in the understanding and theory behind the stability of bipedal legged locomotion [13], it did not lead to significant practical application. The research led to the development of an active rehabilitation device that was tested at a United States rehabilitation hospital; however, it was not used in any other fashion [14]. This initial research was significantly limited by the sensing and design limitations of the time which resulted in a cumbersome, tethered design and a control method that required extensive feedback information. Furthermore, the control strategy was focused on developing a balanced stable gait and not on addressing the variation present in a mobility application.

The next phase of the research came a number of years later and focused primarily in the area of active anthropomorphic exoskeletons for able bodied users. A number of independent projects developed near the same time for the purpose of strength augmentation or assistance. One of these projects was BLEEX from the University of California, Berkeley [15] which is an untethered strength augmentation exoskeleton for the military that was developed as part of a DARPA project. Another device developed under the same DARPA project was introduced by Sarcos of Salt Lake City, Utah [16] which was developed for the same application. The control of these devices was done primarily to augment the users strength by reducing interaction forces between the device and user [17]. Development of these technologies has continued and served as the backbone of devices that are currently working towards commercialization. The Bleex device grew into the Human Universal Load Carrier (HULC) now developed by Berkeley Bionics in association with Lockheed Martin [18], and the Sarcos device became the Raytheon XOS 2 [19]. While these devices have lead to significant advancements in the state of the art for anthropomorphic actuation and sensing, they have fundamental issues that limit their use in mobility applications. Both devices operate with a force based control structure which complicates use in a mobility application because so many of the operational constraints are position based. In addition to that, the systems are mechanically designed to maintain many degrees of freedom and large ranges of motion which are not needed in this application.

Another device that emerged around this time was the hybrid assistive limb (HAL) developed by Cyberdyne in Japan [20]. The device is an actuated full body anthropomorphic exoskeleton that is primarily focused on strength assistance for commercial and home use. This device is much smaller than its military counterparts but is focused on providing a much smaller level of assistance. The device operates in a force paradigm but instead of trying to reduce reaction forces between the device and the user, it responds to muscle action to indicate the users' intent based on EMG signals [21, 22]. This device has since led to the development of HAL-5 also by Cyberdyne [20]. While the design of this device is closer aligned to the needs of the medical application, a problem with this control architecture is that it is entirely dependent on EMG signals getting to the lower extremity muscles in order to function. However, for SCI patients with a severe injury, the integrity of these bioelectric signals could be compromised, thus degrading the overall functionality of the device.

While these strength augmenting or assistive mobile exoskeletons have significant limitations into the arena of severe SCI patients, other research has been conducted that focuses entirely on augmenting impaired gaits. This line of research is in the area of treadmill based exoskeletons for rehabilitation of bipedal walking. Some of these devices have progressed toward commercialization such as the Lokomat which is produced by Hocoma and shown in Figure 1.2 [23]. This device originated from research conducted at University Hospital Balgirst in Zurich, Switzerland. The research led to a device with active hip and knee joints and has a control system based on gait pattern generator along with active control of the treadmill speed [24, 25]. The primary aspect of this control paradigm is that the pattern generator is primarily interested in generating a gait similar to that of an unimpaired user. Thus, the control architecture is not focused on allowing the user to maintain a balanced gait and when transferred to a mobile platform, would not guarantee a user is capable of ambulating without falling. Another commercialized stationary gait rehabilitation device is the Auto-Ambulator developed for proprietary use by Health South Corporation. However, little is known about this device as little has been published regarding its function [26, 27].



Figure 1.2 – Treadmill Based Rehabilitation Exoskeletons (A) Lokomat by Hocoma [23], (B) LOPES from the University of Twente [28], (C) ALEX from the University of Delaware [29]

Other similar devices emerged that have since furthered the capabilities of this type of immobile driven gait rehabilitation as shown in Figure 1.2. One device is LOPES that is developed at the University of Twente in the Netherlands which has explored the actuation and control of more degrees of freedom including an active pelvis [30, 31]. Another device is ALEX that is developed at the University of Delaware and focused on advanced control methods providing variable levels of assistance and force feedback to the user while still maintaining the desired positional constraints in order to create the intended user gait [32, 33]. Much like the commercial devices, a limitation of the application of these devices to a mobile application lies in the difference in safety requirements that are unique to the mobile use.

In recent years, some devices have begun to address the problems associated with being mobile. While many of these projects have focused on proximal joints such as active ankle [34-38] or knee [39] stabilization, only a few have produced mobile devices for the lower extremities specific to SCI patients with severe injuries. One of these devices is REX, as developed by REX Bionics, which is a fully balancing, actuated lower extremity exoskeleton that allows the user to ride [40]. The control of this device is focused entirely on maintaining stability while performing a set collection of maneuvers such as sitting and standing with a joystick for user input. While no information is available regarding the specifics of this device's design or control, it appears to produce a gait consistent with control strategies of robotic walking devices such as Honda's ASIMO [41]. The gait of this device is a great step forward in the area of balance in the presence of a disturbance of the subject's interaction with the machine, but it is significantly limited in the area of its real world application. Because the control is attempting to actively balance, it must create a sufficiently conservative gait to account for very likely uncertainties presented by the system model in the form of the user and its estimation of the environment. The focus on stability leads to a compromised gait that is

extremely slow and unnatural compared to an unimpaired gait and has significant theoretical limitation toward addressing these issues.



**Figure 1.3 – Current Mobility Exoskeletons** (A) REX Exoskeleton by REX Bionics [40], (B) ReWalk Exoskeleton by Argo Medical Technologies [42]

Another mobile lower extremity exoskeleton is ReWalk developed by Argo Medical Technologies which uses a different strategy to address the uncertainty inherent to the mobility application [42]. ReWalk does not attempt to fully balance the user during the gait. Instead, users employ lofstrad crutches to maintain the overall balance of the device while the device progresses the foot [43]. This approach is significantly more robust than that of REX because by giving the user an input through the crutches, the device can benefit from the user's superior ability to process and adapt for variations in the surrounding environment or to disturbances in the balance. Little information is known about the specifics of the device's design or control because it is primarily a commercial device. Despite this, some significant and limiting factors have been apparent in the control of this device that will impair its ability for use in mobility scenarios. One factor is the abnormal gait utilized by the device. This gait may be due to power limitations at the actuation or as an attempt to minimize the disturbances to the stability by significantly limiting joint range of motion, but it leads to increased upper extremity use to maintain body posture throughout the gait. The other limiting factor is the ability of the device to be learned by new subjects. This device has been in development and early subject testing at multiple rehabilitation centers, and very few advanced users have been demonstrated over this extended period of time. If patients cannot quickly learn to use the device and then continue to grow in their mastery of the hardware, they will not be able to safely utilize it for mobility in actual environments.

	Mobile /	Disabled	Assistance	Application	# DOF's	Control	Special Notes
	Tethered	Users	Goal		Controlled	Paradigm	
Lokomat	N / Y	Y	Lower Extremity	Rehab	4	Position	Treadmill based w/ body weight support
LOPES	N / Y	Y	Lower Extremity	Rehab Research	8	Impedance	Treadmill based w/ body weight support
ALEX	N / Y	Y	Lower Extremity	Rehab Research	4	Impedance	Treadmill based w/ body weight support
Raytheon XOS 2	Y / Y	Ν	Full Body	Military, Augment Strength	30 +	Force	Hydraulic Actuation
HULC	Y / N	Ν	Lower Extremity	Military, Augment Strength	4	Force	Hydraulic Actuation
HAL 5	Y / N	Y (Excluding Complete Paraplegics)	Full Body (Available in Lower Extremity)	Commercial, Strength Assist	8	Force	EMG triggered control that acts to augment requested muscle action
ReWalk	Y / N	Y	Lower Extremity	Mobility	4	Position	Patient uses crutches to balance, Unnatural gait kinematics
REX	Y / N	Y	Lower Extremity	Mobility	8	Position	Fully Balanced, Statically Stable Gait

Table 1.1 – Prior Art Summary of Leading Exoskeleton Devices

### **1.2 PROJECT DESCRIPTION**

This project will present the control strategy for a new mobile medical exoskeleton primarily for SCI mobility. The specific device used in the development of this controller is the eLEGS (Exoskeleton Lower Extremity Gait System) which is developed by Berkeley Bionics. The device will utilize lofstrand crutches for the overall balance of the device. This work will focus on the method for determining the desired joint angle trajectories that produce the intended mobility maneuvers. The methods used must be robust to the variation experienced in normal walking due to terrain irregularities or operation of the device. The overall control architecture, as well as the methods for trajectory generation presented, will be limited to the sensing and actuation available on the actual hardware which is discussed in detail in a later section.

The controller will be designed to meet a few specific goals while allowing the user to be mobile while using lofstrand crutches for balance. First, the controller must be structured such that it is applicable to a wide range of maneuvers, environments and control methods. The strategy employed must allow for adding to the available maneuvers to accommodate new needs without significantly limiting previous functionality. Second, the trajectories generated for walking must be similar to that of an unaffected gait. For many SCI subjects, there is potential to regain some muscle control, so the device must generate a healthy gait in order to provide a driven gait style of rehabilitation. Third, the control output must be intuitive and repeatable. This will allow users to quickly learn the function of the device and then continue to improve in their abilities as they adapt to the repeatable behavior of eLEGS.

This work presents an innovative and simple control architecture for SCI mobility exoskeletons. Also included is a novel method of trajectory generation for a near healthy gait that can actively determine the desired joint angles throughout the gait. Initial subject testing with the device on a subject pool is used to quantify the behavior of the device and the results of its interaction with the subject.

## 1.3 SCI PRIMER

Spinal cord injuries are defined by an injury to the spinal cord that results in the impairment of communication between the brain and another part of the body due to an injury to the nerves in the spinal column. These injuries are classified based on the level of injury, which is the last remaining unaffected nerve root in the spinal cord. The nerve roots are split into four sections which are the cervical (C), thoracic (T), lumbar (L) and sacral (S) as shown in Figure 1.4. The individual nerve segments correspond to specific levels of function in the body with the most notable for this application being that injuries in the cervical region correspond to upper extremity and upper torso control [44]. Injuries to the cervical region are classified as tetraplegics, while injuries to the remaining regions are classified as paraplegics [4]. This application is intended primarily for paraplegics and some incomplete tetraplegics because sufficient upper extremity function is needed to operate the lofstrand crutches.



Figure 1.4 – Spinal Column Nerve Roots and Injury Levels

Depending on the individual's injury level and severity, a wide range of symptoms are possible due to a spinal cord injury. The severity of the injury is classified based on the American Spinal Injury Association (ASIA) Impairment Scale or AIS, which gives a letter grade classifying the amount of function found in the impaired regions of the lower extremities. The scale operates as below [45]:

A – Complete Injury: No motor or sensory function preserved at the S5 sacral segment

B – Incomplete Injury: Only sensory function preserved at the S5 sacral segment

C – Incomplete Injury: Motor function preserved at the S5 sacral segment and less than half of the major muscle regions are strong enough to be actively controlled enough to overcome gravity

D – Incomplete Injury: Motor function preserved at the S5 sacral segment and more than half of the major muscle regions are strong enough to be actively controlled enough to overcome gravity

### E – Normal: Normal motor and sensory function

For example, a patient with an injury where the third thoracic segment is the last unaffected nerve root who has complete sensory function and trace motor function would be classified as a T3 AIS C paraplegic with an incomplete injury. The estimated population distribution of subjects within the ranges of injury level and completeness are presented in Table 1.2 [2].

		Complete	Incomplete	
	Tetraplegic	20.4 %	30.1 %	
	Paraplegic	25.6 %	18.5 %	

#### Table 1.2 – SCI Population Injury Level and Completeness Distributions

### 1.4 **BIOMECHANICS PRIMER**

In this work, standard biomechanic terminology and concepts will be discussed that the reader must have an understanding of in order to follow the presented controller design. When referencing the motion of the human body, the body is defined in three independent planes as shown in Figure 1.5. These are the coronal plane which splits the body into front and back segments, the sagittal plane which splits the body into right and left segments, and the transverse plane which splits the body into top and bottom segments. The angular motion of the joints are defined with flexion and extension as movements relative to the sagittal plane where flexion is a motion that decreases the angle between body segments and extension increases the angle. Similarly, abduction and adduction are motions relative to the coronal plane where abduction moves the body segments away from the coronal plane while adduction moves the segments toward it [46]. The primary direction of motion considered during walking is in the sagittal plane, and this is the only direction with active actuation in this device.



Figure 1.5 – Human Body Planes of Motion

A repetitive walking gait for a healthy, unaffected human subject is a very well studied and understood concept. The sagittal plane angles developed during the gait have been evaluated by many different sources; however, while the joint angles vary between individual users, a clear pattern is evident. This can be seen by the sagittal plane hip and knee joint angles developed as shown in Figure 1.6. This normal pattern is delineated into two primary phases of walking, stance and swing as seen in Figure 1.7. The entire stance phase for a single leg typically covers the first 60% of the gait cycle. Also apparent is that the gait has two double stance phases that take 10% of the cycle that are characterized by both legs being on the ground at the same time [50].



Figure 1.7 – Unaffected Walking Gait Phases

## **2** System Description

The device used in this system is an anthropomorphic under actuated lower extremity exoskeleton. In this particular application, the system control strategies that can be implemented are significantly tied to the behavior of the device as well as the available sensor information. If a control strategy is developed that relies on a certain sensor feedback or an expected mechanical stiffness at a joint that is not realizable, then the control can potentially lead to instability. For this reason, it is essential that the performance capabilities of the device are used as the backbone of the control strategies presented in the later sections of this dissertation.

This device is a mechatronic device that has a subject strapped directly in the middle of the system through backpack style torso strapping as seen in Figure 2.1. The device is designed so that it can be easily adjusted to fit a range of users between 5'2" and 6'4" and up to 220 lbs. For this reason, there are a few aspects of the design that require closer inspection. The first area is the mechanical design including the actuation and ranges of motion provided in the system. The second area is the strapping used to secure the user to the device. The final section of the system that requires closer inspection is the available sensor feedback.



Figure 2.1 – Overall view of patient in exoskeleton

## 2.1 MECHANICAL DESIGN

The device is completely mobile and has all power, actuation and processing on board. It is electrically powered and has four electric actuation units to control the hip and knee joints of both legs in the sagittal plane. The actuators at each of these joints have electric motors and use ballscrew and a linkage to transmit torque to the joint. The only actuated degrees of freedom are the flexion and extension directions for the hips and knees. All remaining degrees of freedom are either locked out or passively sprung. Every joint on the machine that allows for motion, whether actuated or passive, includes mechanical hardstops at the extents of the ranges of motion to protect the user in the event of a failure. At the hip joint, as shown in Figure 2.2, the range of motion allowed in the sagittal plane is -20° to 135° flexion. This range was selected to allow for effective walking as well as sitting. The actuation unit for this joint is located just above the axis of rotation and fixed to the torso. Based on the actuation provided, this joint has a peak speed of 14.9 rad/s [856°/s] and a peak torque of 82 Nm [60.6 ft lb]. In the abduction/adduction direction, the joint is normally locked but has a manual release mechanism to allow the joint to freely abduct through a range of motion of 0° to 55° abduction as seen in Figure 2.2. The primary use for the free abduction is for donning and doffing of the device. The hip also allows for rotation in the transverse plane along the axis of the thigh segment as shown in Figure 2.2. This joint is passively sprung through the range of  $\pm$ 45° from the neutral position that holds the foot parallel to the sagittal plane and a large detent torque at the neutral position. This degree of freedom is included for the benefit of turning while in a standing position and then straightening the foot while in swing.



Figure 2.2 – Hip joint degrees of freedom (A) Sagittal plane actuated flexion/extension DOF, (B) Manual adduction/abduction lock DOF and passive rotation DOF in transverse plane

The actuation units used to actuate the hips, and similarly the knees, use a brushless DC electric motor. They were selected to provide the necessary power requirements at the joint to meet normal walking requirements and for standing and going up stairs based on standard biomechanic data similar to that of previous research in this area [51]. The gear ratio between the motor and the joint averages approximately 80:1 throughout the normal operating range of the joints. The transmission used to provide this large gear ratio is a ballscrew fixed directly to the motor, with the nut of the motor fixed to a linkage about the joint to transmit the linear motion of the ballscrew nut into rotational motion at the joint. The specifics of this linkage and the geometry of the actuator are selected to maintain a reasonable lever arm at the joint throughout the entire range of motion of the joint. These linkages differ greatly between the actuation units used at the hip and knee joints because of differences in the ranges of motion and the locations in those ranges that require the peak torque.

At the knee joint, as shown in Figure 2.3, the range of motion allowed in the sagittal plane is 0° to 120° flexion. Similar to the hip, this range of motion was selected to allow for effective walking as well as moving from sitting to standing. Based on the actuation provided, this joint has a peak speed of 15.6 rad/s [896 °/s] and a peak torque of 83 Nm [113.1 ft lb]. The actuation unit for this joint is located above the knee axis of rotation fixed to the thigh. All other degrees of freedom at this joint are locked out.



**Figure 2.3 – Knee joint degrees of freedom** Only a sagittal plane actuated flexion/extension DOF

The ankle joint has no active actuation but is passively sprung through the range of -10° to 20° dorsiflexion as seen in Figure 2.4. The joint is center sprung to a neutral position of 5° dorsiflexion. The torque provided at this joint is much less than can be generated by a healthy human ankle [50]; however, the primary peak torques in an unimpaired gait are due to active push-off in the late stance phase which are not supported by the passive spring assembly. The design decision to not provide active actuation at this joint was made primarily because except for push-off, the passive joint can provide all functions of an unimpaired gait. All other degrees of freedom at this joint are locked out.



Figure 2.4 – Ankle Joint degrees of freedom Only a sagittal plane passive DOF

There are no other mechanically provided ranges of motion in the device although there are some locations where motion happens that affects the behavior of the device. One such area is the sole of the foot. The sole of the foot is a carbon fiber plate that is designed to allow for flexing near the ball of the foot on a standard user. The primary functionality of this motion is to allow the user to maintain a large area of contact with the ground throughout the entire step, including late stance where the heel of the foot can lift off the ground. By allowing the foot to flex, the subject's forefoot maintains ground contact as opposed to progressing onto the tip of the toe as would happen in a stiff foot option, thereby reducing the likelihood of transverse plane rotation about the toe. The other primary source of motion in the device that is not encompassed in the mechanical degrees of freedom is the motion provided to the user's upper torso by the strapping. That motion has the ability to significantly affect the behavior of the device but is allowed because it is a portion of the body that the user has sufficient control over.

## 2.2 BRACING AND ADJUSTMENTS

The device provides three primary adjustments to fit the device to a given user within the range of allowable users as seen in Figure 2.5. The first of these adjustments is the length of the thigh which can be adjusted by loosening a clamp on the thigh of the exoskeleton just below the hip joint. The next adjustment is used to adjust the length of the shank and is located just above the ankle and below the shin bracing. The final adjustment is the hip width, which allows the coronal plane width between the two hip joints to be adjusted to match the user. All of these adjustments are used to align the device's joints with that of the user to ensure a better coupling of the user to the machine throughout the entire range of motion.



**Figure 2.5 – Exoskeleton bracing and adjustments** *Hip width adjustment not shown, located between hip actuators at bottom of torso.* 

The strapping provided by the system is intended to secure the user's lower extremities to the legs of the system and maintain this connection throughout the entire range of motion of the device. The strapping is not attempting to lift or suspend the user; instead, it aims to control the position of the user's body and allow the loads due to the user's weight to be directed to the ground through the feet of the user. Another key facet of this strapping is that it allows for unassisted donning and doffing of the device by the user. This entails all torso strapping coming into the user from behind, allowing the user to transfer into the seated machine and then strap in, with all lower extremity strapping attaching to the user from the side so the user can then pull the legs of the device into theirs as seen in Figure 2.5.

At the torso, there is a harness that attaches to the user's back with backpack style shoulder straps and includes a corset style abdomen strap to synch the body into the machine. This torso strapping is primarily to control the motion of the subject's torso relative to the device torso in the transverse plane. The shoulder straps do not bear any vertical load, and the torso corset is not tightened enough to allow for significant vertical loading of the device. The other significant torso padding is the sacral pad. This pad prevents the torso of the user from slipping out of the back of the device due to a sagittal pelvic tilt, which in early prototypes was a primary source of rubbing and chafing for the user.

The legs incorporate a high thigh strap that is rigid on the front of the leg and fabric on the back. The strap is fixed to the machine above the thigh adjustment to dictate the distance of the strap relative to the torso and not relative to the knee. The purpose of the strap is to hold the user's pelvis into the device when the hip is bent and the torso wants to slide out of the machine parallel to the sacral pad. If the strap was fixed relative to the knee, taller users would have a large area between the strap and the torso allowing for the small amount of deflection provided by the fabric padding to result in large misalignments which lead to binding and rubbing on the user.

Another leg brace that is essential to coupling the user to the machine is the shin brace located on the shank at a fixed distance relative to the knee. This brace typically takes a good deal of load while users are in the machine because it is the primary source of resistance to the user's knees buckling when loaded. To accommodate these forces, the bracing has a large contact area with the skin of the user to reduce the resulting contact pressure. The final strapping provided by the device is the foot binding. There are no large loads imparted on the user by the bindings during normal use. Their primary use is to control the transverse plane positioning of the foot relative to the exoskeleton ankle. Because the user does not experience significant loading in these directions, this strap can be addressed with a number of different methods including a flexible plastic shoe binding or a simple fabric strap.

#### 2.3 ELECTRONICS AND SENSORS

The majority of the electronics of the system reside on the back of the device. The system is run using a Texas Instruments DSP processor running at 100 mhz that is used to do all the analog sensor processing, as well as the control code execution and communication including that done with the motor drives. In addition to the DSP, the custom built electronics board also incorporates a FPGA running at 25 mhz that is used to process the digital sensor inputs and pass the information on to the processor. One of the primary benefits provided by

this architecture is that the FPGA runs significantly faster than the control cycle on the DSP and thus is able to generate a much more accurate calculation of velocities coming from the digital encoders than if the velocity was calculated on the DSP. This high accuracy velocity measurement makes much more feasible the application of controller damping without the standard issues of large noise disturbances.

On the whole, this system has very few sensors considering the large number of degrees of freedom provided to the user. The first sensors in the system are analog pressure sensors located in the feet. These sensors are custom designed and built, and they incorporate a force resistive fabric stack-up to measure heel and toe pressures on each foot. These sensors are designed to measure the load through the foot onto the ground and thus include loading that comes through the foot due to loading in the exoskeleton itself. The signal provided by this sensor is thus related to the ground reaction force for the entire system as opposed to only the downward forces of the user onto the foot, so no sensing is available to isolate loading through the user and through the device.

To keep the sensors simple, some design compromises were made that affect the usefulness of these sensors. First, because the sensor is measuring two large pressure areas at the toe and heel of the foot, the sensor response varies significantly based on the loading conditions. Because the sensor relies on its ability to strain when loaded to change resistance, as the primary load path moves towards the edge or off the sensing area, the response should be expected to vary. Thus, these sensors present the potential to not respond to some extreme loading patterns that, while not typical, are possible. Thus the sensors are primarily used as a software trigger with the understanding that if triggered the foot is guaranteed to be on the ground while if not triggered it may be in slight contact with the ground still. The next issue with the use of these sensors is that the sensor information returned is not equivalent to the ground reaction force of the system; instead, it is only related. While attempts have been made to model the system loading and generate a frequency domain transfer function between load and sensor output [52], the practical use of these types of sensors does not allow it. In application, the loading conditions are not constant through the walking cycle as the center of pressure on the foot is moving significantly. Thus, while a transfer function can be assumed to exist based on a simple spring damper foot model for any specific foot placement and loading condition, there is no reason to expect that the acquired transfer function will apply to all conditions encountered by the sensor so an estimated force could not be trusted.

The next set of sensors are used to measure joint angle of the four actuated joints. Each actuation unit includes an analog potentiometer measuring the travel of the ballscrew nut throughout the entire range of motion, as well as a 2048 count optical encoder fixed to the motor. While both of these sensors measure the linear travel of the ballscrew, the mechanical linkage provides a well defined geometric relationship between the linear travel of the nut on the ballscrew and the joint angle. In this case, the high tolerances of the linkage members as well as their significant stiffness allows for a geometric relationship to be sufficient as opposed to a transfer function. By using the redundant sensing provided, the joint angle can operate without any homing procedure by using the potentiometer to estimate absolute position and

then the encoder to make a better estimation of local relative motion. This means that the system can have the performance of an absolute encoder while only having to package a standard encoder in the mechanical design. Also, by basing all joint measurements off the ballscrew travel and placing a moderate resolution encoder on the motor side of the gear ratio created by the linkage, very high joint angle resolution can be obtained at the joint. For example, at the hip joint a resolution of nearly 1000 encoder counts per joint degree exists, which results in just over 153 thousand encoder counts in the 155° range of motion.

The only other sensors used in the primary machine control is an analog two-axis accelerometer and gyroscope which are used to measure the orientation of the user's torso relative to gravity. The two pieces of sensor information are combined using a Kalman filter to result in the angle of the torso in the sagittal plane relative to gravity. This sensor package has a relatively low bandwidth limit as specified by the manufacturer of 30 hz. The bandwidth has come into play when evaluating some high level control strategies such as active balancing. The system also has access to some optional pieces of sensor information but these are currently used only for high level human machine interface (HMI) applications and not in the active control. These sensors are accelerometers and gyroscopes on each arm of the user for an absolute angle measurement of the upper arm as well as analog forces of loading in each crutch.

## **3** System Dynamics

One of the major factors that has driven the overall architecture of the control algorithms developed for this device is the system dynamics. The dynamics of bipedal walking has been extensively evaluated in a number of resources [53, 54]. The standard methodology for evaluating the human gait is to simplify the system, in order to model only the motions in the sagittal plane. This simplification is typically made because the dynamics resulting in forward progression are the primary focus; however, some models have taken into account the three dimensional nature of the bipedal gait [55]. For the purposes of this application, the sagittal plane simplification will be made because the focus is to take forward steps. In addition to this, it is assumed that the user can handle the out of plane forces using the provided lofstrand crutches.

There are a few primary characteristics of bipedal walking that greatly affect the resulting dynamic equations. The first is that the dynamics are extremely hybrid [56]. The hybrid nature of the bipedal gait is induced by the discrete changes in support due to the feet leaving and contacting the ground. This characteristic results in a series of representative dynamic equations for the different states the system is in. The second characteristic of the system dynamics is they are nonlinear and provide no clear method for linearization. This is because the dynamics appear very similar to an inverted pendulum with varying numbers of links depending on the complexity of the model. The included simplified models are used later in this dissertation for simulations and dynamic analysis.

## 3.1 SIMPLIFIED SAGITTAL PLANE MODELS

#### 3.1.1 3 DEGREE OF FREEDOM MODEL

The simplest of the sagittal plane walking models is that of a stick legged walker as shown in Figure 3.1. This model is representative of a bipedal walker with a single link for the torso and for each leg respectively with no knees. The system has six state variables, the angles of each of the three links and their angular velocities.



Figure 3.1 – 3 Degree of Freedom Walking Model

Using Legrangian dynamics, the dynamic equations for a standard single support phase can be determined as shown in Equation 3.1. These dynamic equations are reasonably complicated and demonstrate nonlinearities even for this significantly simplified sagittal model.

$$M(\theta)\ddot{\theta} + N(\theta,\dot{\theta})\dot{\theta} + G(\theta) = \begin{bmatrix} -T_{St} \\ -T_{Sw} \\ T_{St} + T_{Sw} \end{bmatrix}$$

Equation 3.1

where

$$M(\theta) = \begin{bmatrix} m_L r_L^2 + (m_L + m_T) L_L^2 & -m_L s_L L_L \cos \Delta \theta_1 & m_T r_T L_L \cos \Delta \theta_2 \\ -m_L s_L L_L \cos \Delta \theta_1 & m_L s_L^2 & 0 \\ m_T r_T L_L \cos \Delta \theta_2 & 0 & m_T r_T^2 \end{bmatrix}$$

$$N(\theta, \dot{\theta}) = \begin{bmatrix} 0 & -m_L s_L L_L \dot{\theta}_{SW} \sin \Delta \theta_1 & m_T r_T L_L \dot{\theta}_{St} \sin \Delta \theta_2 \\ m_L s_L L_L \dot{\theta}_{St} \sin \Delta \theta_1 & 0 & 0 \\ -m_T r_T L_L \dot{\theta}_{St} \sin \Delta \theta_2 & 0 & 0 \end{bmatrix}$$

$$G(\theta) = \begin{bmatrix} (m_L r_L + (m_L + m_T) L_L) g \cos \theta_{St} \\ -m_L s_L g \cos \theta_{SW} \\ m_T r_T g \cos \theta_T \end{bmatrix}$$

$$\Delta \theta_1 = \theta_{St} - \theta_{SW}$$

$$\Delta \theta_2 = \theta_{St} - \theta_T$$

The dynamics developed above are representative of the dynamics for the torso, stance and swing links of the simplified model in a single stance phase. Thus, as the walker progresses forward and the leg links oscillate between stance and swing phases, the dynamic equations for the motion of a single link discretely change based on the foot contact conditions. This characteristic is the hybrid dynamics of the system. In this system, it is important to note, there is no double stance phase or double swing phase. Because of the conservation of angular momentum about the swing foot at heelstrike, the support of the body appears to instantly transfer from one foot to the other.

Another significant factor in the dynamics is the hybrid state resets that instantaneously modify the state dynamics as the system transfers from one state to another. In this particular system the reset is due to impact of the swing leg with the ground. The state reset is calculated using rigid body impact dynamics and the conservation of angular momentum [57, 58]. This impact model allows for an instantaneous impact and a discontinuity in angular velocity but not in the angle of any segment. In this simple three link example, three conditions must be met to compute the conservation of angular momentum at impact. Those conditions are a) the conservation of angular momentum of the entire body about the impact foot, b) of the torso link about the hip joint and c) of the original stance leg about the hip joint. Using these three conditions, the state transition reset can be determined as shown in Equation 3.2. A complete

derivation of these equations can be found in the appendix of this dissertation in Section 10. In this equation,  $\theta_1$  is the angle of the stance leg prior to impact and becomes the swing leg following heelstrike, and  $\theta_2$  corresponds to the other leg.

$$Q^{-}\begin{bmatrix} \theta_{1}^{-} \\ \dot{\theta}_{2}^{-} \\ \dot{\theta}_{T}^{-} \end{bmatrix} = Q^{+}\begin{bmatrix} \theta_{1}^{+} \\ \dot{\theta}_{2}^{+} \\ \dot{\theta}_{T}^{+} \end{bmatrix}$$

Equation 3.2

where

$$Q^{-} = \begin{bmatrix} A & -m_{L}r_{L}s_{L} & m_{T}r_{T}(L_{L}\cos(\theta_{2} - \theta_{T}) + r_{T}) \\ -m_{L}r_{L}s_{L} & 0 & 0 \\ m_{T}r_{T}L_{L}\cos(\theta_{1} - \theta_{T}) & 0 & m_{T}r_{T}^{2} \end{bmatrix}$$

$$A = (2m_{L}r_{L} + m_{T}L_{L})L_{L}\cos(\theta_{1} - \theta_{2}) + m_{T}r_{T}L_{L}\cos(\theta_{1} - \theta_{T}) - m_{L}s_{L}r_{L}$$

$$Q^{+} = \begin{bmatrix} m_{L}s_{L}(s_{L} - L_{L}\cos(\theta_{1} - \theta_{2})) & B & m_{T}r_{T}(L_{L}\cos(\theta_{2} - \theta_{T}) + r_{T}) \\ m_{L}s_{L}^{2} & -m_{L}s_{L}L_{L}\cos(\theta_{1} - \theta_{2}) & 0 \\ 0 & m_{T}r_{T}L_{L}\cos(\theta_{2} - \theta_{T}) & m_{T}r_{T}^{2} \end{bmatrix}$$

$$B = m_{L}r_{L}^{2} + (m_{L} + m_{T})L_{L}^{2} - m_{L}s_{L}L_{L}\cos(\theta_{1} - \theta_{2}) + m_{T}r_{T}L_{L}\cos(\theta_{2} - \theta_{T})$$

At this point all the dynamics are present to simulate the three link simplified sagittal walker. This model is a significant simplification from a complete walking model; however, it is still useful in situations where the impact dynamics of heelstrike are the significant characteristic being represented. For this reason, the three link model presented here is used in some of the verification of control developed in later sections of this dissertation.

#### 3.1.2 5 DEGREE OF FREEDOM MODEL

The next sagittal plane walking model incorporates knees into both legs to result in a five link system. The dynamics for this walking model has 10 state variables, in particular, the angles and angular velocities of the torso, two thighs and two lower legs. The dynamics for this system, despite being much longer, are obtained in a similar manner as those of the three link system using legrangian dynamics.

The largest difference that this model presents is the potential for additional hybrid states. When this model is typically used for analysis, the knee joints are restricted to be within the normal operating ranges that knees provide by not allowing them to hyper-extend beyond the angle of the thigh. These additional states lead to a walking state machine with eight total states in it and, in turn, more than just a single reset condition as this model allows 24 state transitions as shown in Figure 3.2. Just as in the three link model, the resets are handled by the concept of conservation of angular momentum. Still, if the ranges of motion condition is not added to the system because in normal operation the model should not encounter the limits of motion, the model will have the same two states as the three link model.



**Figure 3.2 – Hybrid State Machine of 5 Degree of Freedom Walking Model** The following abbreviations are included in the figure for clarity: L1 = Leg 1, L2 = Leg 2, ST = Stance, SW = Swing, Free = Free Knee, Extended = Knee at Full Extension Hard Stop

#### 3.2 COMPLETE SAGITTAL PLANE MODEL

The biggest problem with the simplified sagittal plane models presented so far is that they do not accurately represent the dynamics of the system throughout the entire step. Even by continuing to expand the model, it will still be a significant simplification of the actual system. There are a few factors that exist in the actual system that can be added to a more advanced model. The most obvious method would involve the addition of feet in order to accurately represent the double stance phases of the gait along with realistic ground loading patterns. Another is the addition of range of motion limits that are reached in normal operation of the device such as the ankle plantarflexion and dorsiflexion limits, and the hip extension limit along with the knee extension limit which would all greatly increase the number of states in the model. Another expansion of the model involves the extension to three dimensions because while the largest joint motions happen in the sagittal plane, the out of plane motions play a significant role in the behavior of the system.

The most significant factor limiting the development of a fully representative model is that in actual application, this system relies on the input of the user to achieve stability through the lofstrand crutches. Because there is no method to model the decisions a user will make with their arms to control the crutches when presented with the exoskeleton, it is impossible to simulate that behavior. In turn, the stability of the control implemented on the actual device can only be verified when tested on a wide range of users in applications and not in simulation of a complete or representative model.

## 4 CONTROL DESIGN

The purpose of this device is to create a platform that provides reliable, intuitive and natural walking for patients with impaired lower extremity function due to a spinal cord injury. All aspects of the proposed control strategy are reflected in this focus. Before laying out a control design for this particular application, it is important to identify some of the major requirements that the controller must meet.

The first of those requirements is that because the device involves a human subject, the safety of that subject must be the highest concern at all times. Given that the user is primarily walking, which is an inherently unstable maneuver, the controller must maintain a minimal risk level at all times by not introducing any control methods that may lead to harming the user. Second, the control architecture must have an expandable and adaptive structure so the control methods and insights developed in walking can be applied to other maneuvers. The next goal of the controller is to be capable of generating a natural reciprocal gait for the user. Due to the fact that many of the patients who may use this device have the potential of regaining some amount of muscle control due to the rehabilitative benefits of this type of device, it is important to make sure that they are being trained in a realistic healthy gait.

The next requirements are based on the limitations presented by this device and the complexity of the application. The device is underactuated, it has no actuated degrees of freedom outside the sagittal plane and has very limited sensing. The application itself is so complex that had either complete sensor information or complete actuation been available, it is not entirely clear what method should be used to utilize these new resources. To counter this, the control presented here will utilize the upper extremity actuation of the user to control the overall balance of the device. The adaptability of the human body in the area of walking is unrivaled. By allowing the user to maintain a source of input through their arms, the controller can take advantage of the users superior planning and adaptability.

With this in mind, the final requirement is that the control outputs must be repeatable and reliable so the user can know what to expect from the device, and pilot it in a way that they can use it effectively. As presented, this application has the strange property where the controller will never be capable of guaranteeing the successful completion of the large scale goal of stable walking. The simplest examples are a user choosing to actively knock themselves over or lift their feet off the ground with their forearm crutches. Because, of this the controller can only strive to perform classical closed loop control on the behavior of the individual joint and then have them to behave in a way that the user can learn to work with.

### 4.1 CONTROL ARCHITECTURE

The basic control structure was selected to account for the interaction of the human with the system. Multiple stages of control are used to accomplish the different tasks presented to the controller, as seen in Figure 4.1. This structure allows the deliverable of each

stage of control to be well defined so that the controller is extremely flexible for application to new maneuvers and clear when evaluating functionality.



Figure 4.1 – Overall Controller Architecture

The first stage of control is the Human Machine Interface, which will be referred to as HMI from this point forward. This stage of control is specifically tasked with determining the intended maneuver of the user based on the provided inputs. For example, the HMI is responsible for identifying when the user intends to transition from sitting to standing and then to walking. The HMI is generally viewed as a discrete stage of control due to the fact that the intent classification has a discrete allowable domain depending on what maneuvers are supported by the system. More information about the function of the HMI methods used in this device are provided later in this section.

The second stage is the trajectory generator, which based on the intended maneuver as reported by the HMI along with the current sensor feedback from the device determines what the device should do to accomplish the intended maneuver. For this machine, the output of this stage of control is a set of desired joint angles for each of the controlled joints of the system. It is also possible for the trajectory generator to result in a set of desired joint torques but in this machine the ability to control to a desired torque is not possible due to the lack of sensors. This is the first stage of control that operates continuously and provides most of the design opportunities for the continuous control because it does not fit well into many classical control methods. This is the section of the controller that is the primary research focus of this document and more information regarding how this is handled is provided below.

The final stage of control is the low level controller, which generates the current command for the individual joints to reach the desired motion resulting from the trajectory generation. This stage is a more classical control method as it includes the closed loop tracking of a desired joint angle by adjusting the commanded current to the motor at the joint. This controller can easily be modeled as a continuous time single input single output (SISO) controller where other factors are viewed as disturbances. More information regarding this stage of the controller is available later in the section.

#### 4.2 HUMAN MACHINE INTERFACE

The HMI attempts to convert external inputs either through the HMI sensors discussed in Section 2.3 of this dissertation or a Graphical User Interface (GUI) to select the intended maneuver for the user. To accommodate the discrete dynamic conditions present during bipedal walking, the HMI uses a finite state machine to indicate the intended option of a series of maneuvers. The high level maneuvers that the HMI can move between are currently as follows: Seated, Stand Up, Standing, Walking, Turning and Sit Down. The high level of the state machine is shown in Figure 4.2.



Figure 4.2 – HMI Finite State Machine for Available Maneuvers

In some cases the HMI also behaves to move through the phases of a specific maneuver. The most prevalent maneuver where this is the case is in walking. Here the HMI is used to indicate the user's intent to take each individual step. This is important because while some maneuvers have very definite beginnings and ends like standing up, repetitive maneuvers such as walking required continued involvement by the user for the device to know to continue the action. For example, walking is split into three different phases within the HMI which include an initial state where the subject stands still as well as right and left step phases. It is important to note, that while additional phases exist to execute the walking action, these three states are the ones that identify the intent of the user for the controller to act on. The state machine showing the phases of the walking maneuver can be seen in Figure 4.3.



Figure 4.3 – Walking Maneuver HMI State Machine
The guards in these state machines are set based on the mode of operation of the HMI. The first mode uses the GUI to allow the user or a therapist to move through maneuvers as well as the individual phases of those maneuvers. This simple mode relies entirely on input from the GUI for every transition. The advanced mode, which is based on the research of Katie Strausser, uses the input of the HMI sensors to create these guards. For example, to transition from the right foot step phase to the left foot step phase the advanced HMI looks to see that the right crutch has progressed forward through the arm angle sensor and that the crutch has been loaded. If these conditions are met than the HMI identifies that the user has moved their crutch forward and shifted weight onto that forward crutch thus intending to step forward. In addition to these guards, some foot sensor information is evaluated to identify that the feet are correctly loaded to allow for a safe step. Similar transitions are provided for in the advanced HMI throughout the finite state machine to allow for natural, robust and safe transitioning between states.

The primary benefit provided by the advanced state machine is that instead of requiring direct input by the user, it has the potential to view the entire posture and motion of the user to determine intent. In turn it allows the option to identify the user's intent by perceiving the small motions that the user makes naturally when trying to accomplish that motion. Potentially, the advanced HMI can provide a safer user experience by identifying and preventing false state triggers that could be caused even in the simple mode due to an inadvertent button press. By looking at the entire pose of the subject, the HMI can identify postures that do not match with the selected intent of the user and then ask for clarification or completely block them if they are deemed unsafe. Based on these potential rewards provided by the advanced HMI at this stage of the controller, it is still an active area of research.

## 4.3 LOW LEVEL CONTROL

The task of the low level controller is to generate the desired motor current necessary to get the individual joints to complete their desired actions. The low level controller is currently applied independently to each of the four joints resulting in a single input single output (SISO) control problem. The block diagram for how the low level controller operates is shown in Figure 4.4. It is worth noting that the output of the controller is the desired motor current because, as shown in Figure 4.5, the plant dynamics include the dynamics added to the system by the motor drive. The motor drive has an internal PI controller that is used to track the desired motor current and supply it to the motors.



Figure 4.4 – Block Diagram for Arbitrary SISO Low Level Controller



Figure 4.5 – Detail of Plant Dynamics in Low Level Controller

One freedom that this low level control structure provides is to set up many different low level controllers that can be implemented at a specific joint. The two components that can be manipulated are which state variable to track and which controller to implement. At the hip joint for example, the low level controller can close the loop on the relative joint angle between the thigh and the torso as well as the absolute torso or absolute thigh angles. This structure also allows for the tracking of a desired joint torque in the future when additional sensors are added to provide feedback on the actual joint torque. The controller typically implemented is a version of a PID controller where depending on where it is applied the gains are scheduled to provide the desired feel to the user. The scheduled gains are used to account for significant changes in system dynamics such as a leg moving from stance to swing or to get different behavior such as the stiff control desired while standing still as compared to the smooth soft control provided through the gait.

The performance of this low level controller has been adequate for the requirements of the application. In standard operating conditions, these controllers can be tuned to maintain tracking errors of around ±3 degrees. A representative case is shown in Figure 4.6, where the tracking error of the knee joint to a sinusoidal desired angle using a stiff PD controller with the foot off the ground.



Figure 4.6 – Low Level Controller Tracking Results for Knee Angle

It is important to note that while these controllers can be more aggressively tuned in order to minimize the tracking error, no realistic improvement in performance would be achieved for the actual system. The goal of the controller is to move the subject's leg to the desired joint angles, and a number of sources of error lie between the controlled exoskeleton joint and the user's joint. Error from the coupling of the user's leg to the device leg and sagittal plane flexing of the exoskeleton leg due to the limited mechanical stiffness are the primary contributors. On the actual system these sources of error can easily result in errors of at least  $\pm 2$  degrees. In turn, better tracking would do little to improve the behavior because there is no sensing available of these errors that are after the device joint.

If in the future improvements to the low level controller were desired due to tighter required specifications or a new maneuver which demonstrates degraded performance from the current implementation, two improvements could be applied. The first improvement takes advantage of the fact that while the current technique does not require a model to gain tracking, some information is known about the required performance of the joint that can allow for gravity compensation. Gravity compensation is not a new concept as the benefits of it have been explored in previous research [59, 60] and have also been implemented in other bipedal gait devices [60, 61]. The primary disturbances that this joint must overcome on the swing leg are the result of gravity on the mass of the machine and users leg. By adding this simple model into the system using positive feedback, as shown in Figure 4.7, the low level controller is only responsible for overcoming the error between the actual load needed and the modeled load unlike before where it was responsible for everything at the joint. Because the gravity compensation is implemented through positive feedback, the closed loop dynamics of the resulting plant with the compensation will change and in turn may require the gains of the low level controller to be tuned again to provide stability.



Figure 4.7 – Improved Low Level Controller with Positive Feedback Gravity Compensation

The other potential improvement is to develop a model of the system and implement a multi input multi output (MIMO) controller that addressed each of the joints together as opposed to the independent behavior of the implemented controller. While the SISO system controller is rather effective, the dynamics of the system are coupled as discussed in Section 3. The most obvious location to evaluate this is the thigh, which directly is effected by the torque of both the knee and the hip. In turn, if the hip is trying to control the absolute angle of the thigh and the knee is exerting a torque on the thigh as well, the resulting torque on the thigh could balance and lead to no motion. While this result is not a significant failure, the torque on the hip will still act on the torso and most likely result in an undesirable motion for the user. A MIMO solution utilizing a model will perform better in these conditions because it would inherently account for the coupling of the exoskeleton states.

## 4.4 TRAJECTORY GENERATION

The controller levels discussed in detail up to this point accomplish the tasks of determining the user's intent and then controlling the joints to perform as requested. The trajectory generation stage of the controller is used to bridge the gap between these two stages by determining the desired joint actions based on the intended maneuver. To accomplish this task, the trajectory generation uses a finite state machine to move through the different phase of operation for the intended maneuver. By using the finite state machine to determine the phase of operation, specific trajectories can then be developed in continuous time to achieve the desired machine behavior. To explain the trajectory generation process, this section will detail the complete finite state machine and specific trajectories generated within them for the walking cycle.

The walking finite state machine is the same as that provided by the HMI with the addition of two phases to delineate the single and double stance phases of the gait, as shown in Figure 4.8. The trajectories generated are from the perspective of the right leg where the left leg is 180 degrees out of phase at all times.



Figure 4.8 – Basic Walking Finite State Machine for Trajectory Generation

## Swing Phase

The first phase the right leg enters is the Swing phase. During the swing phase, the desired motion of the leg is for the foot to pick up off the ground, progress forward and then return to the ground with a heel strike. Swing executes in half of the user selected step time.

The individual joint angle trajectories for both the hip and knee joints as a function of time are precalculated and stored in a look up table for use during the control cycle. The design of these trajectories has a significant impact on the behavior of the system while in the walking phase. One potential design method is to use the joint angle trajectories of an unimpaired gait and play them back while the subjects are walking. In application, this method does not work because of a number of differences that exist between an unimpaired gait and a gait in the exoskeleton. One of those differences is that an unimpaired gait has powered ankle flexion to aid in toe clearance while the exoskeleton ankle is sprung to its neutral position of 5 degrees dorsiflexion. The other primary difference is the small deviations that are possible from the

desired position due to low mechanical stiffness, imperfect joint angle tracking and imperfect coupling between the user and the machine. These reasons all point to the implemented trajectories requiring more toe clearance than standard healthy walking to provide some buffer room to allow for these issues to exist while still creating sufficient toe clearance for the user in the device.

The final joint trajectories that were selected are modified to provide additional toe clearance for the machine foot through the swing process. In order to provide the additional toe clearance required, the peak knee flexion increased to 76 degrees and the peak hip flexion increased to 49 degrees. These increases in the range of motion used in normal walking have a significant effect on the gait in that they limit the speed at which an individual step can be taken. In order to keep the joint angle velocities at safe levels for the joints of the user, the minimum step time most users can comfortable handle is 2 seconds which is about twice the step time of an unimpaired gait [47]. The final joint angle trajectories and their comparison to unimpaired joint angle trajectories can be seen in Figure 4.9.



Figure 4.9 – Precalculated Swing Phase Trajectories Compared to Unimpaired Gait

The only other task of this phase is to make sure that the trajectories are zero order continuous at all times. This is only an issue at the beginning of the phase where the joint angles coming into the swing phase may not match the initial angles of the precalculated joint angle trajectories. To accommodate this, the actual trajectories use a linear blend to move smoothly from the initial joint angle references entering the phase to the precalculated ones over a set period of time. The behavior of this blend can be seen in Figure 4.10 and is discussed in detail in the appendix in Section 11.

A final modification to the trajectories is the application of a user selectable desired knee angle at heel strike. The selectable value is varied between 0 and 15 degrees to make weight transfer during early double stance easier for the subject, where it gets easier as the

angle is increased. This is because by bending the leading knee in the first double stance phase of the gait it shortens the front leg of the double stance posture where the user can more easily move forward over that leg than fall back over the trailing leg. This angle is added to the knee angle with another linear blend from the peak of knee flexion to the end of the step. This behavior is also shown in Figure 4.10.



Figure 4.10 – Linear Blending of Precalculated Swing Phase Trajectories for Continuity

At the end of this walking phase the joints hold their final angles until the foot comes in contact with the ground. At that time the phase transitions to the first double stance phase of the state machine.

## 1<sup>st</sup> Double Stance

In the first double stance phase of walking, the knee needs to prepare itself to take the load of the body through the single stance phase. To do this the leg needs get to a user selected bend angle that is close enough to standing to allow the user to shift their weight onto that leg. The knee bend angle is typically set between 0 and 15 degrees. With the knee flexed, it is easier for the user to transfer their weight onto their forward leg. Typically, the knee is position controlled to track the desired joint angle throughout this entire phase, however, if a low knee bend angle is specified where the joint angle can reach within 2 degrees of full extension, then the brake will trigger and lock the joint.

At the hip, the joint stops putting out torque in order to allow the user to shift their body weight forward over the forward leg of the double stance phase. Ideally, this would be accomplished by allowing the joint to maintain a zero torque desired output, but because there is no sensing that allows for accurate torque estimation at the joint, simply zeroing the torque output of the motor was found to be a sufficient alternative. The only errors that this results in, is requiring the forward momentum of the user to overcome any friction present at the joint during this motion. The only controlled motion at the hip are the safety limits used to restrict the free hip from progressing significantly beyond the initial angle. Thus if the joint is bent to more than ±15 degrees from the initial hip angle for this double stance phase, it is position controlled to prevent a runaway situation. This is only to prevent a specific failure scenario and is not in play during normal operation of the device.

After the user has successfully transferred their weight onto the swing leg, the HMI triggers the advance to the next state. At this point the leg transitions to single stance.

## Single Stance

In single stance the leg is responsible for supporting the weight of the body as high as possible to allow for toe clearance of the swing leg and progress the body forward to allow for an acceptable step. To get the hips as high as possible the knee of the stance leg is position controlled through an extension to 0 degrees at which time the joint is locked with the use of the brake that is on the joint. The trajectory for the controlled extension is created by using a linear blend from the initial knee angle to 0 degrees over 30% of the user selected cycle time.

To progress the body forward, the hip undergoes a controlled extension to a user selected desired hip angle. This controlled extension also provides some propulsive assistance to the user during the walking cycle. The user selected hip angle is typically set between 5 and 10 degrees to provide sufficient hip extension at the end of single stance. The trajectory used for the hip is generated with the use of a linear blend from the initial single stance hip angle to the desired extension angle over the 50% of the user selected cycle time. The behavior of both of these joints through single stance can be seen in Figure 4.11.



The ending angles of this phase are held until the user's foot contacts the ground and the state machine transfers to the second double stance phase of the gait.

# 2<sup>nd</sup> Double Stance

The purpose of the second double stance phase in this gait is to allow the user to shift their weight to the front foot. To do this, the knee remains locked. In the event that the state transferred due to foot contact prior to the knee locking, the knee joint is linear blended from the initial knee angle to 0 degrees where it is subsequently locked just as it was in single stance. The hip is position controlled to hold the initial hip angle coming into this phase. This allows the user a known body posture to position while awaiting the next HMI trigger.

The base function for each of the walking phases discussed above is actually quite simple. The majority of the complexity found in this state machine involved handling the errant cases that the state machine may encounter. After adding all of these handling conditions as discussed before to the state machine, it grows to a quite active state machine as shown in Figure 4.12. The main step provided by this state machine centered approach to trajectory generation is that while it does not provide optimal control for the entire walking cycle, it provides a functioning base that allows each of the functions of a normal gait to be carried out with adequate tolerance for the many possible inputs from the user. By having this structure, the function of any one area can be independently evaluated and applied to a real user as will be done in the coming sections of this dissertation.





Synchronized trigger indicates that hip and knee advance states at same time

It is important to note that all trajectories on the device are generated in a similar manner. While only the trajectory generation state machine for walking was provided here, similar state machines exist for the standing up and sitting down maneuvers currently handled by the system. This in turn, can be applied to many other types of maneuvers that will be developed on this device such as turning, and ascending or descending stairs.

# 5 ACTIVE SWING PHASE TRAJECTORY GENERATION

The swing leg joint trajectories serve as one of the most significant aspects of the overall performance of the gait. The currently discussed method for generating these desired knee and hip position trajectories involves generating an offline calculation of the trajectories and then playing them back as a function of time through the step as detailed in Section 4. While this has proven to be a reasonably reliable method for developing these trajectories, it presents some significant issues that limit its long term application.

First, this walk generates a conservative ground clearance profile in order to guarantee that the foot reliably clears the ground throughout a range of different poses. In turn, because the trajectories create a foot profile that successfully clears the ground for a subject leaning significantly forward, it leads to excessive toe clearance for subjects who have either learned to or been coached to walk more upright. With the goal of the device being to generate a repeatable and reliable gait for a wide variety of patients, this is an undesirable result. Second, the simple trajectory generation method makes the assumption that all steps are the same. This is not the case for most steps and especially not for the extreme cases such as first and last steps which only require a half step. This results in strange steps that look unnatural when the precalculated gait trajectories are significantly different from what the joints should actually be doing.

Third, the previously presented trajectory generation technique has no clear method for modifying the gait actively. Typically during testing the therapist or user may want to adjust the step length or independently adjust the toe clearance height that the step is attempting to reach. Because both of these are sagittal plane Cartesian constraints it is not entirely clear how to reflect these desired modifications onto the joint space trajectories. Finally, when the desired joint trajectories are blended to account for initial joint positions, it is unclear what motion is created for the foot in the sagittal plane by the use of the smoothed joint space trajectories. In some cases, smooth joint space trajectories can lead to undesirable results such as the toe clearing the ground at very different heights throughout the step or in some cases not at all.

This method seeks to fill the void and develop an active trajectory generator that will calculate the desired joint trajectories throughout the step. This method will use feedback from the joint kinematics and device pose in combination with Cartesian and joint space constraints to develop joint trajectories leading to the desired foot motion. This method will be developed to improve each of the discussed issues with the previous technique.

# 5.1 ACTIVE TRAJECTORY GENERATION STRUCTURE

Throughout the gait the leg must accomplish a series of different tasks that can be isolated from the other states of the swing phase. The structure used to generate the trajectories takes advantage of this by using a finite state machine to isolate these specific

tasks. This, as with the larger trajectory generation state machine, allows the controller to focus on completing one specific task at a time. The states present in the active trajectory generation state machine are Pre-Swing, Toe Clearance, Toe Extend, Heel Extend, Heel Plant and Toe Plant and are shown in Figure 5.1. The guards for the individual states and the specific methods used to generate these trajectories are discussed later in this section of this dissertation.



Figure 5.1 – Active Swing Phase Trajectory Generation State Machine

The primary unknown throughout this calculation is location of the hips relative to the stance foot. In a simplified system where the hip joint position was either fixed or defined throughout the entire step the desired foot path for the step along with the joint trajectories could be fully defined based off the initial conditions. Because this is not the case, however, the desired joint trajectories can be recomputed every control cycle based on the current estimate of the hip location. In all states, the individual joint angles are calculated using the swing leg geometry and the current hip location estimate. As shown in Figure 5.2, for both the calculation of heel and toe locations the joint angles are completely defined when two parameters are present. This property is used to calculate the desired joint angles by varying which two parameters are provided based on which state in the state machine you are in.



Leg geometry as used in each state of the active method to calculate the desired joint references based on having 2 of the 4 constraints (D<sub>x</sub>, D<sub>y</sub>, θ<sub>Th</sub>, θ<sub>k</sub>) (A Heel Centered Geometry Diagram, (B) Toe Centered Geometry Diagram, (C) Lower Leg Geometry Close-up

Before the individual states are discussed, there are a few concepts that are important to understand. First, the goal of the active trajectory generator is to create a step of a user selected step length and clearance height. Both of these values are selectable by the user on the GUI where the clearance height is the desired number of inches of toe clearance for the step and the step length is the desired length from toe to toe in the resulting double stance phase. It is also important to note that just as in the simple trajectory generation method the step will attempt to complete in 50% of the user selected step time.

Second, this active trajectory generator is meant to replace the simple method discussed in the previous section within the overall state machine. In turn, the active method is only generating trajectories for the swing leg. The trajectories are intended to clear the ground and then reach the desired step length in the desired step time despite the behavior of the stance leg and torso of the user. Because the stance leg and torso is controlled by this input of the user, it can be expected that the trajectories will be different for each step even if all the user selectable settings are the same.

## 5.2 STATE BEHAVIOR AND DESCRIPTIONS

## Pre-Swing

The first state of the active method is the Pre-Swing state. Coming into the swing state the swing leg typically has a fully extended knee which leads to the toe of the foot being located in front of the user's thigh. This is important because in order to arbitrarily place the toe into a desired Cartesian position as is needed to complete the step, the toe must be moved through the geometric singularity caused by the thigh and the toe as shown in Figure 5.3. For this reason, the Pre-Swing phase is responsible for flexing the knee to move the toe through the singular position.



Figure 5.3 – Geometric Singularity Entering Swing Phase

This state takes advantage of one characteristic of an unimpaired gait, where the knee flexes is a sinusoidal motion up to the peak knee flexion angle. The first aspect is estimating the peak knee flexion angle for the swing leg. This estimation is made by calculating the hip to toe length needed for the toe to pass directly under the hip with the desired toe clearance because this is going to be the minimum available clearance length. The estimation is completed once at the beginning of the state. The knee angle trajectory is then calculated as a sinusoidal function from the initial angle to the peak flexion estimate with a half period of 50% of the desired swing time. At this time the hip trajectory is calculated to maintain the absolute angle between the hip and toe throughout the process. Another potential way of making this calculation would be to use the hip to maintain the Cartesian X coordinate of the toe position to minimize the relative motion between the foot and the ground. In application, however, it was evident that because no angle measurement is available for the ankle, the estimate of the toe X position in this state when the foot is contacting the ground and the ankle is not at its neutral position was so poor that it actually increased relative motion. The portion of the trajectory generated for a representative step through this state and the resulting body posture is shown in Figure 5.4 and Figure 5.5.



The guard for progressing to the next state is when the leg has progressed past the singularity position with a small dead band to ensure the toe is sufficiently beyond. The dead band used is an absolute thigh angle 10 degrees beyond the absolute hip to toe angle.

### Toe Clearance

The next state in the active swing method is the toe clearance state. This state is responsible for progressing the foot forward towards the desired plant location and clearing the ground with an allowable clearance height. This is completed by defining the desired relative X and Y coordinates for the toe relative to the hip and using the leg geometry to compute the joint angles.

In order to define the desired X coordinate, a sinusoidal function is used ranging from the current relative toe X position to the estimated ending relative X position with a predetermined period. The explicit equation for the calculation of the desired X coordinate at a specific time can be seen in Equation 5.1.

$$D_{xi} = .5(D_{x0} - D_{xMax})\cos\left(\frac{\pi}{0.95 - P_0}(P_i - P_0)\right) + .5(D_{x0} + D_{xMax})$$

Equation 5.1

where

 $D_{x0}$  Initial Relative X Coordinate for State

 $D_{xMax}$  Estimate of Maximum Relative X Coordinate for Entire Gait

*P*<sub>0</sub> Initial Cycle Percentage for State

*P<sub>i</sub>* Cycle Percentage at Time i for State

It is extremely important that the desired X position is specified relative to the hip and not in absolute coordinates. Because the X position of the hip is not known for times in the future, it is possible that the hip progresses forward faster than the desired toe coordinate resulting in rapid extensions of the knee through the beginning of this state. Alternatively, by specifying desired toe coordinates relative to the hip the length between the hip and desired toe position is bounded and smooth, eliminating the potential for errors based on out of phase motion between the hip and desired toe positions.

The toe's Y coordinate is intended to get up to the user selected clearance height and remain there. In order to get the toe up to this height in a smooth fashion, the desired toe height is blended from the initial position and velocity to the clearance height using a first order continuous blend. More information regarding the specifics of the implementation of this blend can be found in the appendix in Section 11. The blend is carried out over 15% of the swing time so that the foot can quickly get off the ground to allow for the swing phase to carry out.

Once the desired X and Y coordinates are defined, the desired joint angles are calculated using the leg geometry. The resulting joint trajectories and body posture can be seen in Figure 5.6 and Figure 5.7 respectively.



The guard present for leaving this state and progressing to toe extend is reached when the toe has progressed to some dead band in front of the hip which in this case was set to 3". The reason for progressing at this time is that after the toe passes the hip which is the maximum knee flexion location, the constraints on the motion of the system are no longer the desired X and Y coordinates of the toe. At this position in the motion the primary concern is

that the knee extends at a reasonable speed so the user can prepare for stance. So if the anticipated desired toe X resulted in a knee angle that extended extremely slow or even paused, the desired behavior would be to compromise on the step time and toe X progression in order to have a step that felt and appeared more natural to the user.

### Toe Extend

Toe extend is the next state in the active method state machine and is primarily responsible for extending the knee at a reasonable rate while maintaining the user specified clearance height with the toe. The calculation is made by determining the desired knee angle and toe Y position relative to the hip and then using leg geometry to calculate the corresponding hip joint reference angle.

The first aspect to determining the desired knee angle is making an estimate of the ending knee angle based on the current pose of the device. One aspect of this estimation is that no guess has yet been made as to whether ground contact was going to end in a heel strike or a toe strike. To accurately estimate the ending knee angle a crude estimation is made, but there would be significant benefit in knowing more accurately the ending condition of the foot. The estimation used in this implementation is that if the linear distance between the hip and the ending toe position is greater than the maximum possible heel to toe length then the ending knee angle is estimated as if heel strike will occur and otherwise it is assumed toe strike will occur. The ending knee angle is then estimated based on the leg geometry. It is important to note that even if it is estimated that the ending ground contact condition will be a heel strike, no action is taken to position the heel such that it will contact the ground. This is because this estimation is so poor and made so early in the gait that the odds of it being accurate are very slim. So while the ending knee angle is estimated differently, the state still acts to maintain the toe at the clearance height and relies on future states make the decision to change the action. This will be discussed more with the guards of this state along with the heel extend state.

The next aspect to determining the ending knee angle is to estimate the desired time the knee angle should extend over. This is present to protect against the knee angle extending too slowly or pausing as discussed previously. To accomplish this, the desired ending time was selected based on how far the knee was expected to extend between the current angle and the estimated ending knee angle. This allows the active method to compromise on the desired step time and take a slightly faster than normal step in order to maintain normal motion at the knee joint. Once both the ending time and the ending knee angle have been estimated the desired knee angle can be computed using a first order continuous blend as discussed in the appendix in Section 11.

The desired relative toe Y coordinate is selected based on the user specified clearance height and the reference hip angle is calculated based on the resulting leg geometry. The resulting joint trajectories and posture are shown in Figure 5.8 and Figure 5.9 respectively.



The guards for this state allow for three different transitions. The first is the transition to heel extend which happens when the heel is found to be approaching the user selected clearance height at a rate greater than a predetermined threshold. If this transition is triggered, the actions change to reference the heel position in preparation of an assumed heel strike contact. This transition is the first available transition that allows the system to begin preparing for a heel strike and in turn is a conservative guard. This transition only catches the very clear heel strike conditions because the decision is being made earlier in the gait cycle.

The next guards available to this state trigger when the toe progresses to within some dead band of the desired ending X coordinate, where the dead band as implemented is 3". At this time the same crude estimation is used that was for the ending knee angle. If the length from the hip to the desired ending toe location is greater the maximum hip to toe length then the system transitions to heel plant and otherwise it transitions to toe plant. This is the last point at which the active method can transition to preparing for a heel strike ground contact.

## Heel Extend

The next state in the active toe trajectory state machine is the heel extend state. This state is responsible for continuing to extend the knee while keeping the heel at or above the user selected clearance height. The function of the state is very similar to that of the toe extend state but using the heel as the primary focus. As discussed earlier, this state handles the poses that meet the conditions of a very conservative guard. Due to this, the state is typically very short with complete subjects because they generally walk leaned over farther than incomplete or unaffected counterparts. Despite this the state is extremely important to the function of the active method, because it allows the method to emulate the behavior of an unaffected gait by allowing the foot to continue to extend while maintaining proper clearance at the heel.

Because the knee angle is calculated exactly as in the toe extend, except for the ending knee angle always being calculated in reference to the heel, the primary behavioral portion of this state is establishing the desired relative Y coordinate. Entering the state the heel is above the clearance height and moving towards as dictated by the guard coming into the state. Because of this, the heel Y coordinate is set with a first order continuous blend from the initial conditions entering the state to the desired clearance height. At that point with the desired relative Y coordinate and the desired knee angle, the resulting hip joint reference angle can be calculated using the leg geometry.

The guard for this state is if the heel reaches within 3 inches of the desired heel plant x coordinate which triggers a transition to the heel plant state. The resulting joint trajectories and the posture are shown in Figure 5.10 and Figure 5.11 respectively.



### Heel Plant

The next state in the swing phase is the heel plant state. This state is responsible for extending the knee as much as it can without interfering with the ground and moving the heel to the correct step length. This is accomplished using the desired knee angle and the desired heel X coordinate relative to the hip as the constraints for the leg geometry to calculate the reference hip angle.

The knee is extended just as it was in the heel extend state. The hip is calculated to maintain a desired heel x coordinate and move it to the x coordinate needed to meet the intended step length. This is done with a first order blend from the initial heel x coordinate and x velocity to the desired ending coordinate. The resulting joint angle trajectories and posture due to the heel plant state are shown in Figure 5.12 and Figure 5.13 respectively.



The only guard for transitioning out of the state is the foot contacting the ground. At this time the overall walking state machine will transfer out of the swing phase and into the first double stance phase. In the event that the foot does not hit the ground at the intended

time, the knee joint will hold its calculated end angle and the hip will continue to maintain the desired step length.

# Toe Plant

The final state in the active swing phase trajectory generation method is the toe plant state. This state operates exactly as the help plant state, but instead of moving the heel to the desired end x coordinate it controls the toe to its desired end coordinate. This state is an interesting case because it should only execute when the user is in a posture that precludes the completion of a step with a heel strike. In unimpaired walking this is a very rare occurrence and typically only happens when a very small step is taken, however, for new users in the exoskeleton who walk with significant torso lean, this state is often encountered.

The complete active trajectory generation state machine as presented in this section along with the geometric constraints were used to calculate the remaining leg reference information are provided in Figure 5.14.



Figure 5.14 – Summary of Leg Geometry Calculation Constraints

# 5.3 ACTIVE TRAJECTORY GENERATION RESULTS

In application this method works well to generate the joint trajectories online. As Figure 5.15 shows, the trajectories generated for normal upright walking is very similar to that of the simple method. There are added benefits in this application though, as the peak hip flexion angle was reduced by 6 degrees because the active method does not need to be as conservative as the trajectories from the simple method. The real benefits of this technique come when applied to non-standard steps as shown in Figure 5.16 applied to a first step. In this application the motion was more representative of the desired motion and the peak knee flexion was cut by 13 degrees and the peak hip flexion was cut by 6 degrees.



Figure 5.15 – Active and Simple Trajectory Generation Comparison for Upright Walking



Figure 5.16 – Active and Simple Trajectory Generation Comparison for First Step

# 6 TRAJECTORY GENERATION THEORETICAL EXTENSIONS

Up to this point, the trajectory generation methods discussed have shown a wide range of application to the walking gait. Despite the current success, there are some significant limits to their capabilities. The current control cycle as shown in Figure 6.1, only has limited feedback into the trajectory generation process and even then it is only when applying the active method. In the active method, only kinematics are used for feedback to adjust for the geometric constraints of the system where the user is expected to account for the dynamic disturbances. In the simple method of trajectory generation, there is no state feedback feedback used at all as the trajectories themselves are just a feedforward input to the low level controller. Also, there is at no time any feedback on the user selected step parameters to verify that the device is meeting its expected step performance.



Figure 6.1 – Implemented Control Architecture

 $T_{\rm p}$  is the desired step time parameter and  $L_{\rm p}$  is the desired step length parameter

To correct these issues, two modifications are proposed to the current control architecture. The first is to develop a method to incorporate dynamic feedback into the trajectory generation process. With this added feedback, the trajectory generator can actively work to alleviate the burden on the user for overall balance and stability while creating a more natural gait. The second primary modification is the addition of feedback on the state parameters provided by the user to ensure that the step is being executed as desired. The modified control architecture that includes this step parameter feedback can be seen in Figure 6.2. It is important to note that these are referred to as theoretical extensions because they currently require more capability than is provided by the current device and in turn cannot be implemented on the real system as is.



**Figure 6.2 – Modified Control Architecture for Theoretical Trajectory Generation Extensions**  $V_D$  is the desired step time parameter,  $T_D$  is the step time parameter and  $L_D$  is the step length parameter

## 6.1 NEAR OPTIMAL TRAJECTORY GENERATION

In unimpaired bipedal walking the brain is able to generate effective and efficient joint angle trajectories throughout the walking cycle. These trajectories are customized for the needs of the user and work with the users individual dynamics in mind. This extension intends to present a framework for determining the near optimal gait trajectories for an exoskeleton gait based on initial conditions and a desired walking maneuver similar to those trajectories that would be generated by an unimpaired subject given the same input conditions. The method presented will use an optimization format to determine the near optimal trajectories.

### 6.1.1 PROBLEM FORMULATION

#### **Optimization Formulation**

This strategy structures the problem as a nonlinear optimization problem. Below is the optimization formulation that was initially evaluated for this problem as shown in Equation 6.1 with a final cost represented by  $g_F(\theta, u)$  and an individual state cost represented by  $g(\theta, u)$  where both are discussed in detail later in this section.

$$\min_{0 < t < T} \left[ g_F(\theta, u) + \int_0^T g(\theta, u) dt \right]$$

Equation 6.1

subject to

$$M(\theta)\ddot{\theta} + N(\dot{\theta},\theta) + G(\theta) = u$$
$$L_{St}\sin\theta_{St} - L_{Th}\sin\theta_{Th} - L_{Sh}\sin\theta_{Sh} - Foot_{Geo} \ge 0$$

where

$$L_{St} \sin \theta_{St}(T) - L_{Th} \sin \theta_{Th}(T) - L_{Sh} \sin \theta_{Sh}(T) - Foot_{Geo} = 0$$
  

$$L_{St} \cos \theta_{St}(T) + L_{Th} \cos \theta_{Th}(T) + L_{Sh} \cos \theta_{Sh}(T) + Foot_{Geo} = StepL_{Des}$$
  

$$\theta_{Th}(T) = \theta_{Sh}(T)$$
  

$$\theta(0) = \theta_{0}$$
  

$$\theta_{Min} \le \theta \le \theta_{Max}$$
  

$$u_{Min} \le u \le u_{Max}$$

The path constraints included maintain that the toe and heel of the swing foot stay above ground throughout the gait and that the foot finishes the gait on the ground and at the desired step length. The specific angles and geometry of the model used are discussed in greater detail in the system model portion of this section. Another aspect of the path constraints is that they incorporate the foot geometry in the variable  $Foot_{Geo}$ . Because the model does not incorporate an ankle joint, the toe and heel maintain a fixed relationship to the shank throughout the motion and can be added into the optimization to accurately account for the toe and heel coordinates. The specific geometry used in this aspect of the model is application specific and must be determined based on the geometry of the system it is being applied to. Through experimentation it was found that the most robust structure for this problem was to relax the optimization constraints and move the final step length constraint to a final step cost. This leads the optimization routine to find acceptable trajectory results much more reliably. Figure 6.3 shows the difference in resulting trajectories from the relaxed and rigid optimization structures.





The final relaxed optimization formulation is as follows:

$$\min_{0 < t < T} \left[ g_F(\theta, u) + \int_0^T g(\theta, u) dt \right]$$

Equation 6.2

subject to

$$M(\theta)\theta + N(\theta,\theta) + G(\theta) = u$$
$$L_{St}\sin\theta_{St} - L_{Th}\sin\theta_{Th} - L_{Sh}\sin\theta_{Sh} - Foot_{Geo} \ge 0$$

where

$$L_{St} \sin \theta_{St}(T) - L_{Th} \sin \theta_{Th}(T) - L_{Sh} \sin \theta_{Sh}(T) - Foot_{Geo} = 0$$
$$\theta_{Th}(T) = \theta_{Sh}(T)$$

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 $\theta(0) = \theta_0$  $\theta_{Min} \le \theta \le \theta_{Max}$  $u_{Min} \le u \le u_{Max}$ 

The final relaxed optimization formulation, as shown in Equation 6.2, was implemented in Matlab with the use of PROPT from Tomlab. In general this optimization platform was able to accomplish the desired optimization procedure quickly and effectively but there were some minor issues with its limitations that directed this analysis that will be discussed as they are encountered throughout the section.

## System Model

The system model being used for the following analysis is a variation of the 5 degree of freedom model presented in Section 3 of this dissertation. It is a sagittal plane point mass model of a kneed bipedal human walker with a torso and a swing foot. The system model was further simplified to include only four segments, where the knee of the stance leg was omitted and assumed to be locked. This is an acceptable simplification because in application the knee of the stance leg is close to straight for a significant portion of the single stance phase as shown in Section 4. The model diagram can be seen in Figure 6.4.



Figure 6.4 – 4 Link Walking Model Diagram as Used in Trajectory Optimization

The foot is considered to be a zero mass object and is always perpendicular to the swing shank. Removing the dynamics of the swing foot is not a significant source of error because the primary effect that the swing foot has on the system is as an obstacle for toe clearance which is preserved. The system has three inputs in the form of the torques provided at the stance hip, the swing hip and the swing knee.

The system dynamics used for this analysis are simplified to make them easier to input into the optimization format. The first step to developing the simplified system dynamics used

in the analysis is to compute the full system dynamics for the point mass system described above using Legranian dynamics. The resulting dynamic equations that define the system are of the form shown in Equation 6.3. A complete evaluation of these dynamics can be found in the appendix located in Section 12 of this dissertation.

$$M(\theta)\ddot{\theta} + N(\dot{\theta},\theta) + G(\theta) = u$$

Equation 6.3

To input these dynamic equations into the optimization structure they needed to be explicitly solved for the angular acceleration values. In this system of four equations that is a much simpler said than done. When solved using the symbolic toolbox in Matlab, the solved equations for the four accelerations were nearly 3 pages long which led to it being completely infeasible to input into the optimization format. The structural simplification made to the dynamics that makes these dynamic equations useful is the creation of synthetic system inputs. In this case the synthetic inputs used are the angular accelerations of the relative joint angles. This is beneficial because it results in a much simpler dynamic format as shown in Equation 6.4.

$$\ddot{\theta} = A(\theta)v + B(\dot{\theta}, \theta) + C(\theta)$$

Equation 6.4

where

It is important to note that all the initial conditions of the system will correspond to after the toe of the walker has left the ground. The primary reason for this consideration is because human walking is a very hybrid system, in which it exhibits many different sets of continuous dynamics with discrete transitions between the states. The dynamics of the system are significantly different based on whether the foot of the swing leg is on the ground or not so this problem is addressed by placing the system into one of the discrete modes operation. Prior to application this modification will need to be accounted for by placing this technique within a state machine that provides a pre-swing state to prepare the optimization state by picking the foot off the ground, however, this will not be addressed within the scope of this extension.

$$\nu = \begin{bmatrix} \ddot{\theta}_{St} \\ \ddot{\theta}_{Th} \\ \ddot{\theta}_{Sh} \end{bmatrix}$$

#### Cost Function

The cost function is the item that makes this method work so strongly. It allows the desired result to be shaped based on what is more important very succinctly and effectively. To develop the cost function the most important criteria that distinguish one step over another must be identified. To meet the set forth criteria of matching the efficiency and effectiveness of the gait of an unimpaired human, two criteria were identified. First, the torso stability exhibited by the gait, in the form of penalizing for angular velocity of the torso and second, was the power used at the joints in a step. The resulting cost function that was used in this optimization and has shown the most reliable results is shown in Equation 6.5.

$$g(x,u) = c\phi(x,u)$$

Equation 6.5

where

$$\phi = \begin{bmatrix} (\sum_{Torso Angle Velocity})^2 \\ (Torso Angle Velocity)^2 \end{bmatrix} = \begin{bmatrix} (u_{St}(\dot{\theta}_{To} - \dot{\theta}_{St}) + u_{Th}(\dot{\theta}_{To} - \dot{\theta}_{Th}) + u_{Sh}(\dot{\theta}_{Th} - \dot{\theta}_{Sh}))^2 \\ \dot{\theta}_{To}^2 \\ c = \begin{bmatrix} 0.001 & 1 \end{bmatrix}$$

For the cost function when implemented on an exoskeleton there is another feature that could be added to improve the effectiveness of this optimization but it was left out because it is not a factor in unaffected gait. The desired feature was the addition of a clearance height cost. This would have been used to penalize the gait for clearing too close to the ground. This method can be implemented by adding an exponential component to the cost function based on the clearance height. This is important primarily because the optimization routine requires a differentiable cost function. This feature was not needed in this calculation because the to replicate the gait of an unaffected subject the toe is only required to clear the ground

The final cost is used to enforce the relaxed optimization constraints of reaching the desired step length and the desired torso angle as shown in Equation 6.6.

$$g_F(x,u) = c_F \phi_F(x,u)$$

Equation 6.6

where

$$\phi_F = \begin{bmatrix} (Step Length Error)^2 \\ (Torso Angle Error)^2 \end{bmatrix}$$
$$c_F = \begin{bmatrix} 100 & 1 \end{bmatrix}$$

## 6.1.2 OPTIMIZATION RESULTS

## Normal Walking

More existing information exists for the steady walking portion of the human gait than any other portion of human walking. But this information is not clear as to how it must be modified to match each particular set of initial conditions. Here the optimization routine is applied to a set of initial conditions that corresponds to a walking pose found in a steady walking raw data set produced by Winter [47]. A random amount of noise is added to that initial condition to result in a wide variety of optimization solutions.

The masses and lengths used in this analysis were selected to approximate adolescent male used to generate the raw data provided in Winter. The individual lengths and masses were determined based on the anthropomorphic data provided in Winter [47].

The analysis can be easily seen to result in acceptable walking trajectories as shown by Figure 6.5, Figure 6.6 and Figure 6.7. The trajectories can be seen to correlate strongly to the unimpaired trajectories available in literature [47, 48, 49]. This shows that the cost functions selected are sufficient to characterize the decisions made subconsciously in unimpaired bipedal walking.



Figure 6.5 – Near Optimal Swing Phase Body Posture for Normal Walking



The optimizer was also able to develop optimal trajectories unique to each of the different initial conditions as shown in Figure 6.8 and Figure 6.9. These trajectories would have been impossible to plan before hand using existing trajectory generation methods and would only be used given the ability to plan the entire step as is done with this method. The level to which the optimized trajectories match up with the accepted CGA trajectories shows that the trajectory planner was able to generate trajectories similar to those that would be generated by a healthy individual.



Figure 6.8 – Near Optimal Knee Angle Trajectories at Various Initial Conditions



Figure 6.9 – Near Optimal Hip Angle Trajectories at Various Initial Conditions

### First Step

While the ability of this method to match the known data of steady state walking is very useful, the primary benefit is its ability to generate data for unknown regions of the walking gait. The first step is one primary region where using the simple trajectory generation method leads to very poor overall performance of the gait. This method is capable of determining a gait trajectory similar to the trajectories that would be generated by an unimpaired human in the same scenario.

The simulations in this section are completed on a walker with the lengths and masses of the adolescent male as presented by Winter and used in the earlier section. The initial conditions of this section correspond to the first step just after the swing toe has left the ground but there is no angular velocity. This is important about the initial conditions for the first step because no velocity corresponds to the walker having no forward momentum which is by far the worst case scenario for the first step of walking. Despite these initial conditions the trajectory generator is still capable of generating a very acceptable gait trajectory for the first step as shown in Figure 6.10. The relative joint trajectories produced by the trajectory generator are found in Figure 6.11.

One interesting aspect of the first step trajectories is that they let the walker fall forward to gain some momentum and then extends the leg to catch itself. This phenomenon is very similar to that observed in the first step of healthy human walking but due to the lack of extensive cases to evaluate, reliable trajectories have yet to be identified.

This method allows the differences in the effect of particular initial conditions to be clearly identified and leads to being able to identify trajectories to account for the many initial condition variations much easier.



Figure 6.10 – Near Optimal Swing Phase Body Posture for a First Step



Currently, this method has not been applied to the actual system for one primary reason. The optimization routine is extremely calculation intensive and requires far more computing power than is available in the face of a real time system. On a desktop computer running with a processor speed of 2GHz, the optimization takes nearly 10s to complete the calculation. In a real time bipedal application, the swing phase can take between 0.6 and 1 second. So in order to be a realistic option for trajectory generation, the calculation would need to complete in 0.1 seconds at the longest on the embedded 100 MHz processor that is included on the device.

Despite not being implementable, this technique provide a significant result by demonstrating the applicability of this simple cost function to accurately represent the decisions made in unimpaired walking. With future improvements in processing power, or the application of other techniques such as Value Iteration, there may be possibilities for this to be implemented at a later time.

## 6.2 STEP PARAMETER PLANNING

Another limitation to the current implementation, is that the step parameters are currently selected by the user and then fed into the trajectory generation method without verifying that they are executing correctly. Regardless of how the trajectories are generated, enough uncertainty exists in the system due to the model or the input from the user that there are few guarantees that the steps will be conducted as desired by the user. The technique presented here, provides a method for feedback control of the step parameters in order to control that the desired motion is executing.

## Dynamic Analysis

The intended result of the step parameters is that the executed step meets the user selected values. Unfortunately, the dynamics of the gait make it impossible for all the step parameter combinations to be executed given the presented methods for generating step trajectories. The results of trajectories for steps at (12in, 2s), (24in, 4s) and (6in, 1s) will all lead to different stable walking gates despite the fact that the desired velocity is the same for all three gaits. This is because behind the step length and step time that the user selects is a desired velocity that has only one step parameter setting that results in that desired velocity. By changing the user input parameter to a desired walking velocity, a step time and step length can be determined that leads to a stable gait at the desired velocity for all selectable walking velocities.

A stable gait is defined by balanced energy throughout the gait cycle in that the energy input to the system is equal to that lost through the gait cycle as shown in Equation 6.7. At this point, all the work has been to establish the process by which energy is put into the system. The primary contributors to the input energy is the torque input from the actuators and in turn the trajectory generation method used, as well as how the specific user interacts with the device. None of these inputs should be adjusted for the purpose of step parameter planning because the device has no sensing of users interaction with the machine. It is also not clear how the trajectory generation methods presented earlier would be modified to adjust the energy input to the system.

$$\int_{Step} Energy_{Lost} = \int_{Step} Energy_{Input}$$

#### Equation 6.7

The missing piece is that nothing discussed in previous sections provides insight into what inputs control the energy lost through the gait cycle. Literature suggests that the primary source of energy loss through the gait cycle is due to the impact at heelstrike [63]. The energy lost at heelstrike is due to the coefficient of restitution of the vertical component of the center of mass velocity as shown in Figure 6.12. Based on the fact that a percentage of the vertical velocity is lost at impact, by adjusting the ratio of forward velocity to vertical velocity can provide a direct control input to the amount of energy that is lost at heelstrike. Some simple dynamic analysis using a rigid body collision assumption of the leg geometry and dynamics as shown in Equation 6.8, suggests that the length of the commanded step is directly proportional to the energy lost. In turn, following a step the device can determine if more or less energy needs to be burnt off from the following step and act accordingly.



**Figure 6.12 – Step Parameter Planning Point Mass Model** Shows center of mass velocity vectors both before and after heel strike impact

 $Energy_{Lost} \propto V_Y - V_Y' = V_Y(1 - e) = V \sin \theta \propto L_{Step}$ 

Equation 6.8

## Control Design

The controller designed for this application is a discrete control layer that exists in an outer loop compared to the trajectory generation and low level control segments of the control as shown in Figure 6.2. This controller must be discrete because the input necessary to compute a future control action is the resulting velocity of the step and thus new actions can only be determined after each individual step. The control method will use the feedback from the actual resulting velocity of the walking gait to close the loop on the user selected desired velocity and generate outputs of desired step time and step length.

The controller starts with a user selected desired velocity and an initialized desired step length of typically 12 inches. The controller has three simple steps involved in its execution. First, the controller computes a desired step time parameter that matches the current desired step length and velocity. Second, the desired step time and length are fed into the trajectory generator and a step is executed. Third, the desired step length is adjusted using a PD controller based on the error between the desired execution time and the actual execution time as shown in Equation 6.9. This update can be made because as the actual execution time is too fast compared to the desired, the controller is acting on the simple control law established above and increases the step length in order to burn more energy at heelstrike. The control diagram can be seen in Figure 6.13.

$$L_{Desired} \leftarrow L_{Desired} + e_T(k_P + k_D s)$$

Equation 6.9

where

$$e_T = T_{Desired} - T_{Actual}$$

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Figure 6.13 – Step Parameter Planning Block Diagram

 $V_{Des}$  is the desired forward velocity,  $T_{Des}$  is the step time parameter,  $L_{Des}$  is the step length parameter,  $T_{Act}$  is the actual step time and  $e_{\tau}$  is the error in the step time

## Simulation Results

The simulations used to verify the functionality of this controller were done using the three link simplified dynamic model presented in Section 3. Body segment masses were distributed proportional to accepted biomechanic proportions [47] between the torso and each leg, but the lengths of the segments to an arbitrary length of 1 meter. The lengths have no significant effect on the functionality of the controller and thus this simplification was acceptable. The trajectory generation method used in the simulations established sinusoidal hip angles from the starting angle to the intended ending hip angle needed to reach the desired step length with a period of one gait cycle. The desired joint angle trajectories were tracked with a PD low level controller similar to that of the actual device. While both the method of trajectory generation and the type of low level controller implemented will have an effect on the resulting stable step length and step time, the controller has no knowledge of them and handles these differences as disturbances to the system.

The controller was capable of tracking a desired set point velocity from an arbitrary set of initial conditions. As shown in Figure 6.14, when planning step parameters to track a desired walking velocity of 0.38 m/s the actual velocity exhibited some overshoot and then steadied out on the desired velocity. The bipedal walker required seven steps to fall within 10% of the desired velocity and 11 steps to track within 2%.




The response of the controller was also evaluated in the presence of a change in the desired velocity. In this scenario, as shown in Figure 6.15, the controller is able to track between two different desired velocities within 20 steps. An interesting aspect of the tracking of the desired velocity, is that as the reference changes, the closed loop dynamics exhibit a non-minimum phase characteristics as the initial response to the change in the desired velocity is opposite the direction of the error. Again, some overshoot is seen but the actual velocity does settle on the desired velocity.



Figure 6.15 – Step Parameter Planning Varying Reference Velocity Tracking

While this control strategy does provide a way to easily develop a steady walking gait without a system model, some drawbacks were seen in this initial implementation that must be evaluated prior to implementation. The first is that the method can only handle an initial velocity delta of between 0.12 and 0.15 m/s and still maintain a stable step. If too large of an initial velocity delta is present, then the controller has the potential to over correct to the point where the walker cannot complete the requested step and either falls forward over the heelstike foot or does not have enough momentum to get over the stance leg. This could be alleviated by improving the controller from a simple model free linear PD controller, to one that uses the dynamics of the system to either feedforward into the PD controller or by designing a model based controller.

The second drawback of this method is that while it does find step parameters that lead to a stable gait, the generated gait may require a lot of power. This can be seen as the controllers attempts to reach a high velocity, the step length will continue to update to shorter faster steps to reduce the amount of energy lost at heelstrike. This happens because the only input this controller has access to is the step length. In unimpaired walking, the energy input to the system can be modified as well by changing the trajectory generation method. This way the unimpaired gait can find a low power method for reaching the desired velocity. One method for addressing this issue is to limit the range of velocities that are allowed for the user based on the power required to complete the gait. In turn, extremely fast desired velocities that would lead to excessive power input would never be allowed. A second potential solution would be for the controller to modify the desired velocity to find the minimum cost of transport, which is energy per unit distance, for the given trajectory input.

# 7 SUBJECT TESTING

While significant effort has gone into the theoretical background of the controller applied on the exoskeleton, the lack of a sufficient system model requires significant verification through subject testing. This is an aspect of control that is necessary due to the human centered nature of the exoskeleton application and is needed to identify the realistic effectiveness of the device. In addition to this, some of the specific aims of this project are focused on the user response to the system and must be verified based on actual user experience.

One such goal that must be verified is that the device controller is intuitive to the user thus allowing new users to quickly pick up use of the device. This is necessary for use of this type of device because the subjects must be able to use the device at an adequate level to begin perfecting the use of the device. Second, once using the device, the user must be able to improve their level of expertise. The users showing significant improvement in the use of the device is needed to maintain the potential of future applications in the home mobility market. The final aspect that requires significant verification is that the device requires less energy expenditure from the user than current mobility options. This functionality is essential for both rehabilitation and mobility applications because all current mobility options are limited due to their energy expenditure requirements.

All subject testing has been conducted under the guidance of human subject testing protocols. In addition to this, the testing has been conducted in as safe an environment as possible based on the desired results of the testing by either maintain use of an overhead safety tether or by using a dedicated trained spotter. The trajectory generation for all the presented subject testing was done with the simple trajectory generation method and using the therapist controlled HMI method presented in Section 4.

# 7.1 MULTI-SUBJECT PILOT STUDY

The first subject testing verification objective is to determine if the controller is intuitive to the user. There is a minimum level of operator skill that is necessary to use the device at a functional level that includes standing and walking. While the controller is designed to minimize what is required of the user, if only a limited portion of the patient population is capable of reaching that level then the controller would be deemed too difficult. In this case, for the device to have the potential of widespread application across the mobility and rehabilitation markets, it must have a controller that accommodates a large range of users.

## Test Description

The test for this verification objective involved a pilot study at the Motion Analysis and Motor Performance Laboratory at the University of Virginia's Kluge Children's Rehabilitation Center using the mobile exoskeleton. The study involved seven spinal cord injury patients of various injury levels as detailed later in this section. Each patient received a single two hour testing session, where the subject was first introduced to the device and fitted into the machine. Upon arrival anthropometric measures were taken and the subject was fitted with the eLEGS device. Hip, knee and ankle joints of the exoskeleton were aligned with those of the subject. All tests were performed under a safety tether to prevent the subject from falling, and a spotter was close by to assist if necessary.

Once fitted, the exoskeleton raised the subjects into a standing position using a walker for balance. The subjects walked using a walker until they demonstrated a level of skill that allowed for transitioning to crutches as determined by the operator. The walking was done in 20' strait segments, in which subjects typically required 40' of walking before transitioning to lofstrand crutches. Upon transitioning to the lofstrand crutches, the subjects were further assisted until they were able to walk unassisted. The device was operated in the simple HMI mode, in which the therapist has control of triggering each individual step through the GUI. As discussed in Section 4, eLEGS has a user operated mode, however, for early training the therapist controlled mode provides the most safe environment for the subject to learn to operate the device. While in therapist controlled mode, the steps were triggered by the therapist when the subject had transferred their weight over the stance leg.

Then the user was sat down and given a 15 min break at which time a full body set of reflective markers (38) was applied to the subject and exoskeleton with additional markers placed on the lofstrand crutches as shown in Figure 7.1. Following the break, the user was stood again and the motion analysis system was calibrated before walking was resumed.



Figure 7.1 – Multi-subject Pilot Study Motion Capture Marker Locations

Three-dimensional kinematic data was collected using with a Vicon 8 camera motion analysis system at 120Hz. Additionally, ground reaction forces were measured using integrated 6 degree-of-freedom force plates in the center of the walking path. Data from the machine including joint angles, foot sensors, torso angle, and motor currents were also recorded on the exoskeleton computer. The entire time walking in the device at the end of the testing session did not exceed 45 min. At the end of testing the user was inspected for signs of excessive pressure concentrations.

## Patient Descriptions

The pilot study used 7 subjects which included 3 incomplete and 4 complete injuries. The subjects were also selected in accordance with the human subject testing protocol to have a good mix of injury levels, body sizes and gender. The breakdown of the individual subjects can be seen in Table 7.1. The injury to subject 7 is different than that of the other subjects because he has Friedreich's Ataxia which is an inherited disease characterized by progressive nervous system degradation. This disease typically results in spinal degradation that can affect natural gait patterns which are in turn addressed with available gait rehabilitation methods [64].

Subject Number	Injury Type	Time Since Injury [yr]	Age [yr]	Height	Weight [lbs]	Gender
1	Spinal Cyst – C7 – AIS C	32	44	5′ 5″	125	F
2	Paraplegic – T10 – AIS A	9	28	5′ 10″	140	М
3	Paraplegic with minor brain injury – T8 – AIS A	5	25	5′ 7″	121	F
4	Paraplegic – T5 – AIS A	3	22	6′ 1″	165	М
5	Tetrariplegic – C5 – AIS C	5	25	5′ 8″	120	М
6	Paraplegic – T5 – AIS A	2.5	24	6′ 1″	155	М
7	Friedreich's Ataxia (Presents like Incomplete)	38	53	5′ 10″	145	М

Table 7.1 – Multi-subject Pilot Study Subject Summary

## Results

In general the subjects took slow controlled steps during the testing. For all but one of the subjects, the data was collected when the subject was walking unassisted on lofstrand crutches. The only exception was subject 6 who was only able to walk unassisted using a walker during the single testing session. After processing the Vicon 3D kinematic motion capture data, only subjects 2, 3 and 5 had sufficient data to estimate the overground walking velocity and mean step length. The data for the other subjects was collected using the data logged from the computer processor. The summary of all spatial temporal parameters can be found in Table 7.2.

Subject Number	Mean Step Time [s]	Mean Step Length [in]	Mean Velocity [ft/s]	Mean Percent Single Support [%]	Mean Percent Double Support [%]
1	4.43	22.64	0.43	30.21	69.79
2	6.07	22.31	0.31	21.63	78.37
3	5.48	18.70	0.28	25.77	74.23
4	7.51	18.57	0.21	16.79	83.21
5	3.95	24.70	0.52	36.77	63.23
6	7.61	18.09	0.20	17.71	82.29
7	3.62	26.51	0.61	33.39	66.61

Table 7.2 – Multi-subject Pilot Study Spatial Temporal Parameters

The gait of the patients with complete injuries was typically slower than that of the incomplete patients. This slow down was primarily due to an elongation in the double stance portion of the gait cycle which increased by 20% for the subjects with complete injuries. The longer double stance phases lead to 67% longer average step times for the complete injury patients.

The sagittal plane joint angles for these subjects during walking demonstrate that the hip and knee controlled exoskeleton was able to generate a repeatable and reliable gait for the user. The sagittal plane joint angles for the hip knee and ankle are provided in Figure 7.2 and are representative of the sagittal angles for the subjects. The angles do not correspond to unimpaired gait trajectories because of the augmented angles generated by the trajectories and the extended double stance phases of the gait.

One aspect of these angles that are specific interest are the ankle angles. Unlike the hip and knee joints, the ankle is not actively controlled in the sagittal plane and instead is a passive joint that is managed by the user through the input of their arms. At no time in the gait cycle does the joint reach the joint hardstops which means that the user was able to effectively maintain control of the overall body posture throughout the gait with just their upper extremity inputs through the lofstrand crutches. While, future development will seek to lessen the requirement on the user's arms, the typical user has shown the ability to assist with body posture without active exoskeleton intervention to balance the body in the sagittal plane.



(A) Hip joint kinematics, (B) Knee joint kinematics, (C) Ankle joint kinematics. Shaded area indicated single standard deviation about provided joint angle trajectories.

For the subjects with incomplete injuries, other benefits were visible. Subject 5 in particular can walk unassisted without use of the exoskeleton but has an impaired gait that presents itself in the form of limited knee flexion as shown in Figure 7.3. However, with the use of the exoskeleton this patient experienced significantly increased knee flexion as well as a much more symmetric gait as shown in Figure 7.4.



**Figure 7.3 – Subject 5 Sagittal Plane Knee Joint Kinematics without Exoskeleton** Incomplete tetraplegic subject walking using no exoskeleton and no other walking aid. Shaded area indicated single standard deviation about provided joint angle trajectories.



**Figure 7.4 – Subject 5 Sagittal Plane Knee Joint Kinematics with Exoskeleton** Incomplete tetraplegic subject walking using the exoskeleton. Shaded area indicated single standard deviation about provided joint angle trajectories.

## Discussion

The end result of this testing showed that 6 of 7 patients tested in the device were able to walk unassisted using lofstrand crutches in a single 2 hour training session. The one subject who did not progress to unassisted walking was walking assisted using lofstrand crutches and unassisted in a walker at the end of the study. This subject was the one subject who had difficulty controlling their body in the wheelchair. During the testing he had issues with simple tasks such as transferring into and out of his wheelchair and balancing in a seated position. For this reason, he may be a subject that is not a first adopter user who may be screened from future testing using increased prescreening criteria. Despite this, the ability of 6 of the 7 subjects to advance to autonomous walking in a single 2 hour training session indicates that the desired functionality of the device behaving in an intuitive manner has been sufficiently met.

## 7.2 PATIENT PROGRESSION

The next verification step to proving out the design goals of the controller is to show that the users can continue to improve their skills with the device through continued use. As the users spend time in the device they must be able to significantly improve their skill level if they are going to be able to master the potential that the device offers. This is essential in rehabilitation applications, and potentially in mobility as well, by providing an incentive to the users to continue to use the device. This particular issue is one major issue with other mechanical therapy devices such as standing frames. Many of these devices are not used regularly by the users because while the device still provides the potential health benefits of standing, the novelty has worn off and the device no longer challenges and engages the user in trying to get any better.

#### Test Description

The test for this verification goal involved the compilation of independent testing sessions at the Berkeley Bionics testing facility and at UVA and involved two different subjects at different points in the learning cycle of the device. For the testing at Berkeley Bionics the user donned the device that had already been fit and sized for him based on measurements recorded from a previous testing session. The user stood up in the device and walked a series of 20' straight runs until the user felt familiar with the device. At that point, machine data was recorded including joint angles, torso angle, device state and controller output. A typical testing session lasted about two hours with around 45 min of actual walking. The user then sat down and doffed the device. At UVA, the testing sessions were conducted just as presented earlier in this section, but the subject was tested on two consecutive days.

Two different comparisons are evaluated in this test. The first test compares the gait of a single patient from two different phases in the learning cycle to identify improvements in selected gait parameters that are indicative of gait effectiveness. Due to the nature of the device, the controller is consistently being tweaked and between this subjects first testing session and second testing session many incremental modifications had been made resulting in an imperfect comparison. However, currently no subjects have logged as much training time as this subject, so no data exists that compares the improvement of a single patient with no changes in the controller over such a wide spread of the learning cycle. The second comparison is used to isolate this variable, by testing the second subject on the same controller at two different sessions to identify any learning

## Patient Descriptions

The first subject used in the testing is a 24 year old male, T10 AIS A paraplegic. The subject is 5 years post injury at the time of the testing, is 5' 11" tall and weighs 145 lbs. At the time of the first testing the subject had approximately 5 hours in the exoskeleton at the time of the first testing and 50 hours at the time of the second testing. The second subject used in this comparison is a subject 4 from the UVA pilot study presented earlier in this section.

## Results

The comparison for subject 1 shows that he has improved significantly with the device over his training time. Over his training this subject has demonstrated significant improvements in nearly all of the metrics being used to measure ability as shown in Table 7.3. The subject showed a 48.9% improvement in step time, a 103.6% improvement in walking speed and a 25.3% reduction in duration double support. The subject showed only a nominal improvement in step length of 3.8%.

Usage Time [hr]	Mean Step Time [s]	Mean Step Length [in]	Mean Velocity [ft/s]	Mean Percent Single Support [%]	Mean Percent Double Support [%]
5	7.20	24.19	0.28	39.97	60.03
50	3.68	25.12	0.57	55.15	44.85

Table 7.3 – Subject 2	L Spatial Temporal	<b>Parameter Comparison</b>
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The comparison for subject 2 also shows significant improvement with the device between the first two testing sessions as shown in Table 7.4. The subject demonstrated improvements in the same areas as the first subject but to a lesser degree. He had a 34.0% improvement in step time, a 76.2% increase in walking speed and a 5.9% reduction in the duration of double stance support. The subject showed only a nominal improvement in step length of 3.9%.

Table 7.4 – Subject 2 Spatial Temporal Parameter Comparison

Testing Session	Mean Step Time [s]	Mean Step Length [in]	Mean Velocity [ft/s]	Mean Percent Single Support [%]	Mean Percent Double Support [%]
1	7.51	18.57	0.21	16.79	83.21
2	4.96	19.29	0.32	21.70	78.30

## Discussion

While each patient is expected to perform differently and in turn have their own learning curve, these initial data points suggest that a significant amount of the subject learning can be done early in the testing based on these metrics. This data suggests that after only a few testing sessions, subjects may be able to learn a large amount about the function of the device and then continue to improve as they use the device more. Based on these subjects, the areas that have the most significant room to continue improving after the initial learning is the walking velocity and the duration of double support.

These data points suggest that significant learning is possible for the user as they get more familiar with the device. The comparison, however, is only suggestive because there is not direct comparison between the two subjects. To fully quantify the expected nature of the learning curve for many subjects, a multi-subject test must be completed with evaluation of the same subjects at many different points throughout the learning process while using the same device software. This was not conducted at this time, because not enough subjects have extensively used the device and the software is has been actively changing during these subjects learning process.

## 7.3 ENERGY EXPENDITURE

The final verification test is to identify whether the exoskeleton leads to decreased energy expenditure by the subject. This is a very important facet for a mobility device because while current walking alternatives exist for patients with spinal cord injuries, they are known to require nearly nine times more energy to ambulate in than an unaffected walking [65]. Showing this device successfully reduces the energy requirements for ambulation of a SCI subject will greatly improve its viability as an alternative tool for walking therapy.

## Test Description

To test this, an experiment was conducted with a single subject at the Berkeley Bionics facility using a portable VO2 testing machine which measures the rate that oxygen is taken in by the body. Oxygen consumption is a standard method for the verification of energy expenditure during ambulation [66, 67]. The device also records the amount of carbon dioxide exhaled by the body that is used to determine if the body is performing anaerobic work by looking to see if the amount of exhaled CO2 is inconsistent with the amount of oxygen inhaled. This is important because in an anaerobic state, the oxygen consumed is less than the entire amount of energy reserves [68].

The experiment starts by running a baseline test that measures the energy required by the user while sitting in their wheel chair. The test consists of remaining still and measuring the output of the VO2 device for a period of 5 to 7 minutes. This time is to allow the body to reach a steady amount of oxygen consumption during the testing period. The subject then donned the exoskeleton and took 10 to 20 steps to acclimate to the device. At this point the user

walked a 257 foot lap with the instruction to walk at a quick pace. The distance of this lap was selected to provide enough time for the oxygen expenditure to stabilize as data from the VO2 machine was recorded throughout the 6 to 7 minute lap. The subject then rested for 15 minutes during which they doffed the exoskeleton and donned their conventional walking aid. This subject uses two knee foot orthoses (KFOs) as conventional walking aids.

The subject then walked the same 257 foot lap in the braces while the VO2 machine was recording. The user was again asked to maintain a quick pace and told if their pace fell significantly below that of previous testing. The subject then rested 15 minutes at which time they doffed the walking braces and donned the exoskeleton again. Next, the subject walked the same 257 foot lap while in the device recording data and being instructed just as in the first exoskeleton test. The user then rested for 15 minutes again at which time they doffed the exoskeleton. To finish the test, the user once again took a baseline test sitting in their wheelchair.

#### Patient Description

The patient used in this testing is a 43 year old female, T10 AIS C paraplegic. The subject is 18 years post injury at the time of the testing, is 5' 7" tall and weighs 115 lbs. The patient has trace hip adductors that have began firing following stem cell therapy. The patient is a skilled exoskeleton user, having approximately 30 hours of testing at the time of the testing.

#### Results

The baseline tests at the beginning and end of the experiment were taken to verify that there was no significant trend in the data from the beginning of testing to the end. As shown in Figure 7.5, the baseline results were very close to each other. This can also be seen in the mean energy consumption after the signals stabilized as shown in Table 7.5.



Baseline tests taken sitting in wheelchair after oxygen consumption rate leveled off.

•	Average O2 Rate
	[ml/min]
Baseline Test 1	246.86
Baseline Test 2	248.76

Table 7.5 – VO2 Average Rate Comparison for Baseline Tests

The results from the walking tests also show significant distinction between the rate of oxygen consumption for the tests in the exoskeleton and the conventional walking aid. As shown in Figure 7.6, the two exoskeleton tests were very similar to each other and the brace walking test showed a significant amount more oxygen consumption.



**Figure 7.6 – VO2 Rate Comparison for Walking Tests** Exo Test 1 and 2 correspond to laps in exoskeleton while Brace Test corresponds to testing in conventional KFO walking braces.

The mean rate of consumption can be found in Table 7.6 after the values had stabilized. Also included are the mean velocities of each run. The velocities were calculated by evaluating the time needed to travel 207 feet by the user where the first and last 25 feet were not included to omit the temporal effects of getting up to and slowing down from the steady walking speed. Due to the fact that the velocities were substantially different between the three tests, the best comparison value is the cost of transport (COT) of the user which is the volume of oxygen consumed per unit distance traveled as opposed to time based on the average velocity of the individual test. The final piece of information provided is the indexed consumption rates and COT. These values are adjusted from their non-indexed versions by removing the average baseline consumption to compare only the energy consumption associated with the ambulation action. All of these values can be seen in Table 7.6.

	Average O2 Rate [ml/min]	Average Indexed O2 Rate [ml/min]	Average Velocity [mph]	Average O2 COT [ml/ft]	Average Indexed O2 COT [m]/ft]
Exo Test 1	618.44	370.63	0.475	14.79	8.86
Exo Test 2	652.17	404.36	0.508	14.60	9.05
Brace Test	967.10	719.29	0.385	28.58	21.25

Table 7.6 – VO2 Average Comparisons for Walking Tests

#### Discussion

The data demonstrates a significant reduction in the energy expenditure for ambulation when using the exoskeleton when compared to conventional walking aids. The use of the device resulted in a rate of oxygen consumption 53.9% of that consumed when using the walking braces when comparing the indexed rate values. In addition to that, the average velocity when walking in the exoskeleton was noticeable increased which resulted in an even better improvement in the energetic cost of transport. Due to this the indexed cost of transport with the exoskeleton was only 42.1% of the conventional method. This improvement is significant enough to suggest that the energy expenditure required for use of the exoskeleton would not be as significant a barrier to extended use that it is for brace walking.

Another interesting aspect of this test is that some data suggested the results are conservative compared to the potential benefits. The main aspect is that the ratio of oxygen consumed to carbon dioxide exhaled for the brace walking test was near the level that would indicate anaerobic work may have been done. Over the walking time, the ratio averaged 0.995 over the stabilized segment of the testing session. A ratio of 1 indicates the anaerobic limit is reached, however, this limit is not a strict boundary so a tested ratio that is so close to 1 indicates that for this patient the anaerobic limit may have been reached. If it had been reached, the energy consumption estimated by the oxygen intake would be a conservative estimate of the energy used in the brace walking test. If the actual energy consumed during this test were larger the percentage reduction would be even larger for the exoskeleton testing.

While this test does indicate that there may be significant benefits in this area, it is not sufficient to guarantee that the benefits should be expected for the typical patient. The primary issue with the test is that only one subject was evaluated. In the future, similar tests on a larger sample size of subjects can be conducted in order to accommodate for the fluctuations found in testing any one patient. One such fluctuation that existed in this testing is that the subject's braces, while custom fitted for her at the time of purchase, were identified by the physical therapist present during the testing as not being the best design for her condition. If it is the case that typical patients are in non-optimal braces then little change would be expected when the sample size expands, however, if this subject is the one person with poor braces then a large sample size would serve to drown out the perceived improvements shown by her testing.

# 8 CONCLUSION

This work sought to develop a control structure and a method of trajectory generation for eLEGS, a mobility exoskeleton for SCI patients, which is safe, practical and overcomes the limitations presented by existing options. The controller was designed to take advantage of the users upper extremity input using lofstrand crutches and accommodate for the limitations presented by the physical system while meeting three specific goals. First, the controller was structured such that it is applicable to a wide range of maneuvers, environments and control methods. The strategy employed must allow for adding to the available maneuvers to accommodate new needs without significantly limiting previous functionality. Second, the trajectories generated for walking must be similar to that of an unaffected gait. For many SCI subjects, there is potential to regain some degree of muscle control; therefore, the device must generate a healthy gait in order to provide a driven gait style of rehabilitation. Third, the control output must be intuitive and repeatable. This will allow users to quickly learn the function of the device and then continue to improve in their abilities as they adapt to the repeatable behavior of eLEGS.

## 8.1 SUMMARY

In the previous sections of this work, the difficulties found in developing a complete and accurate system model were discussed. Based on these difficulties, a controller structure was presented that works with the motion of the user to produce the intended maneuver without requiring an explicit model. The structure of the controller, as presented in Section 4, is extremely flexible allowing for the easy implementation of new maneuvers or control methods as demonstrated in Sections 5 and 6. The control structure utilized three independent controller blocks to accomplish individual tasks that are the human machine interface, trajectory generator and low level controller. By structuring the controller this way, it provides a structure to build on for future additions and improvements while allowing the subjects to reliably move through the entire gait.

This work also presented a method of trajectory generation in Section 4, which allows a repetitive, safe ambulation to SCI subjects in eLEGS. This trajectory generation method was structured around a state machine to determine the lower extremity motion that best suits the current phase of the gait. The active trajectory generation method presented in Section 5, allows the device to modify the gait actively throughout the swing cycle of the trajectory in order to develop the intended foot motion with a trajectory that is not overly conservative. In Section 6, additional improvements to the trajectory generation process were presented that generated near optimal trajectories via optimization and that planned step parameters for a stable limit cycle gait at a desired velocity. While these extensions have not reached the level of implementation, they have provided significant support for the fact that the trajectory generation centered control structure provides the most viable option for a safe, full featured mobile exoskeleton controller.

In addition to this, all aspects of the trajectory generation were able to maintain these features while producing a gait that is similar to that of an unaffected subject. While it is very possible that a few highly skilled subjects could learn to use the device if the joint trajectories were significantly different than that of a healthy gait, by minimizing these differences the controller allows more users to adapt to using it. In addition, the near healthy gait provides added incentive for subjects with incomplete injuries to use the device because they are actively undergoing driven gait rehabilitation while using the device. Because these subjects potentially have the neurological ability to regain some level of lower extremity function based on the nature of their injury, the rehabilitative nature of the mobility device will be extremely enticing to this patient population.

This work also presented the first results of subject testing on the developed controller. Because of the difficulties presented by modeling the entire system, including the interaction of the user with the device, it is essential that the function of the device is shown in practice as opposed to simulation alone. In Section 7 of this work three primary design goals were evaluated through subject testing. First, the ability of new users to learn to use the device quickly and easily was demonstrated through the results of an initial multi-subject pilot study where 6 of the 7 subjects were able to walk unassisted using forearm crutches after only a single two-hour session of training. This is an extremely crucial result that implies the device will be able to get people up and walking quickly as opposed to requiring months of training prior to experiencing the benefits. Second, the ability of users to continue to improve with the device was demonstrated using an evaluation of one new subject over his first visits and by comparing a skilled user to his first few times of walking. These showed that the users can continue to grow more skilled with the device as they continue to use it, potentially leading to the possibility of subjects realistically using the device in real mobility environments. Finally, the energy expenditure required to use the device was shown to drop significantly from that of conventional walking aids by evaluating oxygen consumption rates for a single subject. This is also a significant result because one of the primary drawbacks from the widespread adoption of conventional walking braces is that they require too much energy from the SCI subject. Thus, by significantly reducing the required energy expenditure the potential for widespread use is still possible.

## 8.2 FUTURE WORK

The development of the controller for this device is the first comprehensive control strategy for a mobility exoskeleton for SCI patients. As the first in its field, the application of this device leads to as many open questions and possibilities as it does answers. Presented here are a collection of the primary areas that require additional work in the future to further the capabilities of this device.

The first area that requires additional attention is the area of pose estimation. Currently, the pose of the device is estimated using only the measured joint angles of the device along with the sagittal plane absolute angle of the torso to calculate a Cartesian coordinate estimate of the current exoskeleton pose. While this is a good start, the joint angles are considered perfect sources of state information and the exoskeleton mechanical assembly is assumed to be perfectly rigid, which are both incorrect assumptions. In addition to this, there is no information regarding the angle of the ankle, so when estimating the toe position, the ankle is assumed to be at the neutral position. As implemented, the toe of the swing leg is only able to be estimated within about 3 inches of its actual location because a few degrees of total error at the joints can lead to significant deviations. This implementation is currently one of the only options because no source of redundant information exists regarding the device's absolute pose. Many options exist, but they will require significant work in modeling and signal processing. One such example is to include a sagittal plane accelerometer and gyroscope in the ankle of the exoskeleton and use the newly available information to generate an absolute angle for the foot of the device so as to provide redundant information on the absolute acceleration of the foot. This information, along with a 7-link sagittal plane model where the links are modeled as rotational springs, could be fused using an extended kalman filter or a particle filter to estimate the actual pose of the device. An accurate estimation of the robot's pose could then be used in many ways to improve the functionality of the device by providing direct feedback into the active swing trajectory method about step length and clearance height or allowing more precise model based control methods for advanced features such as standing or balancing.

This leads directly into the next area that deserves further attention, which is methods for active balancing of the device or balance assistance to the user in standing and walking. The easiest active balancing problem is to actively balance the device while in a stationary standing position because a desired set point can be easily specified, and with the existing actuation at the hip, it is very similar to the classical control problem of the acrobat which is an inverted double pendulum with actuation in the middle. While the source actuation may exist, there are still some major issues regarding system identification of the mass distribution of the user because the device will be designed to work with any user who wears the device. Because the control for this application will most likely be model centered, an inaccurate model may lead to significant control limitations due to safety for the user. Another issue that must be addressed when it comes to balance is whether the hip is the best method for control of balance at all or if the balance can be more robustly implemented with the addition of an active ankle. Because the hip has a direct effect on the relation of the torso to the legs and the torso comprises most of the mass for the system, it is reasonable to believe that it can have a significant effect on the balance of the device. However, because the ankle is located directly at the point of balance, it may present a better location to input small amplitude, but high frequency input to the system by actively stiffening or actuating the device by a method of active control or active dampening. After all of this is evaluated, the next extension for this technology is to the area of single stance balance assistance similar to what an unaffected human subject does. This application is generally more complicated due to the limited ability to predefine a desired set balance point because the single stance phase is a dynamic motion and directly associated with the forward momentum of the gait.

The next area that deserves some specific attention is the use of the stance phases of the gait for forward propulsion. This is extremely important because currently the users gain their forward propulsion primarily with the aid of gravity through falling forward through the stance phases of the gait. It may be possible, through the use of better posture control and balancing, to monitor the forward momentum of the step through the stance phases to control and actively add a level of propulsion to the user while walking. This area is extremely dependent on the first two areas identified for future work, but it is the first that will lead to the direct result of faster overground walking that for many is the critical metric for determining the feasibility of mobility devices in the general market. One aspect that may prove necessary towards executing a final solution is actuation out of the sagittal plane to assist the user with coronal plane weight shift. This is extremely important because the addition of significant forward propulsion is not helpful if the user is not capable of controlling the coronal motion of the body at the higher speeds.

The final area that stands out as needing further work is the area of variable assist for the rehabilitation and mobility effectiveness of high functioning incomplete patients. While the current state of the art in rehabilitation is moving toward force feedback and impedance based control [31, 32], it is unclear how this transfers to mobile walking where, if the user fails to respond to the force feedback, they fall. In addition to that, even in solely mobility applications, the user is able to process and respond to the environment much better than the exoskeleton will be able to in the foreseeable future because of additional sensor inputs such as sight and a more capable form of fusing all of the sensor information together so it makes sense to use the input of these users when it is available. The current approach is focused on completely positioning the user, and while other research has focused on responding to the motions of the user, no unifying method exists that encompasses variable assistance levels from 100% assistance as currently provided to the possible 0% assistance available in other devices.

## 8.3 CLOSING

In the end, this work presents a flexible control structure and a novel method of trajectory generation that is robust and extremely capable. It develops the first comprehensive control architecture for a mobility exoskeleton for SCI patients and lays the framework for a new area of research into the human centered control of mobility exoskeletons. It is implemented and presented in a manner that allows significant research into isolated aspects of the controller function without losing or compromising the function of the remainder of the controller. This work presents the first significant subject testing of a device of this nature, as well as demonstrates significant improvements over the existing drawbacks of current mobility options.

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## **10** APPENDIX: **3** LINK WALKING MODEL IMPACT RESET DERIVATION

The equations of reset at the impact event of heelstrike for the 3 link simplified sagittal plane walking model as provided in Section 3.1.1 are computed based on the conservation of angular momentum. The conservation of angular momentum is calculated using simple rigid body impact model which allows for an instantaneous impact and a discontinuity in angular velocity but not in the angle of any segment. For the 3 member linked chain in this problem, this is calculated by balancing 3 conditions as presented below:

- Conservation of angular momentum of the whole body about the impact foot
- Conservation of angular momentum of the original stance leg about the hip
- Conservation of angular momentum of the torso about the hip

These three conditions are generated by isolating the system into segments that have only one source of impulse forces acting on them and then conserving momentum about the location of the impulse forces. Only impulse forces are considered in this derivation because the integral of non impulse forces over an instantaneous impact is zero as shown in *Equation* 10.1

$$\frac{dJ}{dt} = T \rightarrow \int dJ = \int T dt \rightarrow \Delta J = \Delta T_{(t^- \rightarrow t^+)} \approx 0$$

Equation 10.1

The angular momentum about a point is calculated as shown in *Equation 10.2*.

$$J_A = \sum_i d_{iA} \times m_i v_i$$

Equation 10.2

where

- $J_A$  Angular Momentum about Point A
- $d_{iA}$  Vector from Point A to Center of Mass of Link i
- *m i* Mass of Link i
- v i Velocity of Center of Mass of Link i

The individual computation of the first condition is completed based off Figure 10.1.



A Conservation of angular momentum condition 1 of whole body about impact position, B) Conservation of momentum condition 2 of original stance leg about hip, C) Conservation of angular momentum condition 3 of torso about hip

$$J_{1A}^{-} + J_{2A}^{-} + J_{TA}^{-} = J_{1A}^{+} + J_{2A}^{+} + J_{TA}^{+}$$

Equation 10.3

where

$$J_{iA}^{+} = d_{iA} \times m_{i}v_{i}^{+}$$

$$L_{i} = r_{i} + s_{i}$$

$$d_{1A} = [L_{L}\cos\theta_{2} - s_{L}\cos\theta_{1} \quad L_{L}\sin\theta_{2} - s_{L}\sin\theta_{1}]$$

$$d_{2A} = [r_{L}\cos\theta_{2} \quad r_{L}\sin\theta_{2}]$$

$$d_{TA} = [L_{L}\cos\theta_{2} + r_{T}\cos\theta_{T} \quad L_{L}\sin\theta_{2} + r_{T}\sin\theta_{T}]$$

$$v_{1}^{-} = [-r_{L}\dot{\theta}_{1}^{-}\sin\theta_{1} \quad r_{L}\dot{\theta}_{1}^{-}\cos\theta_{1}]$$

$$v_{2}^{-} = [-L_{L}\dot{\theta}_{1}^{-}\sin\theta_{1} + s_{L}\dot{\theta}_{2}^{-}\sin\theta_{2} \quad L_{L}\dot{\theta}_{1}^{-}\cos\theta_{1} - s_{L}\dot{\theta}_{2}^{-}\cos\theta_{2}]$$

$$v_{T}^{-} = [-L_{L}\dot{\theta}_{1}^{-}\sin\theta_{1} - r_{T}\dot{\theta}_{T}^{-}\sin\theta_{T} \quad L_{L}\dot{\theta}_{1}^{-}\cos\theta_{1} + r_{T}\dot{\theta}_{T}^{-}\cos\theta_{T}]$$

$$v_{1}^{+} = [-L_{L}\dot{\theta}_{2}^{+}\sin\theta_{2} + s_{L}\dot{\theta}_{1}^{+}\sin\theta_{1} \quad L_{L}\dot{\theta}_{2}^{+}\cos\theta_{2} - s_{L}\dot{\theta}_{1}^{+}\cos\theta_{1}]$$

$$v_{2}^{+} = [-r_{L}\dot{\theta}_{2}^{+}\sin\theta_{2} - r_{T}\dot{\theta}_{T}^{+}\sin\theta_{T} \quad L_{L}\dot{\theta}_{2}^{+}\cos\theta_{2} + r_{T}\dot{\theta}_{T}^{+}\cos\theta_{T}]$$

The computation of condition 2 is computed as shown in Equation 10.4 where the velocities are the same as presented previously in this section.

$$J_{1B}^- = J_{1B}^+$$

Equation 10.4

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where

$$J_{iB}^{+} = d_{iB} \times m_i v_i^{+}$$
$$L_i = r_i + s_i$$
$$d_{1B} = [-s_1 \cos \theta_1 \quad -s_1 \sin \theta_1]$$

The computation of condition 2 is computed as shown in Equation 10.5 where the velocities are the same as presented previously in this section.

$$J_{\rm TC}^- = J_{\rm TC}^+$$

Equation 10.5

where

$$J_{ic}^{+} = d_{ic} \times m_i v_i^{+}$$
$$L_i = r_i + s_i$$
$$d_{TC} = [r_1 \cos \theta_T \quad r_T \sin \theta_T]$$

Following expansion and simplification, these three conditions solve to the system presented in Equation 10.6 where  $Q^-$  and  $Q^+$  are the resulting transformation matrices from the simplified system.

$$Q^{-}\begin{bmatrix}\dot{\theta}_{1}\\\dot{\theta}_{2}\\\dot{\theta}_{T}^{-}\end{bmatrix} = Q^{+}\begin{bmatrix}\dot{\theta}_{1}^{+}\\\dot{\theta}_{2}^{+}\\\dot{\theta}_{T}^{+}\end{bmatrix}$$

Equation 10.6

$$Q^{-} = \begin{bmatrix} A & -m_{L}r_{L}s_{L} & m_{T}r_{T}(L_{L}\cos(\theta_{2} - \theta_{T}) + r_{T}) \\ -m_{L}r_{L}s_{L} & 0 & 0 \\ m_{T}r_{T}L_{L}\cos(\theta_{1} - \theta_{T}) & 0 & m_{T}r_{T}^{2} \end{bmatrix}$$
$$A = (2m_{L}r_{L} + m_{T}L_{L})L_{L}\cos(\theta_{1} - \theta_{2}) + m_{T}r_{T}L_{L}\cos(\theta_{1} - \theta_{T}) - m_{L}s_{L}r_{L}$$
$$M_{L}s_{L}(s_{L} - L_{L}\cos(\theta_{1} - \theta_{2})) & B & m_{T}r_{T}(L_{L}\cos(\theta_{2} - \theta_{T}) + r_{T}) \\ m_{L}s_{L}^{2} & -m_{L}s_{L}L_{L}\cos(\theta_{1} - \theta_{2}) & 0 \\ 0 & m_{T}r_{T}L_{L}\cos(\theta_{2} - \theta_{T}) & m_{T}r_{T}^{2} \end{bmatrix}$$

$$B = m_L r_L^2 + (m_L + m_T) L_L^2 - m_L s_L L_L \cos(\theta_1 - \theta_2) + m_T r_T L_L \cos(\theta_2 - \theta_T)$$

# **11 APPENDIX: TRAJECTORY GENERATION BLENDING**

## **11.1 LINEAR BLENDING**

The linear blending used in the trajectory generation is designed to be easy to implement and to only have setup calculations at the beginning of the blend operation. This blend is intended to result in a function that is zero order continuous. In the case that two lines are being blended as shown in Figure 11.1, the calculation is made to blend out the initial error between the two lines over a set amount of time. Thus, the final line can be explicitly defined as a piecewise function as shown in *Equation 11.1*.

$$f(t) = \begin{cases} f_1(t) & t \le 0\\ f_2(t) + e_0(1 - \frac{t}{t_{Blend}}) & 0 < t < t_{Blend}\\ f_2(t) & t \le t_{Blend} \end{cases}$$

Equation 11.1

where

 $e_0 = f_1(t_0) - f_2(t_0)$ 



Blend starting at time 0 with a blend time of 1 s.

## **11.2 FIRST ORDER CONTINUOUS BLENDING**

This blend is similar to that used in the linear blend, but it is attempting to maintain a continuous velocity throughout the blending process. The result of the blend can be seen in Figure 11.2 with the piecewise explicit solution provided in *Equation* 11.2.

$$f(t) = \begin{cases} f_1(t) & t \le 0\\ f_2(t) + At^3 + Bt^2 + Ct + D & 0 < t < t_{Blend}\\ f_2(t) & t \le t_{Blend} \end{cases}$$

Equation 11.2

$$D = e_0 = f_1(t_0) - f_2(t_0)$$
$$C = \dot{e}_0 = \dot{f}_1(t_0) - \dot{f}_2(t_0)$$
$$B = -2\frac{\dot{e}_0}{t_{Blend}} - 3\frac{e_0}{t_{Blend}^2}$$
$$A = \frac{\dot{e}_0}{t_{Blend}^2} + 2\frac{e_0}{t_{Blend}^3}$$



**Figure 11.2 – Trajectory Generation First Order Continuous Blend** Blend starting at time 0 with a blend time of 1 s.

## **12 APPENDIX: NEAR OPTIMAL TRAJECTORY GENERATION DYNAMICS**

The system being modeled for this analysis is a four link point mass walking model with a torso. The system has a straight stance leg and a kneed swing leg as shown in Figure 6.4. The complete description of this model is located in Section 6.1.

The dynamics were solved using Legrangian dynamics and can be stored in the form provided in Equation 12.1. The individual equations that make up this master system are provided in Equation 12.2 to Equation 12.5.

$$M(\theta)\ddot{\theta} + N(\dot{\theta},\theta) + G(\theta) = u$$

Equation 12.1

$$\theta = \begin{bmatrix} \theta_{St} \\ \theta_{Th} \\ \theta_{Sh} \\ \theta_{To} \end{bmatrix}$$
$$u = \begin{bmatrix} u_{St} \\ u_{Th} \\ u_{Sh} \end{bmatrix}$$

$$u_{St} = ((m_{Th} + m_{Sh} + m_{To})L_{St}^{2} + m_{St}r_{St}^{2})\ddot{\theta}_{St} - L_{St}(m_{Th}s_{Th} + m_{Sh}L_{Th})\cos(\theta_{St} - \theta_{Th})\ddot{\theta}_{Th} - m_{Sh}s_{Sh}L_{Th}\cos(\theta_{St} - \theta_{Sh})\ddot{\theta}_{Sh} + m_{To}r_{To}L_{Th}\cos(\theta_{St} - \theta_{To})\ddot{\theta}_{To} - L_{St}\dot{\theta}_{Th}^{2}(m_{Th}s_{Th} + m_{Sh}L_{Th})\sin(\theta_{St} - \theta_{Th}) - m_{Sh}s_{Sh}L_{St}\dot{\theta}_{Sh}^{2}\sin(\theta_{St} - \theta_{Sh}) + m_{To}r_{To}L_{St}\dot{\theta}_{To}^{2}\sin(\theta_{St} - \theta_{To}) + g((m_{Th} + m_{Sh} + m_{To})L_{St} + m_{St}r_{St})\cos(\theta_{St}) Equation 12.2$$

$$u_{Th} + u_{Sh} = -L_{St}(m_{Th}s_{Th} + m_{Sh}L_{Th})\cos(\theta_{St} - \theta_{Th})\ddot{\theta}_{St} + (m_{Th}s_{Th}^2 + m_{Sh}L_{Th}^2)\ddot{\theta}_{Th}$$
  
+  $m_{Sh}s_{Sh}L_{Th}\cos(\theta_{Th} - \theta_{Sh})\ddot{\theta}_{Sh} + L_{St}\dot{\theta}_{St}^2(m_{Th}s_{Th} + m_{Sh}L_{Th})\sin(\theta_{St} - \theta_{Th})$   
+  $m_{Sh}s_{Sh}L_{Th}\dot{\theta}_{Sh}^2\sin(\theta_{Th} - \theta_{Sh}) - g(m_{Th}s_{Th} + m_{Sh}L_{Th})\cos(\theta_{Th})$   
Equation 12.3

$$-u_{Sh} = -m_{Sh}s_{Sh}L_{St}\cos(\theta_{St} - \theta_{Sh})\ddot{\theta}_{St} + m_{Sh}s_{Sh}L_{Th}\cos(\theta_{Th} - \theta_{Sh})\ddot{\theta}_{Th} + m_{Sh}s_{Sh}^2\ddot{\theta}_{Sh}$$
$$+ m_{Sh}s_{Sh}L_{St}\dot{\theta}_{St}^2\sin(\theta_{St} - \theta_{Sh}) - m_{Sh}s_{Sh}L_{Th}\dot{\theta}_{Th}^2\sin(\theta_{Th} - \theta_{Sh})$$
$$+ m_{Sh}s_{Sh}g\cos(\theta_{Sh})$$

Equation 12.4

$$-u_{St} - u_{To} = m_{To} r_{To} L_{St} \cos(\theta_{St} - \theta_{To}) \ddot{\theta}_{St} - m_{To} r_{To}^2 \ddot{\theta}_{To} - m_{To} r_{To} L_{St} \dot{\theta}_{St}^2 \sin(\theta_{St} - \theta_{To}) + m_{To} r_{To} g \cos(\theta_{To})$$

Equation 12.5

$$s_{Link} = L_{Link} - r_{Link} \forall Link$$