A POWERED LOWER LIMB EXOSKELETON SUPPLEMENTED WITH FES

FOR GAIT ASSISTANCE IN PARAPLEGIC PATIENTS

By

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A mi madre, quien me dio todas las herramientas a su alcance.

"To my mother, who gave me all the tools at her disposal"

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CHAPTER I

Regaining standing and walking capacity in paraplegic population

Introduction and problem definition

Just in the U.S. there are 12 000 new cases of paraplegic patients every year. One of the most significant impairments resulting from paraplegia is the loss of mobility, particularly given the relatively young age at which such injuries occur [1-3]. In addition to impaired mobility, the inability to stand and walk entails severe physiological effects, including muscular atrophy, loss of bone mineral content, frequent skin breakdown problems, increased incidence of urinary tract infection, muscle spasticity, impaired lymphatic and vascular circulation, impaired digestive operation, and reduced respiratory and cardiovascular capacities [4]. In the last decades two major approaches to achieve upright position and walking have been reported in the literature: orthoses/exoskeletons type devices and functional electrical stimulation (FES). The combination of passive orthoses combined with FES has also been wildly studied by researchers. A powered orthosis can provide leg movement to facilitate legged locomotion. However, in such a system, the user will not derive the documented physiological and psychological benefits resulting from FES (i.e., resulting from walking under their own power). In order to gain the benefits of FES-aided legged locomotion, avoid the deficiencies of FES, and leverage recent advances in robotics technology, this work presents a powered lower limb exoskeleton supplemented with FES for gait assistance in paraplegic patients.

Document organization

This document is organized into six chapters. Chapter I present the introduction, problem definition and an overview of the work reported by other researchers pursuing standing position and walking in paraplegic patients. Chapters II to V are papers submitted to journals or peer-reviewed conferences

where the author of this dissertation has a significant contribution. Specifically, chapter II presents the Vanderbilt exoskeleton concept and focuses on its mechanical design aspects. Chapter III details the electronic design and the device control scheme. Chapter IV focuses on efficiency of the device, and reports on a T10 complete SCI subject who performs sit to stand, walking and sitting maneuvers. In chapter V a cooperative controller that enables the unit to use FES is presented and tested. Each paper contains an abstract that provides a more detailed description. The dissertation conclusions are presented in Chapter VI.

<u>Overview</u>

Orthoses to enable legged locomotion

In an effort to restore some degree of legged mobility to individuals with paraplegia, several passive and powered lower limb orthoses have been developed and described in the engineering literature. The simplest form of passive orthotics are long-leg braces that incorporate a pair of ankle-foot orthoses (AFOs) to provide support at the ankles, which are rigidly coupled to leg braces that lock the knee joints against flexion. The hips are typically stabilized by the tension in the ligaments and musculature on the anterior aspect of the pelvis. Since almost all energy for movement is provided by the upper body, these (passive) orthoses require considerable upper body strength and a high level of physical exertion, and provide very slow walking speeds. A more sophisticated orthosis, the hip guidance orthosis (HGO), is described in [5]. The HGO incorporates hip joints that rigidly resist hip adduction and abduction, and rigid shoe plates that provide increased center of gravity elevation at toe-off, thus enabling a greater degree of forward progression per stride. Another variation on the long-leg orthosis, the reciprocating gait orthosis (RGO), incorporates a kinematic constraint that links hip flexion of one leg with hip extension of the other, typically by means of a push-pull cable assembly. As with other passive orthoses, the paraplegic individual leans forward against the stability aid, utilizing gravity to provide hip extension of the stance leg. Since motion of the hip joints is reciprocally coupled, the gravity-induced hip extension also provides contralateral hip

flexion (of the swing leg), such that the stride length of gait is increased. Examples of this type of orthosis, and studies of its efficacy, are described in [6-12],

In order to decrease the high level of exertion associated with passive orthoses, some researchers have investigated the use of powered orthoses, which incorporate actuators to assist with locomotion. Historical efforts to develop powered orthoses to aid in paraplegic mobility include [13-16] More recently, Ruthemberg [17] developed a powered orthosis for evaluating design requirements for paraplegic gait assistance. In [18-20], a powered orthosis was developed by combining three electric motors with an RGO, two of which were located at the knee joints to enabled knee flexion and extension during swing, and one of which assisted the hip coupling, which in essence assisted both stance hip extension and contralateral swing hip flexion. The orthosis was shown to increase gait speed and decrease compensatory motions, relative to walking without powered assistance.

In [21-24], the authors describe control methods for providing assistive maneuvers (sit-to-stand, stand-to-sit, and walking) to paraplegic individuals with the powered lower limb orthosis HAL, which is an emerging commercial device with (in the incarnation utilized in the aforementioned publications) six electric motors (i.e., powered sagittal plane hip, knee, and ankle joints). As discussed in [21-24], the authors propose the use of a zero-moment-point (ZMP) method to actively control balance during movement. Like the powered lower limb orthosis HAL, two additional emerging commercial devices are the ReWalk powered orthosis (Argo Medical Technologies) and the ekso powered orthosis (Ekso Bionics). At the time of writing of this document, studies have not been published regarding the design, performance characterization, or efficacy of these systems. Despite this, based on product information available from the respective companies, ReWalk utilizes a wrist-mounted keypad to select between modes of operation (while utilizing upper body tilt to gate subsequent steps while walking), while the ekso orthosis utilizes instrumentation on the forearm crutches to indicate user intent.

Functional electrical stimulation

Functional electrical stimulation (FES) is a means of artificially eliciting muscular contractions from either skin surface or implanted electrodes. The muscular physiology in many spinal cord injured persons is essentially intact. Complete injury in the spinal cord, however, prevents the electrical signals sent from the central nervous system (CNS) from traveling via neural pathways to respective muscles. As such, paralysis is not a deficiency of the muscle, but rather an inability of the CNS to initiate a muscular contraction. FES utilizes electrical stimulation of peripheral motor nerves, using either skin surface or implanted electrodes, to elicit muscular contractions. In contrast to passive (or powered) orthoses, FES actively involves the musculature of the lower as well as the upper body, and thus provides a means of total body exercise that has been shown to alleviate some of the physiological impairments resulting from immobility [25-30] Because the power for movement is derived metabolically, the FES system need only provide signal power to the nerves, and therefore (unlike a powered orthosis), the external (i.e., battery) power requirements are extremely modest.

Bajd and Kralj [31-34] proposed a protocol of stimulation that indirectly stimulates hip flexors via stimulation of the common peroneal nerve, which in turn elicits the flexion withdrawal reflex (i.e., like stepping on a tack). Specifically, stimulation of the common peroneal nerve is utilized to elicit the flexion withdrawal reflex, which provides simultaneous hip and knee flexion, while a (direct stimulation) quadricep channel is used to provide knee extension. As with passive orthotic-based gait, hip stability is provided by the C-posture. The quadriceps and common peroneal stimulation channels are sometimes supplemented with a gluteal stimulation channel, which can provide additional stability to the hips and increase stride length by providing hip extension. FES-aided gait with surface electrodes is therefore provided with as few as four to six channels of stimulation. Variations and advances incorporating this flexion withdrawal reflex-based surface electrode protocol have been developed and evaluated by several groups [35-38]. Based on the protocol of Bajd and Kralj, an FES gait restoration system called ParaStep was introduced on the commercial market in 1994 (http://www.sigmedics.com/TheParastep/theparastep.html). The system has since

been fit to between approximately 400 [39] and 750 [40] SCI users. Studies of the system indicate that it is useful for standing and walking [26, 27, 41, 42] and for providing several psychological and physiological therapeutic benefits, including maintaining cardiovascular health, increasing physical self-concept, and decreasing depression [25, 26, 43].

Although FES-based systems have shown promise, electrically stimulated muscles suffer from rapid fatigue [7], relatively poor controllability of stimulated muscles, and the inability to control degrees-of-freedom outside the sagittal plane. Hybrid systems, which combine FES with an orthosis, can address several of the shortcomings of FES-aided gait. Use of an orthosis reduces the duty cycle of muscle stimulation during stance, helps to reduce the complexity of gait by imposing kinematic constraints, and enables closed-loop control of electrically stimulated muscle by providing a mounting platform for joint angle sensors. Recent research in hybrid FES gait restoration systems using surface and implanted electrodes include [6, 9, 10, 13, 16] and [14, 15, 20], respectively.

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CHAPTER II

Preliminary Evaluation of a Powered Lower Limb Orthosis to Aid Walking in Paraplegic Individuals

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Abstract

This paper describes a powered lower-limb orthosis that is intended to provide gait assistance to spinal cord injured (SCI) individuals by providing assistive torques at both hip and knee joints. The orthosis has a mass of 12 kg and is capable of providing maximum joint torques of 40 Nm with hip and knee joint ranges of motion from 105° flexion to 30° extension and 105° flexion to 10° hyperextension, respectively. A custom distributed embedded system controls the orthosis with power being provided by a lithium polymer battery which provides power for one hour of continuous walking. In order to demonstrate the ability of the orthosis to assist walking, the orthosis was experimentally implemented on a paraplegic subject with a T10 complete injury. Data collected during walking indicates a high degree of step-to-step repeatability of hip and knee trajectories (as enforced by the orthosis) and an average walking speed of 0.8 km/hr. The electrical power required at each hip and knee joint during gait was approximately 25 W and 27 W, respectively, contributing to the 117 W overall electrical power required by the device during walking. A video of walking corresponding to the aforementioned data is included in the supplemental material.

Introduction

There are currently about 262,000 spinal cord injured (SCI) individuals in the United States, with roughly 12,000 new injuries sustained each year at an average age of injury of 40.2 years [1]. Of these, approximately 44% (5300 cases per year) result in paraplegia. One of the most significant impairments resulting from paraplegia is the loss of mobility, particularly given the relatively young age at which such injuries occur. Surveys of persons with paraplegia indicate that mobility concerns are among the most prevalent [2], and that chief among mobility desires is the ability to walk and stand [3]. In addition to impaired mobility, the inability to stand and walk entails severe physiological effects, including muscular atrophy, loss of bone mineral content, frequent skin breakdown problems, increased incidence of urinary tract infection, muscle spasticity, impaired lymphatic and vascular circulation, impaired digestive operation, and reduced respiratory and cardiovascular capacities [4].

In an effort to restore some degree of legged mobility to individuals with paraplegia, several lower limb orthoses have been developed and described in the literature. The following literature review focuses

on orthoses that were developed specifically for restoration of mobility in paraplegic individuals. For recent surveys that consider passive and powered exoskeletons in a more general context, the reader is referred to [5-7]. Also, it should be noted that considerable research has been conducted on the use of functional electrical stimulation (FES) to restore legged mobility to paraplegics, although this topic is also not reviewed here. For a recent review of progress in FES-based gait restoration, the reader is referred to [8].   A number of passive orthoses have been developed to restore legged mobility to paraplegics. The simplest form of passive orthotics are long-leg braces that incorporate a pair of ankle-foot orthoses (AFOs) to provide support at the ankles, which are coupled with leg braces that lock the knee joints in full extension. The hips are typically stabilized by the tension in the ligaments and musculature on the anterior aspect of the pelvis. Since almost all energy for movement is provided by the upper body, these (passive) orthoses require considerable upper body strength and a high level of physical exertion, and provide very slow walking speeds. The hip guidance orthosis (HGO), which is a variation on long-leg braces, is described in [9-11]. The HGO incorporates hip joints that rigidly resist hip adduction and abduction, and rigid shoe plates that provide increased center of gravity elevation at toe-off, thus enabling a greater degree of forward progression per stride. Another variation on the long-leg orthosis, the reciprocating gait orthosis (RGO), incorporates a kinematic constraint that links hip flexion of one leg with hip extension of the other, typically by means of a push-pull cable assembly. As with other passive orthoses, the user leans forward against the stability aid while unweighting the swing leg and utilizing gravity to provide hip extension of the stance leg. Since motion of the hip joints is reciprocally coupled through the reciprocating mechanism, the gravity-induced hip extension also provides contralateral hip flexion (of the swing leg), such that the stride length of gait is increased. Examples of this type of orthosis, and studies of its efficacy, are described in [12-19]. In [20, 21], the authors describe a variation on the RGO, which incorporates a hydraulic-circuit-based variable coupling between the left and right hip joints. Experiments presented in [21] indicate improved hip kinematics with the modulated hydraulic coupling.

In order to decrease the high level of exertion associated with passive orthoses, some researchers have investigated the use of powered orthoses, which incorporate actuators and an associated power supply to assist with locomotion. Historical efforts to develop powered orthoses to aid in paraplegic mobility include [22-24]. More recently, [25] developed a powered orthosis for evaluating

design requirements for paraplegic gait assistance. In [26-28], a powered orthosis was developed by combining three electric motors with an RGO, two of which are located at the knee joints to enable knee flexion and extension during swing, and one of which assists the hip coupling, which in essence assists both stance hip extension and contralateral swing hip flexion. The orthosis was shown to increase gait speed and decrease compensatory motions, relative to walking without powered assistance. In [29-32], the authors describe control methods for providing assistive maneuvers (sit-to-stand, stand-to-sit, and walking) to paraplegic individuals with the powered lower limb orthosis HAL, which is an emerging commercial device with (in the incarnation utilized in the aforementioned publications) six electric motors (i.e., powered sagittal plane hip, knee, and ankle joints). Like the powered lower limb orthosis HAL, two additional emerging commercial devices include the ReWalk powered orthosis (Argo Medical Technologies) and the eLEGS powered orthosis (Berkeley Bionics). Both of these devices were developed specifically for use with paraplegic individuals, although no studies have been published characterizing the performance of these devices, or discussing their efficacy.

This paper describes a powered lower limb orthosis that, like the devices already mentioned, is intended to provide gait assistance to paraplegics by providing sagittal plane assistive torques at both hip and knee joints. Although as previously stated, studies have yet to be published providing technical information on ReWalk and eLEGS, the device described here is at minimum different in the fact that it neither includes a portion that is worn over the shoulders, nor a portion that is worn under the shoes. Also, the device described here has a significantly lower mass relative to the respective masses reported for the other two devices to date in the popular media. This paper describes the salient features and characteristics of the orthosis, and discusses the experimental implementation of the orthosis on a paraplegic subject (T10 complete). Data is presented characterizing the (hip and knee) joint angle trajectories during walking, the step-to-step repeatability of these trajectories (as enforced by the orthosis), and the average walking speed resulting from these trajectories. Additional data characterizes the electrical power consumption and corresponding battery life associated with the walking trials.

The Vanderbilt Powered Orthosis

The powered lower limb orthosis, shown in Figs. 2-1 and 2-2, was designed to provide powered assistance in the sagittal plane at both hip and knee joints. Each joint is powered by a brushless DC motor through a 24:1 gear reduction, which provides each joint with a maximum continuous joint torque of 12 Nm, and shorter duration maximum torques of approximately 40 Nm. The knee motors are additionally equipped with electrically controllable normally locked brakes, such that the knee joints remain locked in the event of a power failure. Each brake consists of a spring-loaded solenoid which, in the absence of power, provides torsional resistance in a drum brake configuration. The brake remains locked during the stance phase of gait and is released during the swing phase of gait and also during sit-to-stand and stand-to-sit transitions by energizing the solenoid. The range of motion for the hip joints is 105º in flexion and 30º in extension, while the range of motion for the knee joints is 105º in flexion and 10º in hyperextension. Hip ab/adduction is accommodated by compliance embedded into the hip segment. Such compliance is intended to provide stability to the wearer, while disallowing excessive adduction during swing, in order to prevent scissoring during walking. The orthosis is intended to be worn in conjunction with a standard ankle foot orthosis (AFO), which provides support at the ankle and prevents foot drop during swing. The structure of the orthosis is a composite of thermoplastic reinforced and supplemented with aluminum inserts. Sensors in the orthosis include potentiometers in both hip and knee joints, in addition to accelerometers located in each thigh link. As shown in Figs. 2-1 and 2-2, hook-and-loop straps on the hip segment, thigh segments, and shank segments secure the orthosis to the user with integrated padding in place to distribute pressure from the straps and protect the wearer from skin abrasion. The total orthosis mass is 12 kg (26.5 lbs). The orthosis is prevented from sliding down the body primarily by the two thigh straps immediately above the knee joints, and by the orthosis hip segment, which affixed around the subject's waist, just above the gluteal musculature. A mass breakdown showing individual component masses of the orthosis is given in Table 2-1.

Figure 2-1. Vanderbilt gait restoration orthosis oblique view.

Figure 2-2. Vanderbilt gait restoration orthosis frontal view.

Component	Mass (kg)	Mass Distribution
Joint Actuation	3.57	30%
Thigh Frames	4.08	34%
Hip Brace	2.10	17%
Shank Frames	1.09	9%
Battery	0.68	6%
Electronics	0.50	4%
Total	**12.02**	**100%**

Table 2-1 Mass breakdown of Vanderbilt Orthosis.

The powered orthosis additionally includes a custom distributed embedded system (DES), the components of which are located in the hip and both thigh segments. A functional diagram of the DES is given in Fig. 2-3. The DES is powered by a 29.6 V, 3.9 A•hr lithium polymer battery, and as indicated in Fig. 2-3, includes a power management module, a computation module, electronic signal conditioning and sensor interface module, power electronics, and communication electronics to interface components within the DES and between the DES and a host computer. The power management module provides, from the battery, linearly regulated ±12 and +3.3 V, which are used for signal conditioning and computation, and are derived from intermediate ±12.5 and +5 V switching regulators for efficient conversion. The main computational modules consist of two 80 MHz PIC32 microcontrollers, each with 512 kB flash memory and 32 kB RAM, and each of which consume approximately 400 mW of power. The microcontrollers are programmed in C using MPLAB IDE and the MP32 C Compiler (both from Microchip Technology, Inc.). A control tether connects the microcontrollers on the orthosis to a laptop computer via an RS-232 interface, such that the orthosis control can be supervised by the laptop host via the real-time interface provided by MATLAB Simulink RealTime Workshop. The two microcontrollers drive the brushless motors via four-quadrant switching servoamplifiers, and also drive the knee brakes via pulse-width-modulated (PWM) power transistors. One of the two main (twin) DES boards is shown mounted within the thigh link in Fig. 2-4.
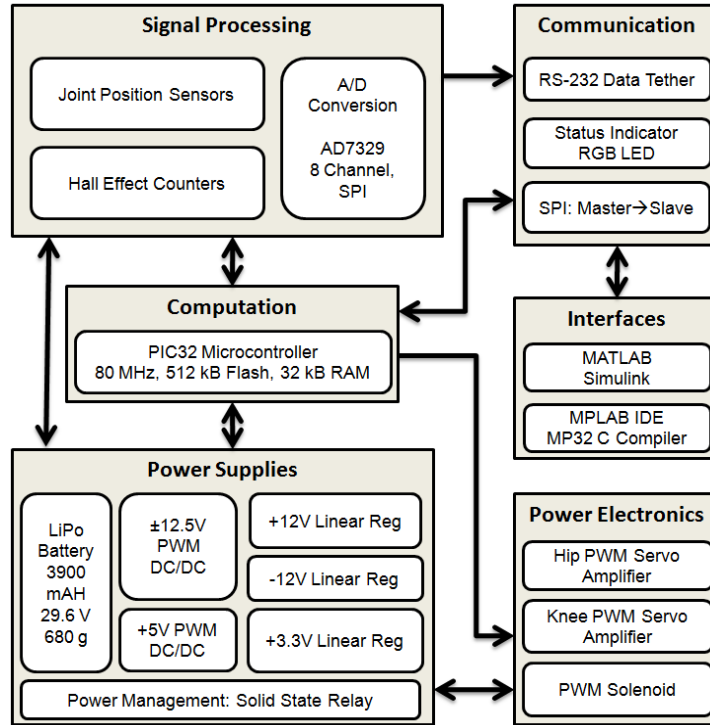
Figure 2-3. Embedded system framework.

Figure 2-4. Embedded system circuit board.

Orthosis Control

The orthosis controller consists of a state-flow system with four states, as shown in Fig. 2-5. Each state is defined by a set of joint angle trajectories, which are enforced by high-gain proportional-derivative (PD) control loops. Joint angle trajectories were preprogrammed for each motion based on normal biomechanical walking trajectories, obtained from a recording of the joint angle trajectories generated by a healthy subject while wearing the orthosis. For the data presented herein, switching between states was initiated by voice command of the user. The voice commands were keyed into the laptop host computer via an operator, which moved the state machine from one state to the next.

Figure 2-5. State-flow diagram. Boxes represent static states. Transitions are labeled between states, where LFS = left full-step, LHS = left half-step, RFS = right full-step, RHS = right half-step.

Experimental Implementation

In order to substantiate the ability of the powered orthosis to provide gait assistance, the previously described orthosis and controller was implemented on a paraplegic subject. The subject was a 35-year-old male (1.85 m, 73 kg) with a T10 complete injury, 8 years post injury. The evaluations were conducted at the Shepherd Center (Atlanta GA, USA), a rehabilitation hospital specializing in spinal cord injury. The testing was approved by both the respective Vanderbilt University and Shepherd Center Institutional Review Boards. All evaluations described herein were conducted within a standard set of parallel bars. Figure 2-6 shows the test subject wearing the orthosis while standing and walking, respectively, in the parallel bars. For the data presented in the subsequent sections, the evaluation protocol was as follows. The subject stood from a wheelchair with footrests removed by issuing a "stand" voice command. Note that the footrests, if not removed, would obstruct the subject's ability to bring his feet close to the chair, and therefore would impede his ability to transition from sitting to standing. Once comfortable standing,

the subject issued either a "left-step" or "right-step" voice command, and subsequently, a "step" voice command to initiate subsequent steps. Once near the end of the parallel bars, the subject issued a "half-step" command, which returned him to the standing configuration. The subject then turned in place in the parallel bars by lifting his weight with his arms and incrementally twisting around in order to walk in the opposite direction. This process repeated, typically for four to eight lengths of the parallel bars, at which point the subject sat (in his wheelchair, by issuing a "sit" voice command), so that data from the walking trial could be recorded (i.e., uploaded and saved to the host computer). A video depicting a lap of walking is provided in the supplemental material that accompanies this paper. Data indicating hip and knee joint angles and electrical power consumption corresponding to over ground walking are given in the following subsections.

Figure 2-6. Subject with T10 complete spinal cord injury walking wearing the powered orthosis.

Gait Kinematics

Figure 2-7 shows measured joint angle data from 23 right steps and 23 left steps, overlaid onto the same plot. Note that an approximate one-second delay exists between each right and left step, during which time the subject adjusted his upper body in preparation for commanding the next step. Figure 2-8 shows the same data shown in Fig. 2-7, but with the delay between steps replaced with a vertical dashed line (which indicates a discontinuity in time), with the time base replaced with a percent stride base, and with the left and right joint angles overlaid onto the same plots. In this manner, the knee and hip joint angles can be qualitatively compared to standard joint kinematics during walking, which is typically represented as a function of stride. These normal biomechanical trajectories (taken from [44]) are also plotted in Fig. 2-8 as dashed lines. The repeatability of the joint angle data over these 23 strides, and the similarity of

such data to normal biomechanics (particularly with respect to the amplitude of knee flexion, and the amplitude of hip flexion and extension), indicate that the powered orthosis is able to provide appropriate and repeatable gait assistance to the user during walking. The gait represented by this data is characterized by an average overground walking speed of 0.22 m/s (0.8 km/hr or 0.5 mi/hr).
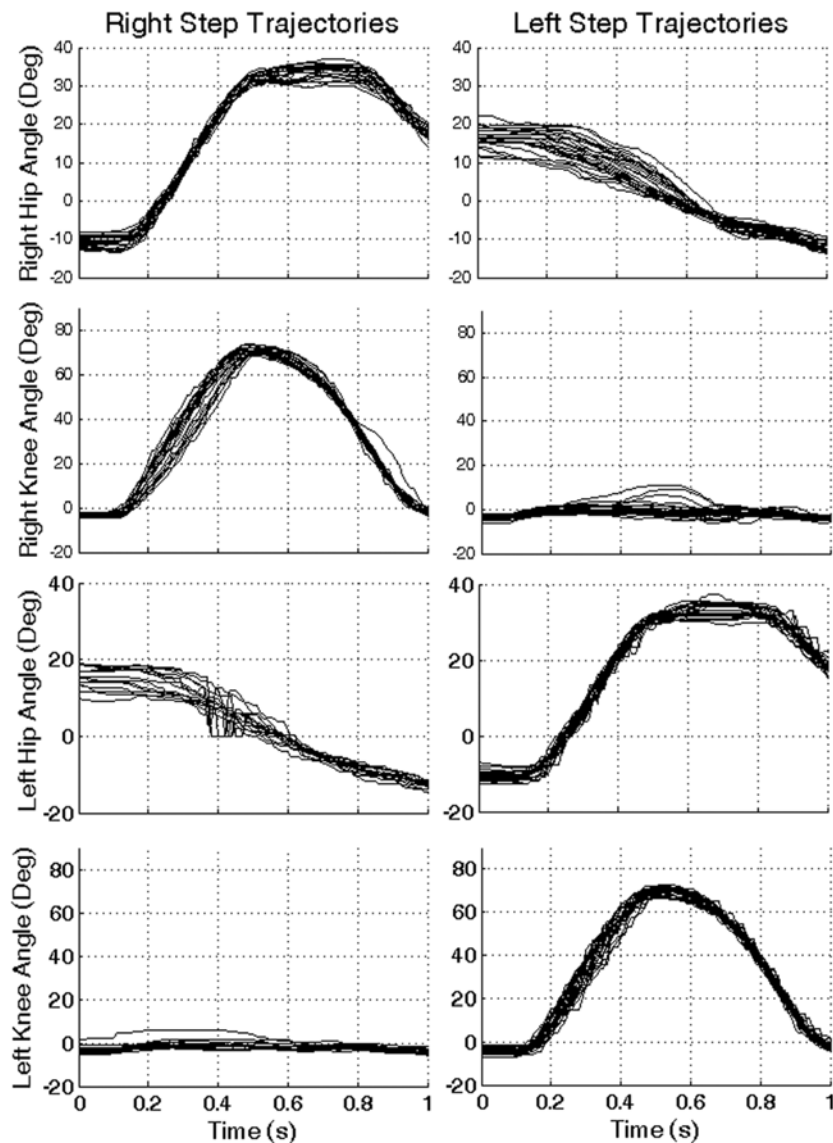
Figure 2-7. Measured joint angles during 23 right steps and 23 left steps.
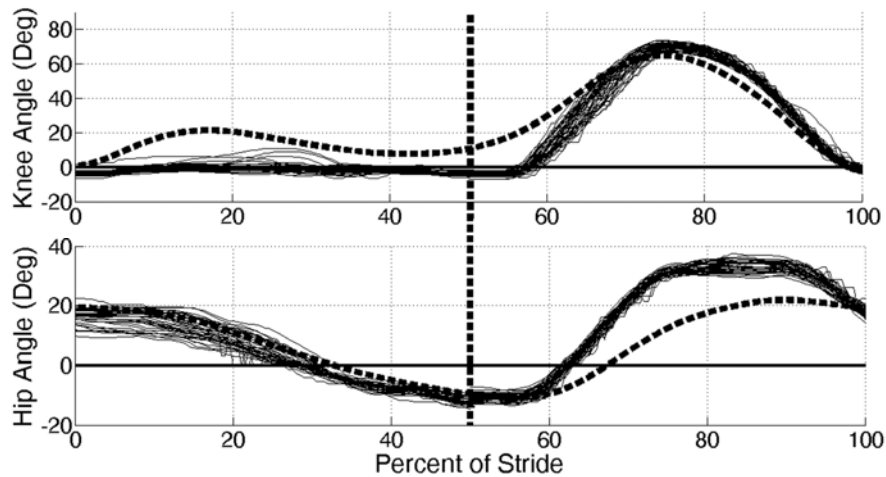
Figure 2-8. Overlaid joint angles for 23 gait cycles. The dashed vertical line represents an approximately one-second pause taken by the subject between steps. The dashed trajectories represent normal biomechanical data for level ground walking at a medium cadence.

Electrical Power Consumption

Electrical power consumption was recorded during the walking represented by Figs. 2-7 and 2-8. The electrical power required by the servoamplifiers, corresponding to the data shown in Fig. 2-8, is shown averaged over all 46 steps (or 23 strides) in Fig. 2-9, which represents an average power consumption of approximately 35 W for each knee actuator (during the active stride), and approximately 22 W for each hip actuator (during the stride). In addition to requiring electrical power during right and left steps, the joint actuators also used power to maintain joint stiffness in the double support states (i.e., while the subject shifted his weight to prepare for the next step). For the 46 step sequence previously described, the total electrical power required by each actuator was 27 W on average for each knee motor and 21 W on average for each hip motor during the swing phase of gait, and 26 W and 29 W of average power for the knee and hip motors, respectively, during the stance phase of gait. The knee brakes additionally required

on average approximately 7 W of electrical power during swing, but did not require any power during stance (i.e., they are normally locked brakes). Finally, the average electrical power required by the remainder of the distributed embedded system was measured as 7.2 W. The average measured electrical power consumption for each component and each phase of the walking cycle is summarized in Table 2-2. With a one-second average pause between steps (corresponding to the 0.22 m/s walking data represented by Fig. 2-8), the total electrical power required by the system was 117 W. Recall that the battery pack included in the powered orthosis prototype described herein is a 680 g lithium polymer battery with a 115 W-hr capacity. Based on the walking data of Fig. 2-9 and Table 2-2, the battery would provide approximately one hour of continuous walking between charges. At the previously stated (measured) average overground speed of 0.8 km/hr (0.5 mi/hr), the powered orthosis would provide a range of approximately 0.8 km (0.5 mi) between battery charges. Note that the range could be easily increased, if desired, without incurring a significant mass penalty, by increasing the size of the battery, which currently constitutes 6% of the system mass (see Table 2-1). For example, doubling the size of the battery pack would double the range and result in an overall device mass of 12.7 kg, as opposed to 12 kg, as implemented here.
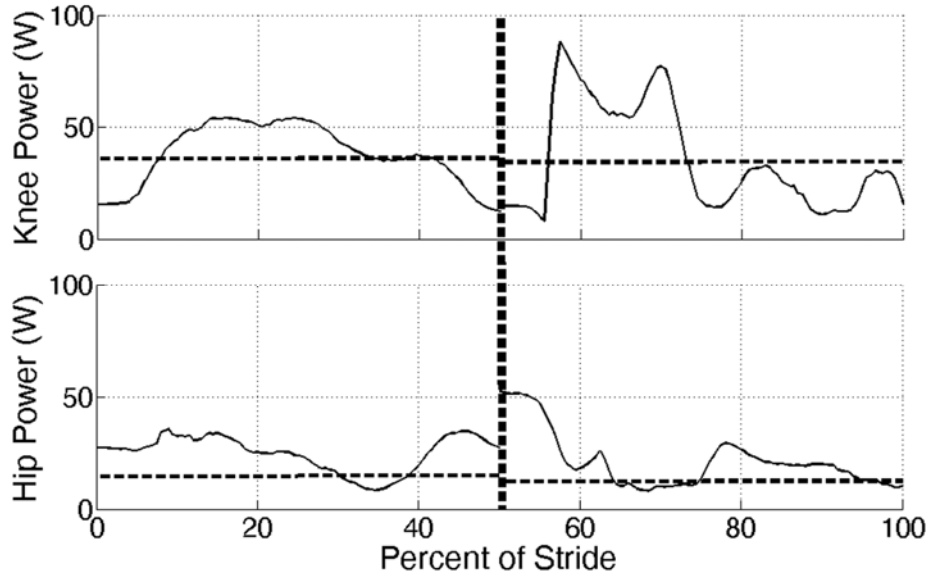
Figure 2-9. Average measured electrical power at the knee and hip joints over one gait cycle (average taken from 23 gait cycles). *The dashed vertical line represents a pause of approximately a second which the subject would take in between steps. The dashed horizontal lines represent average electrical power within one gait cycle.

Component	Power During Stepping (W)	Power Between Steps (W)	*Average Power During Walking (W)
Swing Knee Motor	34.8	19.6	27.2
Stance Knee Motor	35.6	16.5	26.1
Swing Hip Motor	21.4	19.8	20.6
Stance Hip Motor	23.9	34.0	29.0
Embedded Electronics Boards	7.2	7.2	7.2
Swing Knee Brake	13.5	0	6.7
Stance Knee Brake	0	0	0
Total	**136.4**	**97.1**	**116.8**

*Average power assumes a 1 second pause in between steps (i.e., steps are being taken during 50% of the time during "walking")

Table 2-2 Vanderbilt Orthosis electrical power consumption.

Audible Sound Level

A digital sound level meter was used while walking with the orthosis. The average sound level, as measured one meter away from the orthosis, was 55 ± 2 dBA (with an ambient noise level of 38 dBA).

<u>Discussion</u>

The powered orthosis has been developed with a strong focus on ergonomics and user acceptance. High priority was given to low mass and minimal body coverage. Additionally, the authors have attempted to minimize the profile of the orthosis in the frontal plane, which adds 3.2 cm at the hip and knee joint, and 4.8 cm at mid-thigh, such that a user is able to sit in an armchair or wheelchair. Similarly, the hip segment protrudes approximately 3.2 cm posteriorly from the user's lower back, such that it should not significantly interfere with a seat back. The orthosis does not extend above mid-abdomen and requires nothing to be worn over the shoulders and nothing above the lower back, which presumably renders the device less noticeable when sitting at a desk or table. The compact design of the orthosis is greatly facilitated by the integration of the distributed embedded system within the orthosis structure. With quick disconnects at each hip joint, the orthosis is easily separated into three modular components – right leg, left leg, and hip segment – for ease of donning and doffing and also for increased portability. An important aspect of the device is the normally locked knee brakes, which presumably provide a fail-safe in the event of power loss (e.g., a fully depleted battery). The combination of low joint impedances at the hip in the absence of control, along with locked knees, renders the device the essential equivalent of a pair of long-leg braces in the event of power loss, such that the user is neither in danger, nor completely immobilized by such an unplanned occurrence.

One trade-off associated with the design of the orthosis is the need for custom fitting relative to users of different sizes. Of particular importance is that the sagittal plane centers of rotation of the orthosis joints be concentric with the approximate corresponding centers of rotation of the user's physiological hip and knee joints, and the width of the hip piece correspond to the width of the user's hips. In the absence of shoulder straps or structure under the shoes, the orthosis must fit the wearer well in

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order to effectively support the weight of the orthosis and provide the desired functionality.

Conclusion

This paper describes a powered lower limb orthosis developed to assist gait in spinal cord injured individuals. Experimental results from walking trials with a T10 complete paraplegic indicate that the orthosis is capable of providing a repeatable gait with knee and hip joint amplitudes that are similar to those observed during non-SCI walking. Electrical power measurements with the current battery pack and control algorithms indicate a battery life of approximately an hour, and a corresponding walking range of approximately 0.8 km. The authors expect this range to increase with efficiency in enhancements in the control algorithms, and can of course be most easily increased with a larger battery pack, which (given the size of the current battery pack), is unlikely to significantly increase the mass or size of the device. Future work includes the addition of automated, sensor-based gait mode transitions, such that the device can operate without voice commands.

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CHAPTER III

A Method for the Autonomous Control of Lower Limb Exoskeletons for Persons with Paraplegia

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Abstract

Efforts have recently been reported by several research groups on the development of computer-controlled lower limb orthoses to enable legged locomotion in persons with paraplegia. Such systems must employ a control framework that provides essential movements to the paraplegic user (i.e., sitting, standing, and walking), and ideally enable the user to autonomously command these various movements in a safe, reliable, and intuitive manner. This paper describes a control method that enables a paraplegic user to perform sitting, standing, and walking movements, which are commanded based on postural information measured by the device. The proposed user interface and control structure was implemented on a powered lower limb orthosis, and the system was tested on a paraplegic subject with a T10 complete injury. Experimental data is presented that indicates the ability of the proposed control architecture to provide appropriate user-initiated control of sitting, standing, and walking. The authors also provide a link to a video that qualitatively demonstrates the user's ability to independently control basic movements via the proposed control method.

Introduction

One of the most significant impairments resulting from paraplegia is the loss of mobility, particularly given the relatively young age at which such injuries occur [1-3]. In addition to diminished mobility, the inability to stand and walk entails significant physiological impairments, including muscular atrophy, loss of bone mineral content, frequent skin breakdown problems, increased incidence of urinary tract infection, muscle spasticity, impaired lymphatic and vascular circulation, impaired digestive operation, and reduced respiratory and cardiovascular capacities [4].

In an effort to facilitate legged locomotion in individuals with paraplegia, several computer-controlled lower limb orthosis systems have been, and are being, developed and described in the research literature. Some of these include hybrid FES-systems, which combine a computer-controlled orthosis with computer-controlled functional electrical stimulation (FES) of leg muscles, such as the systems described by [5-9]. Recently, a number of powered lower limb orthoses, or exoskeletons, have also been described for purposes of gait assistance for persons with paraplegia, including those described by [10-16]. In addition to these systems, two other exoskeleton systems, developed by

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commercial entities, are those by Argo Medical Technologies (ReWalk) and Berkeley Bionics (eLEGS). Technical information regarding these two systems have not yet appeared in the engineering literature.

In the aforementioned publications describing computer-controlled orthoses (i.e., [5-16]), the authors focus on the capacity of their respective systems to provide legged mobility, but do not focus on specifically on control methods that enable the user to autonomously command various movements. In order to demonstrate mobility, these approaches have either incorporated push-button controls on the stability aid, or have incorporated a system operator, who operates the system (e.g., from a host computer) on behalf of the paraplegic individual. The emerging commercial systems, ReWalk and eLEGS, presumably provide for autonomous user control. To the authors' knowledge, however, no information has been published in the engineering literature regarding the control methods incorporated by either of these systems. Based on product information available from the respective companies, the ReWalk exoskeleton appears to utilize a tilt-sensor on the torso to gate subsequent steps while walking, and utilizes a wrist-mounted keypad to select between other types of movements. The eLEGS exoskeleton reportedly utilizes instrumentation on the forearm crutches or walker to gate subsequent steps while walking.

To the authors' knowledge, no publication in the engineering literature has described and demonstrated a method that enables a paraplegic user to intuitively and autonomously control (i.e., without push-button controls or the assistance of a system operator) the basic movements associated with legged mobility (i.e., sitting, standing, and walking). As such, this paper presents a control architecture for a powered lower limb orthosis (or exoskeleton) designed to enable a paraplegic user to autonomously navigate through these movements, without the use of their hands or the aid of an external operator. Specifically, the control architecture enables the user to switch between sitting, standing, and walking, based on the user's upper body movement. The control architecture was implemented on a powered lower limb orthosis and evaluated on a paraplegic subject with a T10 motor and sensory complete injury (i.e., American Spinal Injury Association, ASIA, A classification). The ability of the user to autonomously control the system was assessed by having the paraplegic user repeatedly perform a timed-up-and-go (TUG) test, which is a standard clinical measure of legged mobility [17]. The paper describes the control architecture and its implementation, and presents experimental results of the TUG

tests. The test results support the ability of the proposed control architecture to enable user-autonomous control of the basic movements associated with legged mobility.

Powered Orthosis Prototype

Although the proposed control interface is generally applicable to a number of computer-controlled lower limb orthoses (such as those previously described), for purposes of this paper, the architecture was implemented on the powered lower limb orthosis shown in Fig. 3-1. Specifically, the orthosis shown in Fig. 3-1 incorporates four motors, which impose sagittal plane torques at each hip and knee joints. As seen in the figure, the orthosis contains five segments, which are: two shank segments, two thigh segments, and one hip segment. Each thigh piece contains two brushless DC motors which are used to drive the hip and knee articulations through a speed-reduction transmission. Each joint can provide up to 12 Nm of continuous torque and 40 Nm for shorter (i.e., 2-sec) durations. As a safety measure, both knee joints include normally locked brakes, in order to preclude knee buckling in the event of a power failure. The system does not contain foot or ankle components, but is designed to be used in conjunction with a standard ankle foot orthosis (AFO) to provide stability for the ankle, and to preclude foot drop during the swing phase of gait. Physical sensing in the orthosis consists of Hall-effect-based angle and angular velocity sensing in each hip and knee joint, and 3-axis accelerometers and single-axis gyroscopes in each thigh segment. A pair of microcontrollers located in the thigh segments, provide low-level control of the orthosis. In the current implementation, the microcontrollers communicate with a host computer via a data tether, which facilitates controller development and data visualization. All power on the orthosis is provided by a lithium polymer battery located in the hip segment (see Fig. 3-1). A functional schematic of the embedded system on the orthosis is shown in Fig. 3-2.
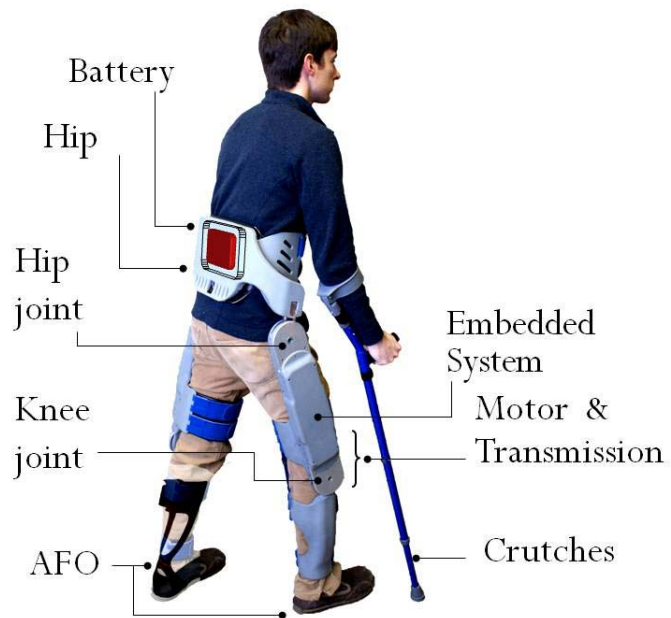
Figure 3-1. Powered lower limb orthosis.
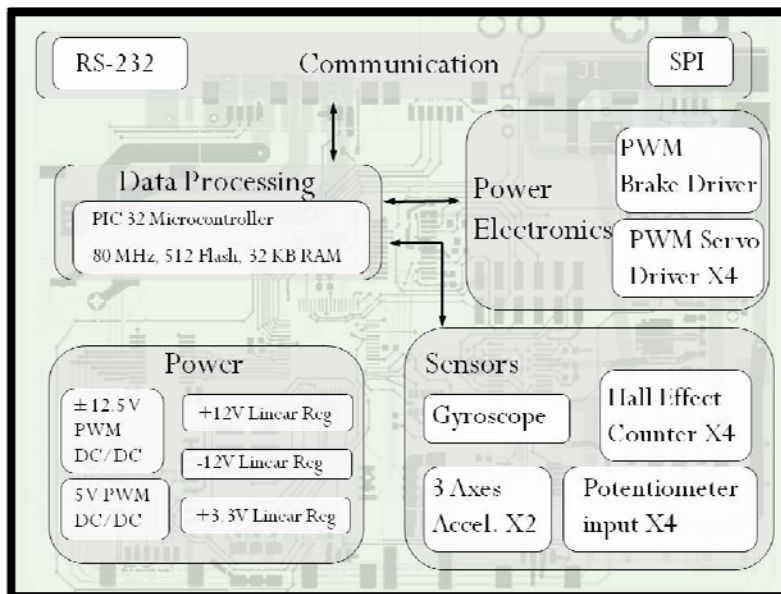
Figure 3-2. Functional schematic of embedded system.

Powered Orthosis Control Architecture

Joint-Level Controllers

The general control structure of the orthosis consists of variable-impedance joint-level controllers, the behavior of which is supervised by an event-driven finite-state controller. The joint-level controllers consist of variable-gain proportional-derivate (PD) feedback controllers around each (hip and knee) joint, where at any given time, the control inputs into each controller consists of the joint angle reference, in addition to the proportional and derivative gains of the feedback controller. Note that the latter are constrained to positive values, in order to ensure stability of the feedback controllers. With this control structure, in combination with the open-loop low output impedance of the orthosis joints, the joints can either be controlled in a high-impedance trajectory tracking mode, or in a (relatively) low-impedance mode, by emulating physical spring-damper couples at each joint. The former is used where it may be desirable to enforce a predetermined trajectory (e.g., during the swing phase of gait), while the latter is used when it may be preferable not to enforce a pre-determined joint trajectory, but rather to provide

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assistive torques that facilitate movement toward a given joint equilibrium point (as in transitioning from sitting to standing), or to impose dissipative behavior at the joint (as in transitioning from standing to sitting).

Finite-State Control Structure

The joint-level controller receives trajectory commands, as well as PD gains, from a supervisory finite-state machine (FSM), which (for sitting, standing, and walking) consists of 12 states, as shown in Fig. 3-3. The FSM consists of two types of states: static states and transition states. The static states consist of sitting (S1), standing (S2), right-leg-forward (RLF) double support (S3), and left-leg-forward (LLF) double support (S4). The remaining 8 states, which transition between the four static states, include sit-to-stand (S5), stand-to-sit (S6), stand-to-walk with right half step (S7), stand-to-walk with left half step (S11), walk-to-stand with left half step (S10), walk-to-stand with right half step (S12), right step (S9), and left step (S8).

Each state in the FSM is fully defined by the combination of a set of trajectories, and a set of joint feedback gains. In general, the latter are either high or low. The set of trajectories utilized in six of the eight transition states are shown in Fig. 3-4. For all the trajectories shown in Fig. 3-4, the joint feedback gains are set high. The final angles of the trajectories shown in Fig. 4 for the various joints define the constant joint angles that correspond to the static states of RLF double support (S3), LLF double support (S4), and standing (S2). Three states remain, which are the static state of sitting and the two transition states of sit-to-stand and stand-to-sit. The static state of sitting (S1) is defined by zero gains, and therefore the joint angles are unimportant. The transition from stand-to-sit (S6) consists of a zero proportional gain and a high derivate gain (i.e., damping without stiffness). Thus, the joint angles are also immaterial for this state, assuming they are constant. Finally, the sit-to-stand (S5) state is defined by standing (S2) joint angles, and utilizes a set of PD gains that ramp up from zero to a value that corresponds to a high impedance state. Table 3-1 and Fig. 3-4 summarize the trajectories and nature of the feedback gains that together define completely the behavior in all states of the FSM shown in Fig. 3-3.
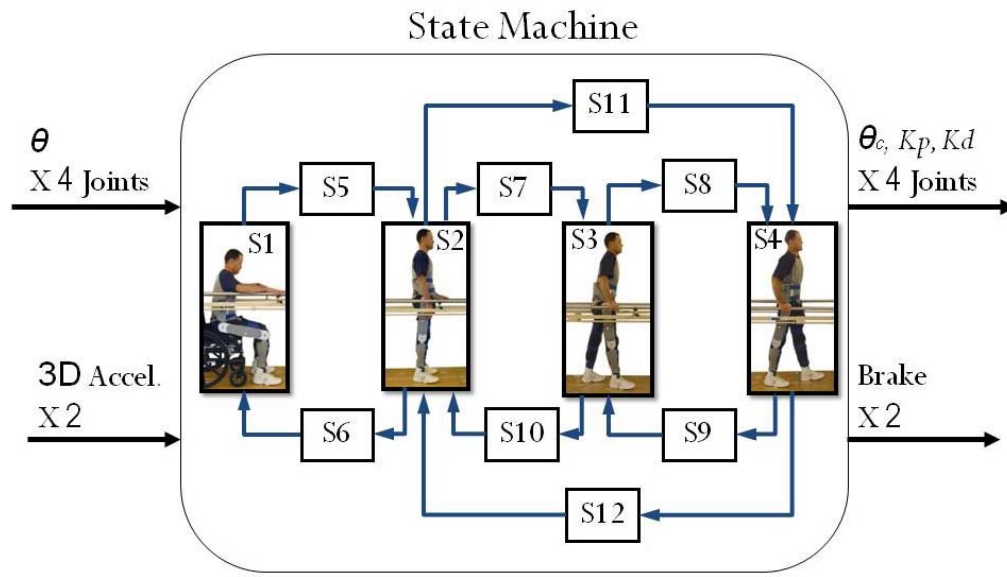
Figure 3-3. Finite state machine for sitting, standing, and walking.
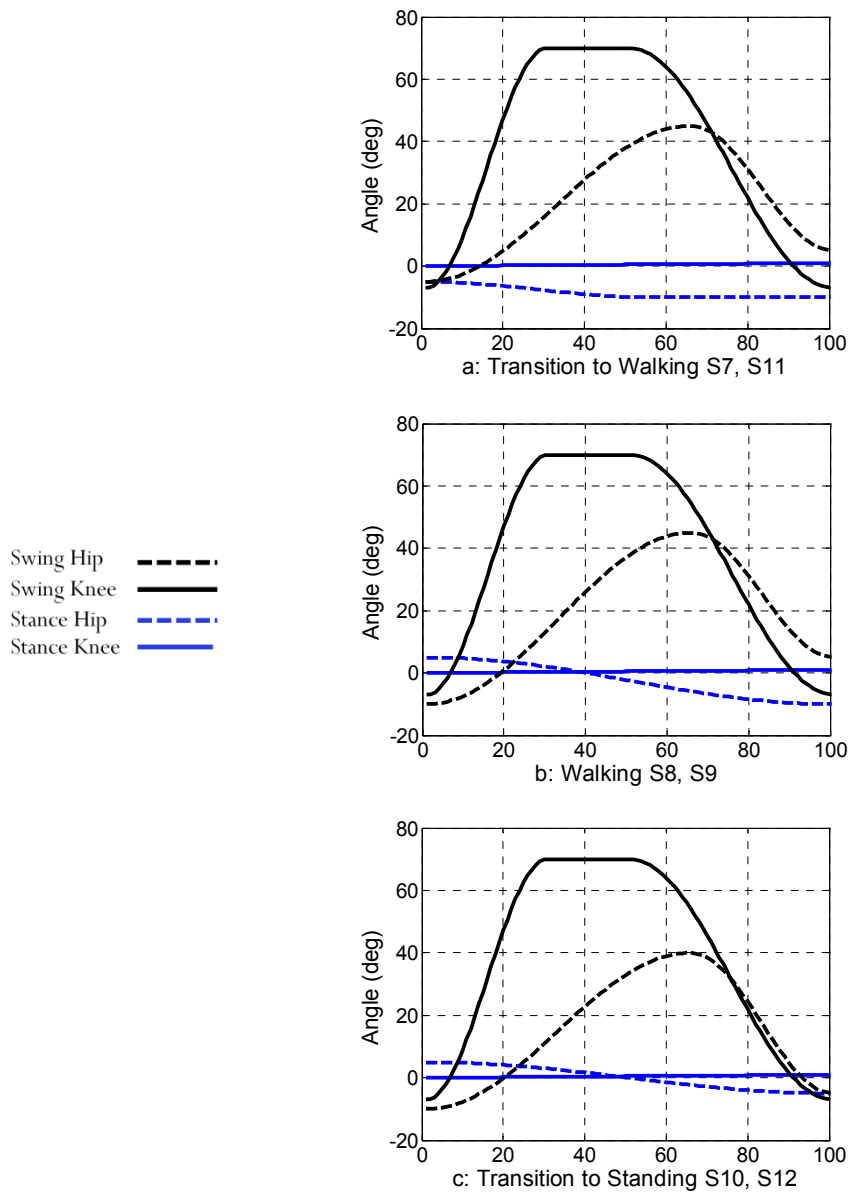
Figure 3-4. Walking trajectories corresponding to finite states as indicated.

CONTROL CHARACTERISTICS IN EACH STATE

State	Type	Gains	Control Priority
S1- Sitting	Static	Low	NA
S2- Standing	Static	High	Position
S3- Right Forward	Static	High	Position
S4- Left Forward	Static	High	Position
S5- 1 to 2	Transition	N.A	Gain
S6- 2 to 1	Transition	N.A	Gain
S7- 2 to 3	Transition	High	Trajectory
S8- 3 to 4	Transition	High	Trajectory
S9- 4 to 3	Transition	High	Trajectory
S10- 3 to 2	Transition	High	Trajectory
S11- 2 to 4	Transition	High	Trajectory
S12- 4 to 2	Transition	High	Trajectory

Table 3-1 Joint controller characteristics within each state.

Switching Between States

The volitional command of the basic movements in the FSM is based on the location of the (estimated) center of pressure (CoP), defined for the (assumed quasistatic user/orthosis) system as the center of mass projection onto the (assumed horizontal) ground plane. This notion is illustrated in Fig. 3-5, which indicates the approximate location of the CoP, relative to the forward-most heel. It is assumed that, with the use of the stability aid, the user can affect the posture of his or her upper body, and thus can affect the location of the CoP. By utilizing the accelerometers in the orthosis, which provide a measure of the thigh segment angle (α in Fig. 3-5) relative to the inertial reference frame (i.e., relative to the gravity vector), in combination with the joint angle sensors (which provide a measure of the configuration of the orthosis and user), the orthosis controller can estimate the location of the CoP (in the sagittal plane). More specifically, in this estimation, the authors assume level ground; that the heels remain on the ground; that the head, arms, and trunk (HAT) can be represented as a single segment with fixed inertial properties; and that out-of-sagittal-plane motion is small. Given these assumptions, along with estimates of the length, mass and location of center of mass of each segment (right and left shank, right and left

thigh, and HAT), the controller can estimate the projection of the CoP on the ground. Let the distance from the forward-most heel to the CoP be X_c, where a positive value indicates that the CoP lies anterior to the heel, and a negative number indicates the CoP lies posterior to the heel (see Fig. 3-5). From a state of double support (S3 or S4), the user commands the next step by moving the CoP forward, until it meets a prescribed threshold, at which point the FSM will enter either the right step or left step states, depending on which foot started forward. From a standing position (S2), the user commands a step by similarly moving the CoP forward until it meets a prescribed threshold, but also leaning to one side in the frontal plane (as indicated by the 3-axis accelerometers in the thigh segments), which indicates that the orthosis should step with the leg opposite the direction of frontal plane lean (i.e., step forward with the presumably unweighted leg). That is, leaning to the right (and moving the CoP forward) will initiate a left step, while leaning to the left (and moving the CoP forward) will initiate a right step. In order to transition from a standing state (S2) to a sitting state, the user shifts the CoP rearward, such that the CoP lies behind the user. Finally, to transition from a sitting to a standing state (S1 to S2), the user leans forward (as illustrated in Fig. 3-6a), which shifts the CoP forward to a predetermined threshold, which initiates the transition from sitting to standing. Note that the right portion of Fig. 3-6 shows the case where the user's CoP is not sufficiently forward to initiate a transition from sitting to standing. Finally the transition from (either case of) double support to standing (i.e., from either S3 or S4, to S2) is based on the timing associated with crossing the CoP threshold. That is, if the CoP does not cross the CoP threshold within a given time following heel strike (i.e., if the controller remains in either state S3 or S4 for a given duration), subsequent crossing of the CoP threshold will transition to standing (S2) rather than to the corresponding double support configuration. That is, a sufficient pause during gait indicates to the system that the user wishes to stand, rather than continue walking forward. A summary of all switching conditions, governing the user interface with the FSM controller, is given in Table 3-2.

The previous discussion indicates that the user-initiated right and left steps occur when the estimated location of the CoP (relative to the forward heel) exceeds a given threshold. The authors have found that this approach provides enhanced robustness when this threshold is a function of the step length. That is, despite high-gain trajectory control in the joints of the orthosis during swing phase, scuffing of the foot on the ground, as occasionally occurs, in combination with compliance in the orthosis

structure, can alter the step length during walking. In the case of a small step length, the forward thigh is nearly vertical, and the user is more easily able to move the CoP forward of the forward heel. In the case of a large step length, the forward thigh is forms a larger angle with the vertical, and moving the CoP forward is more difficult. As such, the CoP threshold during walking was constructed as a linear function, where the CoP threshold (i.e., the amount the CoP must lie ahead of the forward heel) decreases with increasing step size.
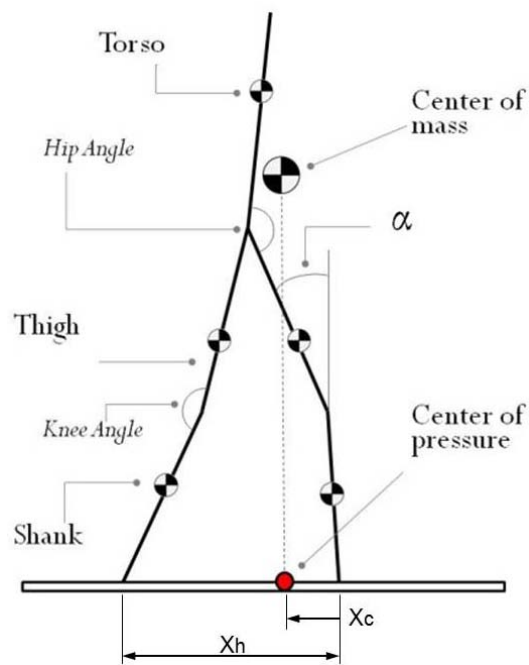
Figure 3-5. Schematic indicating estimated stride length length (X_h) and center of pressure (X_c), both estimated based on the configuration of the orthosis.
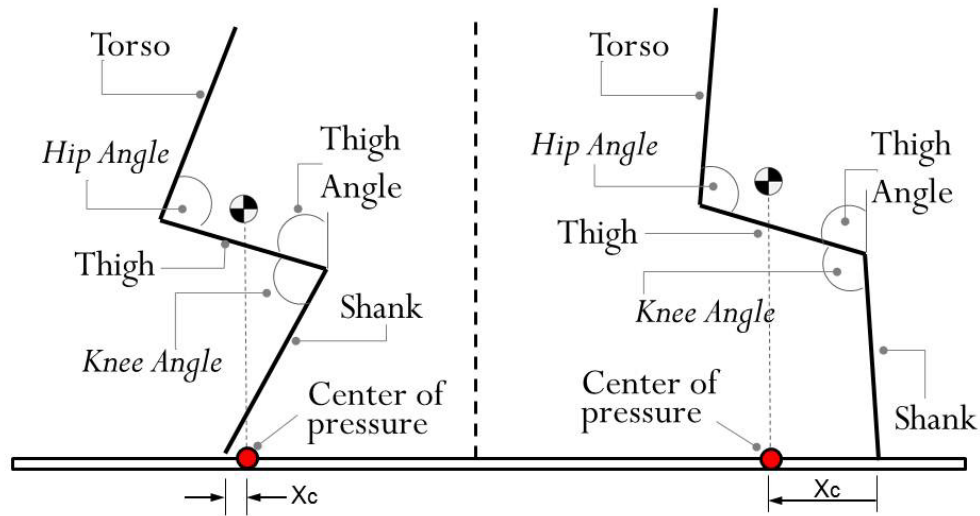
Figure 3-6. Schematic indicating the use of center of pressure (X_c) estimate for purposes of sit-to-stand and stand-to-sit transitions.

STATE MACHINE SWITCHING CONDITIONS

Transition	Condition
S1 to S5	The user leans forward and pushes up.
S5 to S2	Hip and knee joints meet the Standing (S2) configuration.
S2 to S7	The user leans forward and left.
S7 to S3	Hip and knee joints meet the Right Forward (S3) configuration.
S3 to S8	The user leans forward.
S8 to S4	Hip and knee joints meet the Left Forward (S4) configuration.
S4 to S9	The user leans forward.
S9 to S3	Hip and knee joints meet the Right Forward (S3) configuration.
S3 to S10	The user pauses for a predetermined period prior to leaning forward.
S10 to S2	Hip and knee joints meet the Standing (S2) configuration.
S2 to S6	The user leans backward.
S6 to S1	A predetermined time has lapsed.
S2 to S11	The user leans forward and right.
S11 to S4	Hip and knee joints meet the Left Forward (S4) configuration.
S4 to S12	The user pauses for a predetermined period prior to leaning forward.
S12 to S2	Hip and knee joints meet the Standing (S2) configuration.

Table 3-2. Walking trajectories corresponding to finite states as indicated.

The proposed control architecture (defined by Fig. 3-3, Table 3-1, and Table 3-2) was implemented on the previously described powered lower limb orthosis, and the ability of the system to enable a user to autonomously perform the basic movements associated with legged mobility (i.e., sitting, standing, and level walking) was assessed in preliminary trials conducted with a paraplegic subject. The subject was a 35-year-old male (1.85 m, 73 kg) with a T10 motor and sensory complete injury (i.e., ASIA A), 9 years post injury. The evaluations were conducted at the Shepherd Center (Atlanta GA, USA), a rehabilitation hospital specializing in spinal cord injury. The testing was approved by both the respective Vanderbilt University and Shepherd Center Institutional Review Boards. All data presented here corresponds to walking conducted using a walker as a stability aid. The subject is shown wearing the orthosis and using the walker in Fig. 3-7.

The ability of the powered orthosis and control architecture to provide autonomously commanded sitting, standing, and walking was assessed by having the subject autonomously perform a timed-up-and-go (TUG) test. The TUG test is a standard clinical measure for assessing legged mobility [45]. In this test the subject starts seated in a chair, and given a start command, stands up, walks forward three meters, turns around in place, walks back to the starting point, and sits down in the chair. In order to assess the ability of the subject to autonomously control movements of the orthosis, this test was repeated a number of times, until the subject was comfortable performing the test. Once comfortable with the task, the subject was asked to repeat the TUG test three times. The set of data that corresponds to the third of these three TUG tests is shown in Fig. 3-8. Specifically, the figure shows the right and left hip and knee joint angles corresponding to this TUG test, along with the corresponding states of the FSM. In the sequence, the user starts in the sitting state (S1), after which the system enters the sitting to standing mode (S5), in which both hips and both knees provide torques to facilitate joint extension. Following S5, the state history depicts a series of consecutive steps, followed by a period of standing (S2), during which the subject turned in place, with the aid of the walker. The first series of steps is then followed by a second series, during which the subject returned to the chair. Once at the chair, the subject again entered standing mode (S2), allowing the subject to turn in place, prior to returning to a seated position in the chair. A video of actual TUG test corresponding to this data can be viewed at:

Recall that the threshold for the CoP during walking is function of the step length. Figure 3-9 shows the system state, the estimated CoP (X_c), and the CoP switching threshold ($X_{\hat{c}}$) for several steps (of slightly varying length). As seen in the figure, the CoP threshold ($X_{\hat{c}}$) varies with step length (X_h). In general, when the CoP (X_c) exceeds the threshold at the end of the swing phase trajectory, the controller will switch immediately to the contralateral swing phase (i.e., switching between S8 and S9). If the CoP does not cross the CoP threshold at the end of swing phase, the controller will remain in the respective double support phase (S3 or S4) until the user shifts the CoP to cross the CoP threshold.

Figure 3-10 presents the sequences of finite states corresponding to each of the three TUG tests. The subject completed the three tests in 103, 128, and 112 s, respectively. The average time to complete the sequence was 114 s, with a standard deviation of 8.6 s (7.5%). The consistency between trials (i.e., standard deviation of ±7.5%) indicates that the control approach appeared to provide a repeatable means for the subject to control the basic movements associated with legged mobility.

Figure 3-7. Photographic sequence showing standing, a left step, and a right step.
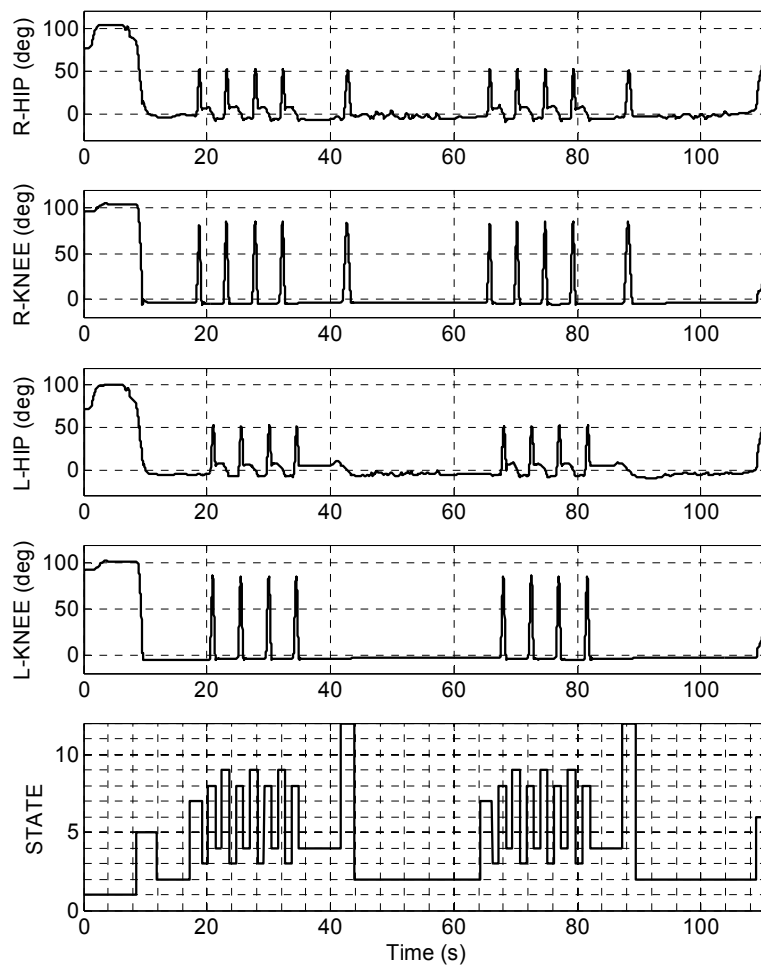
Figure 3-8. Joint angles and controller state during the third TUG test.
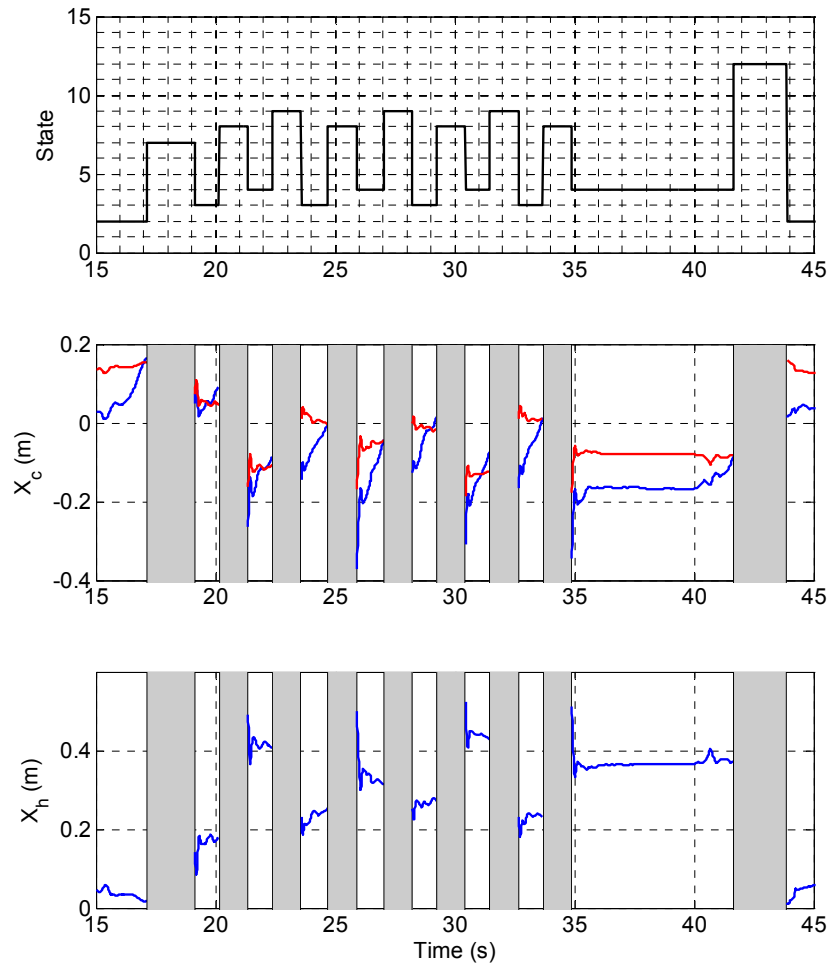
Figure 3-9. Data excerpted from Fig. 3-8. Top row: finite state corresponding to a sequence of steps. Middle row: center of pressure estimate (X_c, blue) and center of pressure threshold ($X_{\hat{c}}$, red). Bottom row: step length estimate (X_h).
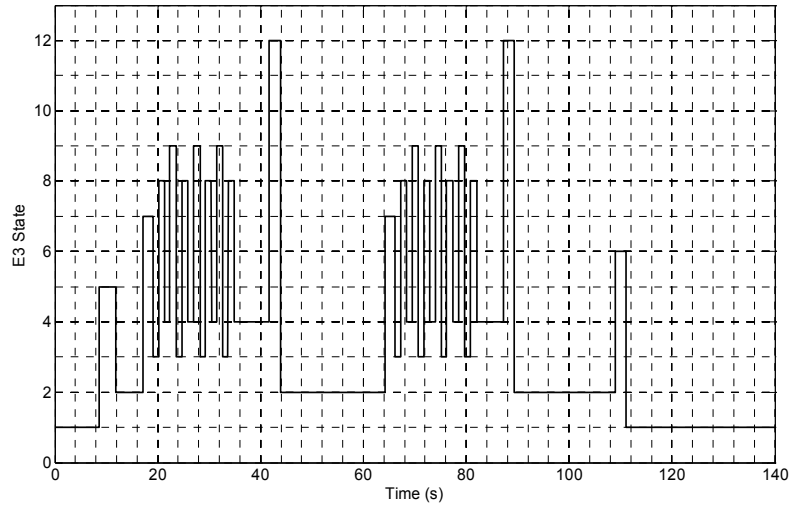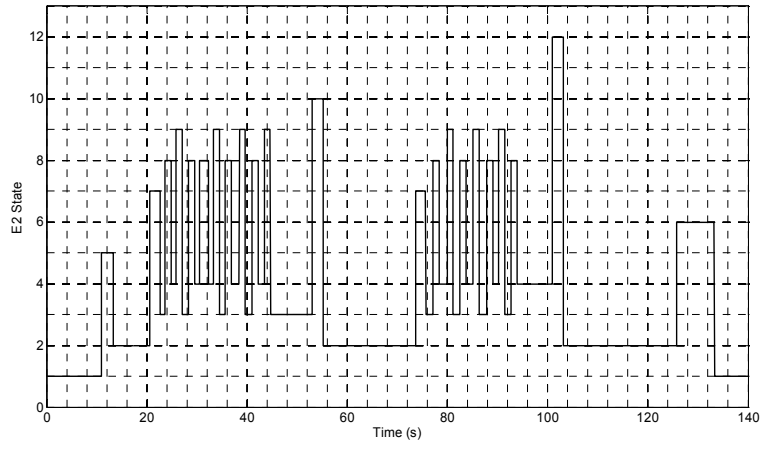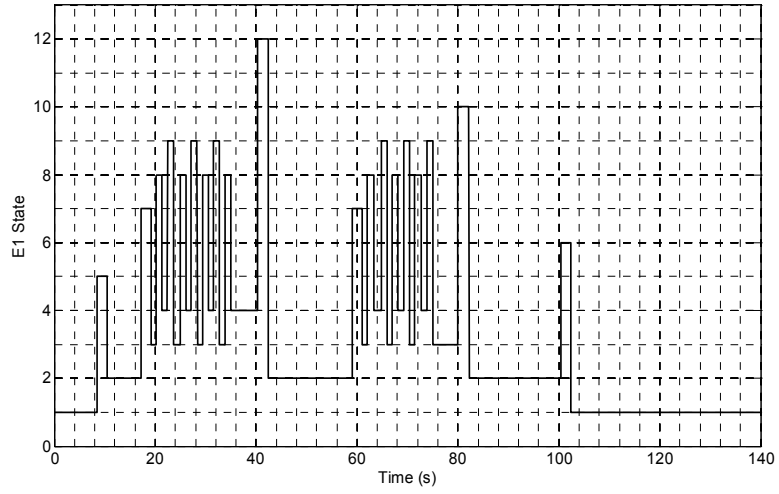
Figure 3-10. Finite states corresponding to each of the three TUG tests.

Conclusion

This paper presents a method for the control of a powered orthosis that enables autonomous (user-controlled) basic legged mobility, including sitting, standing, and walking, for persons with paraplegia (i.e., enables the user to autonomously navigate through these movements, without the aid of push-buttons or an external operator). The architecture, summarized by Fig. 3-3 and Tables 3-1 and 3-2, incorporates a finite state structure, in which the joints assume either high or low output impedance, depending on the current finite state. Switching between finite states is largely dependent on an estimate of the location of the CoP relative to the forward heel. The approach was implemented on a powered lower limb orthosis and was assessed by having a subject with a T10 complete injury autonomously perform a series of timed-up-and-go tests. The ability of the subject to perform these tests, and the consistency of the movement between tests, indicate that the control methodology was effective in enabling the user to autonomously perform the basic movements associated with legged mobility (i.e., sitting, standing, and walking).

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CHAPTER IV

A Preliminary Assessment of Mobility and Exertion in a Lower Limb Exoskeleton for Persons with Paraplegia

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Abstract

This paper describes a functional assessment of a powered lower limb exoskeleton designed to provide gait assistance to persons with paraplegia. The authors propose an assessment protocol for assessing the mobility and exertion associated with systems that provide legged mobility assistance for persons with SCI. The mobility aspect of the assessment protocol is based on two well-established assessment tools in the clinical community, which are the timed-up-and-go test and the ten meter walk test. The exertion aspect of the assessment is based on the change in heart rate entailed in the two standard tests, in addition to the Borg rating of perceived exertion. The proposed assessment protocol was implemented on a single SCI subject, and the mobility and exertion associated with four cases of mobility and stability aids was assessed. The results indicate that the powered exoskeleton affords similar mobility and requires a somewhat lower level of exertion relative to long-leg braces with a swing-through gait, and affords significantly improved mobility with significantly less exertion relative to long-leg braces with a reciprocal gait.

Introduction

One of the most significant impairments resulting from paraplegia is the loss of mobility [1]. In addition to diminished mobility, the inability to stand and walk entails significant physiological impairments, including loss of bone mineral content, frequent skin breakdown problems, increased incidence of urinary tract infection, muscle spasticity, impaired lymphatic and vascular circulation, impaired digestive operation, and reduced respiratory and cardiovascular capacities [2].

In an effort to facilitate legged locomotion in individuals with paraplegia, several computer-controlled lower limb orthosis systems have been, and are being, developed and described in the research literature. Such orthoses include hybrid FES-systems, which supplement functional electrical stimulation (FES) of leg muscles with a computer-controlled orthosis; and fully powered orthoses (or exoskeletons), which utilize electric motors as the primary form of motive assistance. Recent examples of the former include those described in [3-7], while recent examples of the latter include those described in [8-14], in addition to the ReWalk (Argo Medical Technologies) and eLEGS (Berkeley Bionics) systems, two emerging commercial systems that have not yet been described in the engineering literature.

Despite the number of emerging systems designed to provide legged mobility assistance for individuals with paraplegia, there is currently a lack of published data by which the capabilities of each can be comparatively assessed. That is, although several such mobility assistance systems have been characterized, each system has generally used different metrics. The absence of standardized metrics is a significant impediment in uniformly and comparatively assessing the capabilities provided by a given system. This paper proposes an assessment method, and uses it to assess the mobility and level of exertion associated with the legged mobility provided by a powered lower limb exoskeleton, relative to two cases of legged mobility afforded by long-leg braces (considered as the standard current intervention for legged mobility).

A Proposed Set of Assessment Metrics

Metrics Used in Prior Publications

In [4], the authors indicate ability of the hybrid-FES system to provide legged mobility to an SCI subject by showing the variation in knee joint angle kinematics over a number of strides. In [7], the authors indicate the efficacy of the hybrid-FES system in providing legged mobility, relative to other legged mobility interventions, on four paraplegic subjects using as primary measures the average walking speed; percent increase in heart rate and blood pressure; oxygen uptake and carbon dioxide exhalation; and variation in hip and knee joint angle kinematics over a number of strides. In [8], the authors assess the efficacy of a powered orthosis, relative to a passive reciprocating gait orthosis, on four paraplegic subjects by measuring average walking speed; average step length; and the vertical and lateral motion of each subject's head. In [10], the authors quantitatively characterize the efficacy of a powered lower limb exoskeleton primarily by characterizing the walking speed with two paraplegic individuals. In this publication, the authors use heart rate, respiration rate, skin color, and perspiration levels to qualitatively assess level of exertion; the ability to maintain eye contact to qualitatively assess cognitive effort; and the ability to catch a ball to qualitatively assess standing stability. In [11] and [14], the authors evaluated the ability of a powered lower limb exoskeleton to provide sit-to-stand and stand-to-sit maneuvers to a paraplegic user by reporting the hip and knee joint angles during these maneuvers, and also by reporting the force exerted by the user's arms on a horizontal bar. In [13], the authors assess the ability of a

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powered exoskeleton to assist persons with incomplete SCI by comparing hip and knee joint angles and stride length for walking with and without the exoskeleton. Finally, in [15], the authors demonstrated the ability of a powered exoskeleton to provide walking by comparing hip and knee joint kinematics to healthy kinematics, and by reporting average walking speed.

A Standardized Set of Metrics

Although the metrics utilized in the aforementioned publications provide insight into the efficacy of each respective system, these measures in general lack uniformity and standardization, and none collectively characterize the basic functionality of a lower limb exoskeleton, which consists of standing, walking, turning, and sitting. As such, the authors propose the use of standardized walking tests that collectively characterize the degree of mobility provided by, and level of exertion associated with, a lower limb exoskeleton system for spinal cord injured persons.

A recent survey of outcome measures for persons with SCI identifies seven primary measures associated with functional ambulation [16], which include the Timed Up and Go (TUG) test, the Ten Meter Walk Test (10MWT), the Six Minute Walk Test (6MWT), the Spinal-Cord Injury Functional Ambulation Inventory (SCI-FAI), the Functional Independence Measure (FIM), the Spinal Cord Independence Measure (SCIM), and the Walking Index for Spinal Cord Injury (WISCI-II). Of these, the first three are timed measures, the latter three are categorical assessments of ambulation, and the SCI-FAI has components of both. For purposes of assessing the efficacy of mobility systems for providing legged assistance to individuals with SCI, a measurable standardized metric is preferred relative to a classification, since it largely removes subjectivity from the assessment, and further provides a means of characterizing exertion in addition to mobility. As such, for purposes of this paper, the measurable assessments (TUG, 10MWT, and 6MWT) were favored over the observational assessments (SCI-FAI, FIM, SCIM, and WISCI). With regard to the first of these measurable assessments, the TUG test measures the time required for a subject to stand from a seated position, walk three meters, turn, walk back three meters, turn, and return to the seated position. This test, which was originally proposed in [17], has been shown to have high test-retest reliability as a mobility measure across a wide spectrum of patient populations, including persons with stroke impairment, Parkinson's disease, arthritis, cerebellar disorders, and unilateral lower limb amputation [17-22]. The Ten Meter Walk Test (10MWT) measures the

time for a patient to walk ten meters, not including any acceleration or deceleration phases. Like the TUG test, the 10MWT has also been shown to have a high degree of validity and test-retest reliability in assessing the functional mobility of persons with neurological mobility impairment [23-26]. Finally, the Six Minute Walk Test (6MWT) measures the distance a person can walk in six minutes. This test is ideally performed using a straight walkway approximately 30 m long (e.g., a hallway), where the subject turns around a marker following every 30 m length. This measure was originally proposed to assess cardiovascular and respiratory capacity in persons with heart or lung diseases[26, 27], but has also been utilized as a functional mobility assessment for persons with neurologically impaired mobility [28-30].

Since the TUG test is the only one of the aforementioned timed assessments that encompasses sit-to-stand, stand-to-sit, turning, and walking; since these movements constitute the basic set of legged mobility functionality; and since the TUG test has been shown to have a test-retest reliability correlation coefficient of 0.98 among the SCI population [26], the TUG test was selected as an obvious independent measure for characterizing legged mobility systems for SCI. Further, since the TUG test is used across a wide range of impairments, it offers the added benefit of comparison of the SCI target population to a much broader patient population.

While the TUG test largely characterizes the ability of a subject to perform functional transitions (sit-to-stand, stand-to-walk, walk-to-stand, turn in place, and stand-to-sit), the 6MWT and the 10MWT largely characterize a person's (essentially steady state) walking speed. Accordingly, the two measures have been shown to have a strong correlation (i.e., a correlation coefficient of 0.95) in the SCI population [26]. Further, like the TUG test, both measures have been demonstrated to have a high test-retest correlation coefficient (both approximately 0.98) [26].

Despite the high degree of correlation between the 10MWT and the 6MWT, it has been suggested that the 6MWT can provide a measure of endurance (or exertion) that is not as well captured by the 10MWT, and it has been suggested that to "provide the most comprehensive battery [of testing] it will be important to include a measure of endurance such as the 6MWT" [31]. The authors agree with this assertion; however, the 6MWT is more difficult to administer than the 10MWT, since it requires ambulation over a 30 m walkway. Since an effective standardized assessment procedure should be administered in as universal a setting as possible, and in as short a period as possible; since both the

10MWT and 6MWT assess average walking speed, and since both demonstrate similar reliability; since the two measures are strongly correlated; and since the 10MWT is more easily administered and requires far less space than the 6MWT, the authors propose the use of the 10MWT to characterize the steady-state walking functionality of the lower limb exoskeleton systems. Note that use of the 10MWT, in lieu of the 6MWT, is further supported by [31], who suggest that the 10MWT provides "the most valid measure of improvement in gait and ambulation" for the SCI population, and [30], who assert that "the 10MWT appears to be the best tool to assess walking capacity in SCI subjects." As such, the TUG and 10MWT assessments are utilized herein to characterize the efficacy of a lower limb exoskeleton system in providing legged mobility to a person with paraplegia, and the basic transitions associated with it. The authors acknowledge, however, that supplementing these outcome measures with the 6MWT would provide a more complete assessment of the efficacy of this and other lower limb gait assistance systems.

A gait assistance system should provide effective mobility without an undue level of exertion. The single measure typically utilized in both the TUG and 10MWT is the time required to complete each respective test. This measure provides a quantitative measure of mobility (the former largely characterizing the efficacy of gait transitions, the latter characterizing primarily the efficacy of steady-state walking), but neither characterizes the level of exertion associated with these activities. In order to characterize the level of exertion, the time to complete these tests is supplemented with measurement of the pre- and post-test heart rate, which is commonly used to measure physical exertion. This physiological measure of exertion is further supplemented by rating each test with the Borg Rating of Perceived Exertion (RPE) Scale, as proposed in [32], which is well-validated and widely used indication of perceptual effort, in which a subject rates his or her level of exertion on a scale of 6 to 20, where 6 corresponds to "no exertion at all" and 20 corresponds to "maximal exertion." On this scale, casual healthy walking corresponds to a rating of 9.

Thus, the proposed assessment of a powered legged assistance system for persons with SCI consists of the combination of a TUG test and a 10MWT, where the mobility provided by the system is characterized by the time required to complete each test, and the exertion required by the system is characterized by the change in pre- and post-test heart rate, and by the Borg Rating of Perceived Exertion associated with each test.

<u>Comparative Assessment of Two Mobility Aids</u>

The previously discussed assessment procedure was utilized to comparatively assess the efficacy of a powered lower limb exoskeleton system in providing legged mobility to a paraplegic subject with a T10 motor and sensory complete injury (i.e., American Spinal Injury Association, ASIA, A classification). Specifically, this paper presents the aforementioned mobility and exertion measures for the TUG and 10MW tests for four mobility aid combinations, which include 1) the Vanderbilt powered lower limb exoskeleton with a walker; 2) the Vanderbilt powered lower limb exoskeleton with forearm crutches; 3) a set of long-leg braces with a walker, using a swing-through type gait; and 4) a set of long-leg braces with a walker, using a reciprocal type gait. A video is included with the supplemental material that shows the subject walking with each of the four methods.

Vanderbilt Lower Limb Exoskeleton

The Vanderbilt lower limb exoskeleton, shown in Fig. 4-1, provides powered assistance in the sagittal plane at both hip and knee joints. The exoskeleton consists of a hip segment, a right and left thigh segment, and a right and left shank segment. The hip segment contains a lithium polymer battery which powers the exoskeleton, of which each thigh segment contains a pair of brushless DC motors, which actuate the hip and knee joints respectively through speed reduction transmissions. The knee joints are additionally equipped with normally-locked brakes, in order to preclude knee buckling in the event of a power failure. Although the exoskeleton does not explicitly contain a foot segment or ankle joint, it is designed to be used in conjunction with a standard ankle foot orthosis (AFO), which provides stability at the ankle, and precludes foot drop during the swing phase of gait. The total mass of the exoskeleton, including the battery, is 12 kg (26.4 lb). A more detailed description of the exoskeleton design, including a description of the embedded electronics system, is given in [15].

As required by the assessment procedures previously described, the powered lower limb exoskeleton enables sit-to-stand transitions, standing, stand-to-walk transitions, walking, walk-to-stand transitions, and stand-to-sit transitions. In order to enable the user to have autonomous control of these maneuvers, a user interface approach was developed based on the user's ability to affect his or her

center of pressure via the use of his or her upper body, in combination with a stability aid. Specifically, based on sensors embedded in the exoskeleton, the control system estimates the location of the user's center of pressure (CoP), defined as the user's center of mass projection onto the (assumed horizontal) ground plane, and uses the distance between the CoP and the location of the forward ankle joint as the primary command input. Thus, the user transitions out of a given activity (sitting, standing, or walking) by leaning forward or back, such that the CoP moves in an anterior or posterior direction, which commands the controller to transition to a different activity mode. This approach enables the user to autonomously perform the TUG test and 10MWT presented here, without the assistance of an external operator. A more detailed description of the exoskeleton control architecture and user interface, which discusses more specifically the conditions required to move between activities, is given in [33]. Finally, note that the turning maneuver (performed twice in each TUG test) does not entail a separate control mode, but rather is performed in the standing activity mode, with the use of the stability aid, by incrementally twisting the upper body and turning in place. This follows the typical turning methodology utilized with long-leg braces (see discussion below).

Figure 4-1. Vanderbilt lower limb exoskeleton.

Long-leg Braces

The mobility and level of exertion associated with the powered exoskeleton were compared to the respective mobility and level of exertion associated with long-leg brace ambulation. Long-leg braces are the most common legged mobility aids used by persons with paraplegia. The braces used in this case study, which are representative of this type of mobility aid, are shown in Fig. 4-2. These braces consist of a thigh segment, shank segment, and integrated shoe for each leg. The knee joint of each leg consists of a latching hinge joint, such that the joint can remain flexed while donning or sitting, but mechanically locks at full extension, and remains locked during use. Following use, the user can unlatch the knee joints with

the posterior lever, which facilitates a more natural seated posture, and simplifies the doffing procedure. Most long-leg braces incorporate posterior bail locks, which release the knees as the metal bail (located behind and slightly above the knee) is forced upward by the edge of a seat as the user leans backwards. In addition to locking knee joints, each leg of the long-leg braces incorporates an articulated ankle joint, which allows limited ankle dorsiflexion, but precludes ankle plantarflexion.

Long leg braces such as those shown in Fig. 4-2 require the use of a stability aid, and are typically used with either a swing-through or reciprocal method of ambulation. The reciprocal method is an approximation of healthy gait, in which the subject alternatively takes left and right steps. Specifically, the subject uses the stability aid to alternately lean left and right, which unweights the swing leg, while simultaneously leaning forward, such that gravity can act to swing the leg forward. This method tends to be slower and more laborious than the swing-through method, in which the subject moves the stability aid forward, resulting in a forward leaning posture, and unweights both legs simultaneously, such that gravity acts to swing both legs through together. This method does not require leaning side-to-side, and since one leg need not remain on the ground while the other is in swing, it generally affords greater ground clearance and larger strides. Note that the swing-through method is typically used with a "spreader bar," as shown in Fig. 4-2, which mechanically couples both legs together near the ankle joints.

A typical sit-to-stand maneuver with braces starts with the subject seated in a chair, with the knee joints fully extended and locked, with the legs fully extended in front of the user, and the heels of the shoes in contact with the ground. Using a walker as a stability aid, as was the case with the assessments reported here, the user pushes upward in the walker, such that his or her legs are drawn up through the walker, until in a fully upright position. A typical stand-to-sit maneuver requires that the user position himself or herself in front of a chair, then bending forward at the waist, and essentially falling backward into the chair as the chair lifts the bail locks, unlocking the knees. Finally, turning in place is afforded by incrementally rotating the stability aid by twisting the upper body, then unweighting the legs to reorient them with the upper body and stability aid. Using this method, a user can typically turn 180 deg in three or four increments with a walker, and approximately twice as many increments with forearm crutches.

Figure 4-2. Long-leg braces used in assessments (shown with spreader bar attached for swing-through gait).

Assessment Methodology

The mobility and level of exertion associated with the powered lower limb exoskeleton was assessed with the previously described metrics, and compared to the respective mobility and level of exertion associated with long leg brace ambulation. These assessments were performed on a single paraplegic subject with a T10 motor and sensory complete injury (ASIA A classification). The subject was 35 year of age, 9 years post-injury, 1.85 m (6 ft) tall, and with a body mass at the time of testing of 73 k (160 lb). The subject is shown wearing the lower limb exoskeleton and long leg braces in Figs. 4-3 and 4-4, respectively.

Figure 4-3. T10 complete paraplegic subject wearing Vanderbilt exoskeleton.

Figure 4-4. Subject wearing long-leg braces.

TUG Test Protocol

For the TUG test, the (lightly-colored) floor was marked with two lengths of dark tape placed 3 m (10 ft) apart, which designated the starting position and turning position, respectively. A wheelchair with locked wheels and footrests removed was used as the chair for the TUG test, and was positioned fully behind the starting position. The subject was instructed to wait for the verbal queue to start, then stand, walk until his body crossed the turning mark, turn, walk back to the chair, turn, and sit. The total time was recorded from the initial verbal queue, to the time the subject returned to a seated position in the wheelchair.

In order to standardize exertion measurement, the subject's heart rate was taken exactly one minute prior to the start of each TUG test, and exactly 30 sec following each TUG test. The heart rate measurement was taken with an automated monitor (Dynamap V100 by General Electric), which required approximately 20 sec from initiation (i.e., donning of finger clip) to measurement. Specifically, the TUG test was initiated 60 sec after reporting the subject's heart rate, while heart rate measurement was initiated 30 sec after TUG test completion (and as such the heart rate measurement following each TUG test was reported approximately 50 sec after completion of the test). The subject was allowed to practice the TUG test until he felt comfortable performing it. Once the subject was comfortable performing the TUG test, the test and associated heart rate and perceived exertion measurements were performed three consecutive times. The subject was allowed a period of rest between each test, until he felt rested and ready to perform the next test.

10MWT Protocol

For the 10MWT, the floor was marked with two lengths of dark tape placed 10 m (33 ft) apart. In this test, the subject ambulated at a steady-state through the 10 m walkway (i.e., the subject starting walking several meters prior to the first mark, and continued to walk through the second mark. The starting and ending times were recorded based on the subject's body crossing the respective marks. As with the TUG test, the subject's heart rate was taken approximately 60 sec prior to and 60 sec following each test. Since the 10MWT does not start and end in a seated position, however, and since these measurements were taken in a seated position, the intervals between heart rate measurement and the

start and end of the 10MWT were somewhat more approximate than during the TUG test. As with the TUG test, the subject completed the 10MWT three consecutive times, and was allowed to rest between tests until he felt ready to perform the next.

Mobility Aid and Stability Aid Combinations

Four cases of ambulation were assessed. The subject was able to independently perform both the TUG test and 10MWT with both a walker and with forearm crutches, and therefore both cases were assessed. When using the long-leg braces, the subject was able to independently perform these tests using a walker as a stability aid, but was not able to perform these tests with forearm crutches. As such, ambulation with the long-leg braces was conducted with a walker as stability aid, for the cases of reciprocal and swing-through ambulation.

For both cases using the exoskeleton, the TUG test started and ended in a seated position with both knees flexed. For both cases using the long-leg braces, the TUG test started in a seated position with both knees fully extended and locked (i.e., the time required to extend and lock the knees prior to standing, was not included in the TUG test times). Although no assistance was provided to the subject during any of the tests, all tests involved the use of a gait belt and close monitoring by a trained physical therapist, as per the Institutional Review Board approval corresponding to these assessments.

Borg Rating of Perceived Exertion

Rather than rate each trial individually, the subject was asked to rate the collective battery of tests (i.e., three trials of a TUG test, and three trials of a 10MWT) for each combination of mobility and stability aid, using the Borg scale of perceived exertion. As such, each mobility and stability aid combination was assessed by the subject with a single Borg score.

Results and Discussion

The results of the assessment for each the four cases of ambulation are summarized in Table 4-1. Additionally, a video is provided with the supplemental material that provides a qualitative understanding

of the legged mobility provided in each case. For each of the four cases of ambulation, the table lists the average TUG test time (in seconds) and corresponding standard deviation (in parentheses) for the three TUG trials; the average 10MWT time and corresponding standard deviation for the three 10MWT trials; the average change in pre- and post-test heart rate, and associated standard deviation across the three trials for each test type; and the subject's rating of perceived exertion corresponding to the battery of assessments for each ambulation case. A single-degree-of-freedom analysis of variance (ANOVA) was conducted to assess the extent to which the mean measurement of each test type was significantly different from each of the other test types. The mobility and exertion measures, and the extent to which they are different, are discussed in the sections below. Note that the ANOVA was conducted using a confidence level of 90%, unless otherwise noted in the discussion.

Walking Method	TUG Time (seconds)	Heart Rate Change (%)	10MWT Time (seconds)	Heart Rate Change (%)	BORG Perceived Exertion
LL Braces + Walker (Reciprocal)	178 (14)	41.8 (17.1)	109 (7)	18.4 (5.9)	16
LL Braces + Walker (Swing-Through)	118 (3)	19.0 (7.2)	89 (17)	16.1 (2.9)	13
Exoskeleton + Walker	107 (5)	10.1 (4.6)	81 (10)	5.4 (9.5)	12
Exoskeleton + Forearm Crutches	120 (4)	3.9 (5.4)	89 (4)	-1.2 (10.7)	10

*Results are average and (standard deviation) of three trials.

Table 4-1. Summary of Assessment Data.

Mobility

The principal measures of mobility consist of the average TUG test time and the average 10MWT time. The average TUG times for each mobility aid case, bracketed by plus and minus one standard deviation, are shown graphically in Fig. 4-5. As reported in Table 4-1, the exoskeleton with walker provided the fastest average TUG test time (107 sec), followed by the long-leg braces with a swing-through gait (118 sec), the exoskeleton with forearm crutches (120 sec), and lastly by the long-leg braces

with reciprocal gait (178 sec). Based on the ANOVA, however, the difference in means between the exoskeleton and swing-through cases is not significant, while the difference between these means and the mean of the reciprocal gait with braces is significant by a substantial margin. The 10MWT times showed a similar trend, as shown graphically in Fig. 4-6. Specifically, the exoskeleton with walker provided the fastest average 10MWT time (81 sec), followed by the long-leg braces with a swing-through gait and exoskeleton with forearm crutches (each of which had an average time of 89 sec), and lastly by the long-leg braces with reciprocal gait (109 sec). As with the TUG times, based on the ANOVA, the difference in means between the former three cases is not significant, while the difference in means between the slowest of the first set (i.e., long-leg braces with swing-through gait) and the long-leg braces with reciprocal gait is significantly different (with a confidence level of 83%). Therefore, the mobility component of the assessment indicates that the exoskeleton (with either walker or forearm crutches) provides similar mobility to a swing-through gait with braces and a walker, while all of these provide significantly improved mobility relative to the reciprocal gait afforded by the braces with a walker.
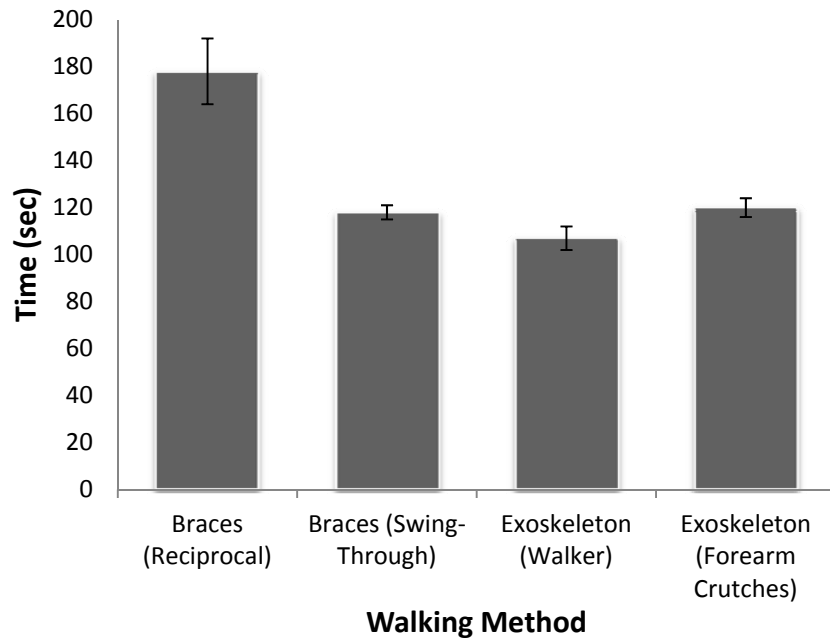
Figure 4-5. Graph of TUG test completion times in each walking method.
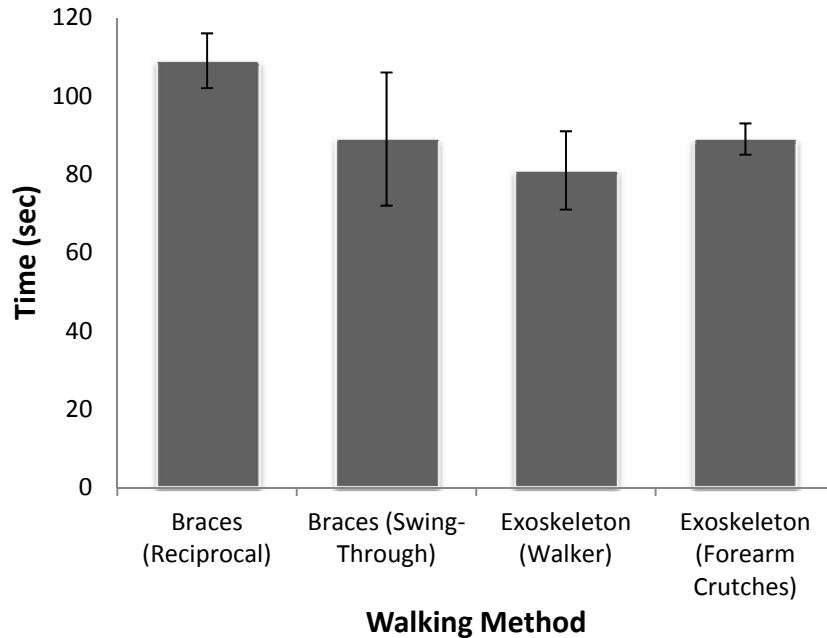
Figure 4-6. Graph of TMWT completion times in each walking method.

Exertion

The principal measures of exertion are the percent change in heart rate during the TUG and 10MW tests, in addition to the Borg perceived exertion ratings. The percent change in heart rate is defined as the difference between the post and pre-test heart rates, divided by the pre-test heart rate (multiplied by 100). The average change in heart rate for both TUG and 10MWT is summarized graphically in Figs. 4-7 and 4-8. For both the TUG and 10MWT, the average change in heart rate was the smallest for the case of walking with the exoskeleton with forearm crutches (3.9% and -1.2%, respectively, were the latter indicates that on average, the heart rate decreased during the 10MWT). The other cases of walking were similarly ordered for both the TUG and 10MWT. In order of increasing average heart rate for the TUG and 10MWT, the remaining cases were ordered as the exoskeleton with

walker (10.1% and 5.4%, respectively), the swing-through gait with the braces and walker (19.0% and 16.1%, respectively), and the reciprocal gait with the braces and walker (41.8% and 18.4%, respectively). Despite the fact that the average change in heart rate was much lower for the exoskeleton with crutches and walker, respectively, relative to the braces with swing-through gait, the ANOVA indicates that the difference in means is not always statistically significant. Specifically, for the TUG test, the difference in means for the three lower average changes in heart rate is not statistically significant; these are, however, all statistically significantly lower than the case of reciprocal walking with braces. For the 10MWT, the difference in means of the two exoskeleton cases are not statistically different; the difference in means of the two cases with braces are also not statistically different; while the difference in means between these two groups are significantly different (but with the variance in data, only with a 61% confidence level). As such, for the TUG test, the exoskeleton (with both walker and crutches) and the swing-through gait with braces and walker required statistically similar levels of exertion, while the reciprocal gait with the braces and walker required a significantly greater level of exertion. With the 10MWT, both cases with the exoskeleton required similar levels of exertion, while both cases with the braces required significantly increased exertion.

The aforementioned measurements are reinforced by the Borg perceived exertion ratings. As previously mentioned, the subject provided a single, collective Borg perceived exertion rating for each mobility case, as graphically depicted in Fig. 4-9. The subject's ratings of perceived exertion correlates strongly with the order of mean changes in heart rate indicated in Figs. 4-7 and 4-8. That is, in order of increasing effort, the subject rated the exoskeleton with crutches as the least taxing (RPE of 10, which is slightly more than equivalent to healthy casual walking); followed by the exoskeleton with walker (RPE of 12); braces and walker with a swing-through gait (RPE of 13); and braces and walker with a reciprocal gait (RPE of 16). Thus, legged mobility with the exoskeleton was generally considered as "light exercise" while legged mobility with the braces was considered as "hard (heavy) exercise," according to the Borg scale [32].
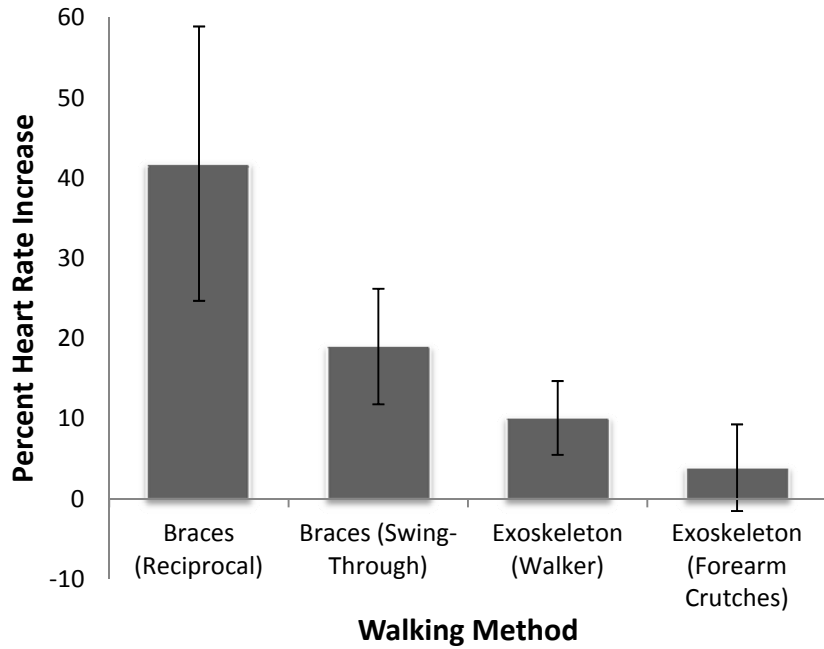
Figure 4-7. Graph of user exertion during TUG test in each walking method.
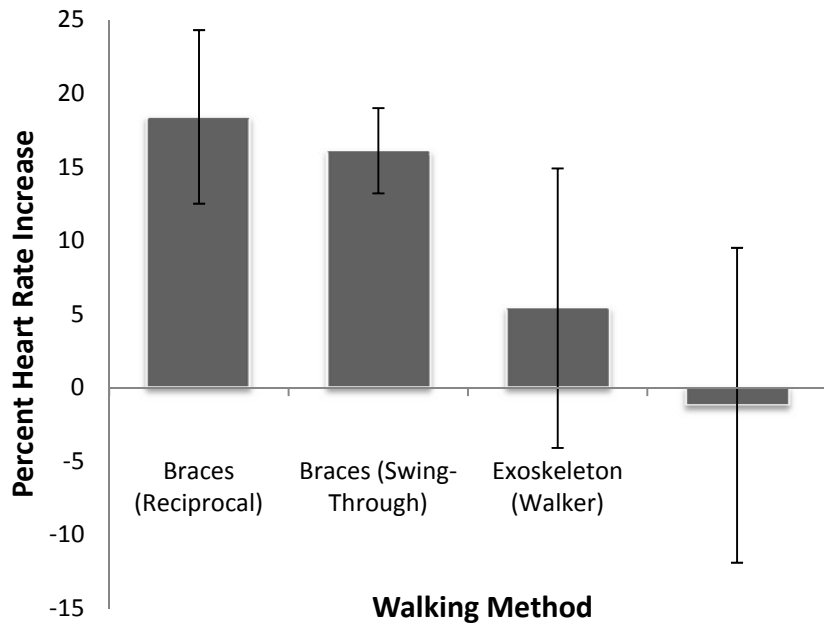
Figure 4-8. Graph of exertion during TMWT in each walking method.
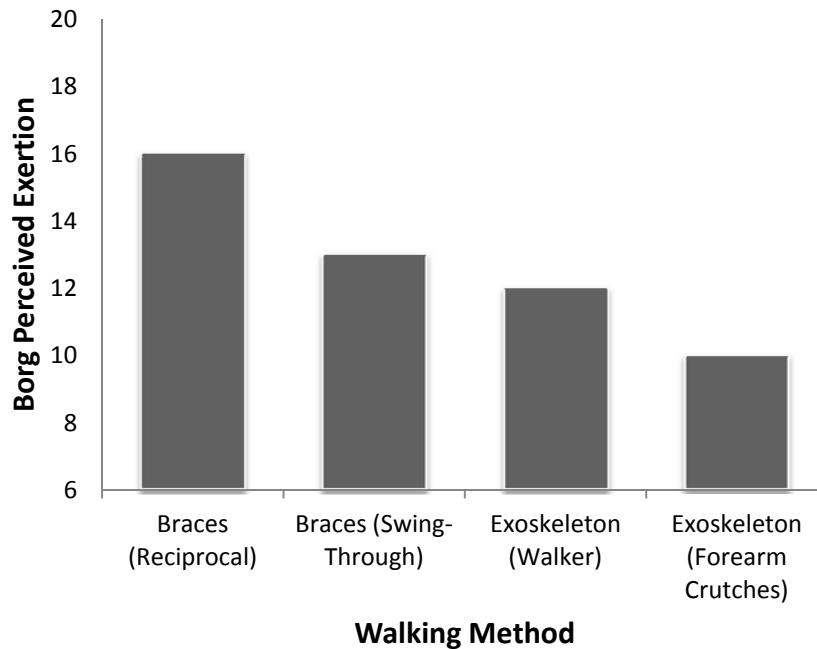
Figure 4-9. Graph of user-perceived exertion during standardized walking tests in each walking method. Note that the Borg Scale ranges from 6 ("no exertion") to 20 ("maximal exertion").

Consideration of Walking as a Fundamentally Reciprocal Activity

For purposes of completeness, the authors included the case of swing-through gait in the assessment and discussion presented here. One can legitimately argue, however, that "legged mobility" as applied to human subjects implies walking; walking fundamentally entails a reciprocal movement of the legs; and therefore, a swing-through gait should not be included in assessing the efficacy of a mobility aids for facilitating legged mobility in persons with SCI. If one therefore considers only the three cases of mobility aids that provide a reciprocal gait (i.e., the two exoskeleton cases and the reciprocal gait with braces and walker), then in all cases, the exoskeleton (with either stability aid) provides significantly faster TUG and 10MWT times, with significantly lower levels of exertion, relative to the braces. Specifically, the exoskeleton provides on average 37% faster TUG times and 22% faster 10MWT times, and

concomitantly entails an 83% lower change in heart rate for the TUG test, and an 89% lower change in heart rate for the 10MWT (note here that the two exoskeleton cases were averaged together, since as previously discussed, these cases are statistically similar). As such, based on the previously discussed measures of mobility and exertion in legged mobility aids, the exoskeleton provides significantly improved mobility, and concomitantly provides significantly less exertion, relative to the reciprocal walking enabled by long-leg braces and a walker.

Conclusion

Systems have started to emerge that provide legged mobility for persons with paraplegia. In this paper, the authors propose an assessment methodology to assess the efficacy of such systems, specifically with regard to mobility and exertion, based on the TUG test and 10MWT (two well-established assessments in the clinical community). Using this assessment methodology, the authors assessed four cases of legged mobility on a single paraplegic subject. Specifically, legged mobility was assessed with long-leg braces and a walker, with both a reciprocal and swing-through gait, and was also assessed with a powered lower limb exoskeleton, with both a walker and forearm crutches as a stability aid. The assessments suggest that the degree of mobility provided by the exoskeleton was statistically similar to the degree of mobility afforded by long-leg braces with a swing-through gait, while the level of exertion was similar to the swing-through gait for the TUG test, and significantly lower than the swing-through gait for the 10MWT. The assessments further suggest that the degree of mobility provided by the exoskeleton was significantly better than the degree of mobility provided by long-leg braces with a reciprocal gait, while the level of exertion required by the former was significantly lower than that required by the latter.

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CHAPTER V

A cooperative controller to integrate FES in a lower limb powered exoskeleton for persons with paraplegia

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Abstract

This paper describes a cooperative controller that combines the torque and power contribution of an externally powered lower limb exoskeleton, with the torque and power contribution of electrically stimulated muscle, in order to combine the benefits of each. The cooperative control structure is described in the paper, and the control approach is implemented on a T10 complete paraplegic individual and a prototype lower limb exoskeleton. Specifically, the control approach is implemented on the knee motor of the exoskeleton, in conjunction with electrical stimulation of the quadriceps muscle groups. The efficacy of the control approach is validated in three sets of experiments, consisting of weighted leg lifts, sit-to-stand maneuvers, and stair ascent maneuvers, respectively. Measurements from these experiments indicate that the control approach effectively combines the respective efforts of the motor and muscle, such that good control performance is achieved, with substantial torque and energy contributions from both the biological and non-biological actuators.

Introduction

There are approximately 12,000 new cases of paraplegia in the US each year [1]. Since the US constitutes approximately 4% of the world's population, the worldwide annual incidence of paraplegia is perhaps 25 times greater (i.e., 300,000 new cases per year). Although paraplegia entails a number of disabilities, surveys of this population indicate that the loss of legged mobility is among the most significant [1-4].

A number of interventions have been developed to facilitate legged mobility in paraplagics. Passive approaches for legged mobility assistance include leg braces, knee-ankle-foot-orthoses (KAFOs), and reciprocating-gait-orthoses (RGOs). In an effort improve upon these approaches, researchers have introduced sources of power to facilitate legged locomotion. Such sources of power can generally be categorized as either metabolic or external. Specifically, a number of researchers have investigated the use of functional electrical stimulation (FES) to artificially control leg muscle contraction, thereby utilizing the metabolic power supply of the paraplegic subject to facilitate legged locomotion. Some of the research

efforts in this regard include [5-7]. Note that, in addition to providing power for locomotion, FES-aided gait has also been shown to provide a number of associated physiological benefits, some of which include increased muscle strength, increased bone density, decreased spasticity, and improved cardiovascular health [8-10]. Despite this, FES-based systems entail a number of challenges; most notably, muscles are difficult to control, particularly in the absence of adequate sensory information, and the synchronous recruitment of muscle fibers used in FES promotes rapid muscle fatigue.

Rather than utilize the metabolic power source provided by stimulated muscle, other researchers have developed lower limb exoskeletons, which utilize external power (e.g., a battery pack) to provide a source of motive power to facilitate legged locomotion. Some of the efforts in this regard include [11-19]. This approach avoids the controllability and fatigue issues associated with FES, but such approaches do not offer several of the physiological benefits associated with FES, and also forego the convenience of leveraging the metabolic power supply.

In this paper, the authors combine the use of an externally powered lower limb exoskeleton with FES, such that the resulting system maintains a high degree of controllability and circumvents problems associated with muscle fatigue, while also leveraging the availability of the metabolic power supply, and providing the physiological benefits associated with FES. The proposed approach is somewhat similar to a hybrid FES approach, as developed and described by others, see for example [20-24], but unlike any previously described hybrid approach, this paper combines FES with a fully externally powered lower limb exoskeleton. The authors describe the control approach utilized to coordinate the cooperative control between the stimulated muscle and the exoskeleton motors. In order to validate the efficacy of the approach, the cooperative controller was implemented in a set of three experiments in which the quadriceps muscle group of a paraplegic individual was controlled in conjunction with an existing lower limb exoskeleton as described in [17-19]. Specifically, the cooperative controller coordinated the electrical stimulation of the right and left quadriceps muscle groups in conjunction with the respective knee motor of the exoskeleton, in order to provide coordinated control of the respective knee joint. The efficacy of the cooperative controller in combining muscle and motor efforts is demonstrated under three experimental conditions, as described in the results section of the paper.

<u>Lower Limb Exoskeleton Prototype</u>

The Vanderbilt exoskeleton [17, 19] is a fully powered lower limb orthosis, designed to allow a paraplegic user to stand and walk. The 26.5 pounds device has five links and four articulations as shown in Figure 5-. The device is electrically actuated and it is powered by a 29.6V lithium polymer battery that is contained in the hip piece of the unit. All the actuation and power transmission is in the thigh links as well as the electronics required for control. The exoskeleton is designed to be used with standard forearm crutches (or a walker) and a standard ankle foot orthoses (AFO).
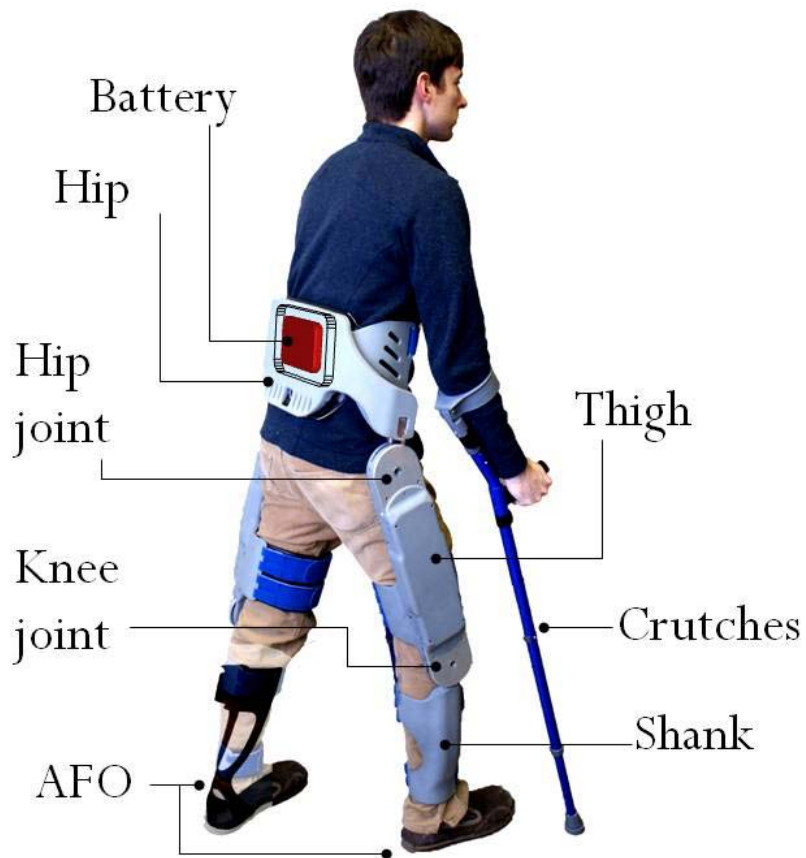
Figure 5-1. Vanderbilt lower limb exoskeleton.

Cooperative Controller Design

The essence of the control design is to use as much muscle effort as possible, and then "fill in" the remaining required joint torque with the externally powered joint actuator. As described in prior publications, in the absence of FES, the (externally powered) joint-level exoskeleton controllers, incorporate a proportional derivative (PD) control structure (see [19]). Assuming a constant stimulation frequency sufficient to provide a fused contraction, the nominal amplitude of muscle effort (i.e., strength of contraction) can be modulated in an FES controller by controlling the (current) amplitude and/or pulse width of the stimulation waveform. Since the effect of modulating these two variables is essentially the same, the stimulation pulse width was held constant in this work, such that the muscle effort was modulated via a modulation of the stimulation amplitude. In controlling muscle stimulation, one would ideally have a model of the muscle, describing the dynamic relationship between stimulation input and muscle torque output. Since this model is generally not available, and since it is well known that the model is characterized by a highly time-varying component, the control approach used here assumes a generic form for this model, and incorporates an adaptive control approach to effectively estimate the gain and phase lag introduced by the muscle stimulation dynamics. As such, the cooperative controller incorporates a gradient-based adaptive component that estimates the gain and phase characteristics of the muscle, and uses these estimates to adjust the relative effort and timing of the muscle stimulation, relative to the torque commands issued to the electric motor. A schematic of the cooperative controller is shown in Figure 5-2.
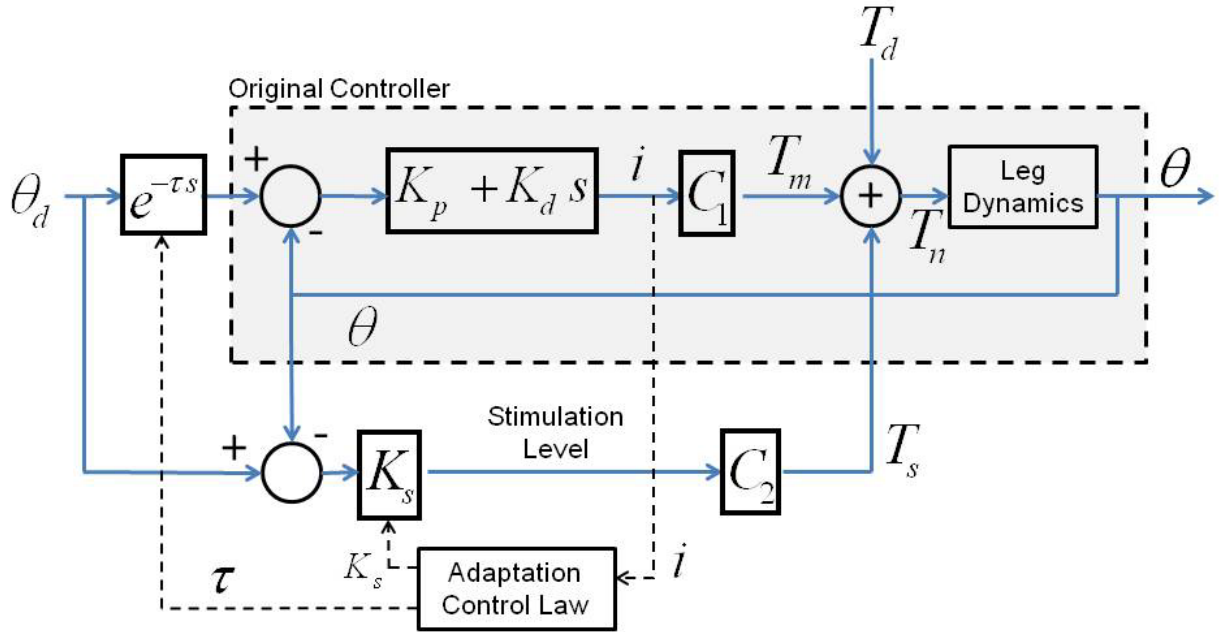
Figure 5-2. Block diagram of the motor controller supplemented with cooperative, adaptive control of FES.

Since, as previously mentioned, the control objective is to use as much muscle torque as possible, we can alternatively characterize the objective as minimizing the magnitude of actuator torque effort. Thus, the controller is derived by first establishing the cost functional:

$$J = \frac{T_m{}^2}{2} \qquad \qquad \text{Eq1}$$

Where Tm is the joint actuator torque. We characterize the gain and phase characteristics of the muscle dynamics by Ks and τ, respectively, and assuming the cost functional is a differentiable function of these characteristics, we can characterize the gradient of the cost functional as:

$$\begin{bmatrix} \frac{dK_s}{dt} \\ \frac{d\tau}{dt} \end{bmatrix} = -\gamma \frac{\partial J}{\partial (K_s, \tau)} = -\gamma T_m \frac{\partial T_m}{\partial (K_s, \tau)} \qquad \qquad \text{Eq2}$$

Referring to the system structure shown in Figure 5-2, we can write:

$$T_m = T_n - T_s - T_d$$

$$T_m = (T_n - T_d) - (\theta_d e^{\tau s} - \theta)K_s c_2$$

Where T_m is the actuator torque, T_s the torque provided by the muscle, T_n the net torque at the knee joint, and T_d the joint torque disturbance caused by any external loading. On Figure 5-2, c_1 and c_2 are conversion factors from current to torque and stimulation level to torque respectively.

Taking the partial derivatives of T_m respect the parameters (K_s, τ) and replacing in equation 2 we obtain the appropriate rate of change of K_s and τ as a function of measurable variables:

$$\frac{\partial T_m}{\partial K_s} = -c_2(\theta_d e^{\tau s} - \theta)$$

$$\frac{\partial T_m}{\partial \tau} = -c_2 K_s(\dot{\theta}_d)$$

$$\begin{bmatrix} \frac{dK_s}{dt} \\ \frac{d\tau}{dt} \end{bmatrix} = \gamma c_2 T_m \begin{bmatrix} (\theta_d e^{\tau s} - \theta) \\ K_s(\dot{\theta}_d) \end{bmatrix} = T_m \gamma_r \begin{bmatrix} (\theta_d e^{\tau s} - \theta) \\ K_s(\dot{\theta}_d) \end{bmatrix} \qquad \text{Eq.3}$$

Where

$$\gamma_r = \gamma c_2$$

γ_r is an adaptation control gain that modulates the rate of adaptation.

Experimental Implementation

The previously described control approach was implemented on a paraplegic subject wearing the previously described lower limb exoskeleton. These experiments were conducted using electrical stimulation of the quadriceps muscle, and therefore only the knee joint utilized cooperative control, and only when requiring a torque in knee extension (i.e., the quadriceps cannot provide a knee flexion torque). The subject had a T10 (ASIA A) complete injury, is a 36 year-old male, 9 years post injury, is 1.9m (6 ft 1

in) tall, and has a mass of 73 Kg (160 lb). All experiments were conducted at the Shepherd Center in Atlanta GA USA, a hospital specializing in spinal cord injury. All experiments required the presence of a physical therapist, as per the experimental protocol approved by the Vanderbilt Institutional Review Board.

Electrical stimulation

Electrical stimulation was provided by a custom module that plugs into the main exoskeleton control board. Both the stimulator module and the main board are shown in Figure 5-3. As implemented, the same microcontroller that controls the joint actuator also controls the electrical stimulator. The stimulator provides a biphasic square-wave stimulation waveform. In the experiments described here, the stimulation frequency was set to a constant 50 Hz, and the pulse width set to 200 μs. As previously described, the stimulation amplitude was set by the structure of the adaptive controller. Note that, for safety purposes the stimulation amplitude was saturated at 120 mA. A pair of surface electrodes was placed on each of the right and left quadriceps. The stimulation leads were run proximally, through the waistband of the subject's pants, where they connected with mating leads from the previously described stimulator.
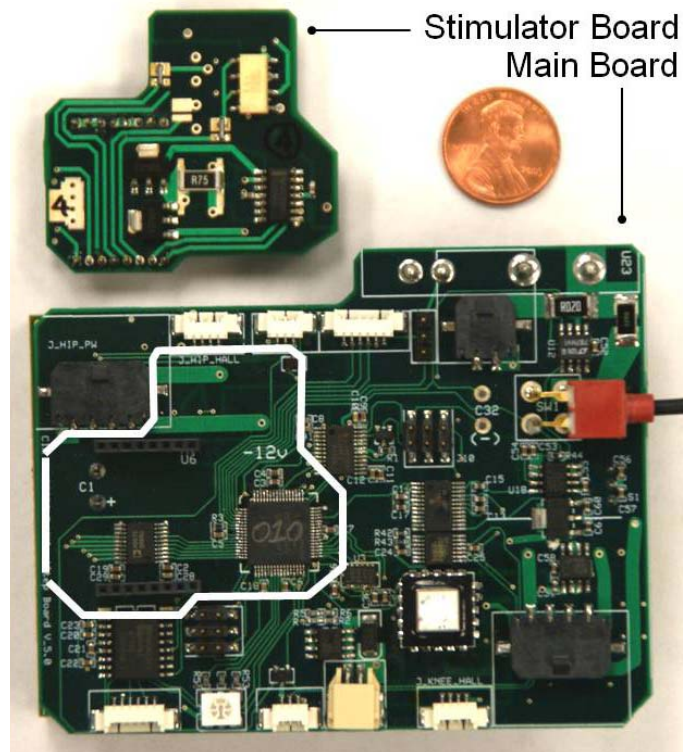
Figure 5-3. Exoskeleton embedded system with plug-in FES module.

Experimental setups

Three experimental setups were used to evaluate the cooperative controller. A series of trajectory-controlled leg lifts were used to assess the efficacy of the cooperative controller to adapt to appropriate gain and phase parameters, and to provide cooperative control under light loading conditions. A series of sit-to-stand maneuvers, and a series of stair ascent maneuvers, were used to assess the efficacy of the cooperative controller during mobility maneuvers, both of which require significant extensive torque at the knee joint. In all of the experiments described below, the adaptation gain γ_r was set to 1×10^{-3}, and the control loop was implemented at a sampling rate of 1 kHz. In the leg lift experiments, the PD control gains were set to kpc1 = 62 Nm/deg and kdc1 = 1.87 Nms/deg, which are appropriate gains for trajectory tracking. In the sit-to-stand and stair ascent maneuvers, the cooperative controller was used in the format of the sit-to-stand controller reported in [19]. In this controller, rather than use fixed PD control gains and a

varying joint angle trajectory, the joint controller uses a step input in the joint angle trajectory, and ramps the PD control gains from zero to the same values used in the leg lift experiments over a period of 1.75 s. The following subsections describe each of the experimental conditions.

Leg lifts while sitting

The first set of experiments involved the subject performing a series of seated leg lifts while wearing a 2.75 kg (6 lb) ankle weight. Specifically, the knee joint was commanded to move in a sinusoidal trajectory from a flexion angle of 50 deg, to a flexion angle of 15 deg (where 0 deg would be a fully extended knee). The sinusoidal trajectory had a 1 Hz period, with a 5 s pause between each successive leg lift. In all experiments, the cooperative controller with FES assistance was alternated with control by the exoskeleton only.

Sit to stand maneuver

In this set of experiments, the subject performed a series of sit-to-stand maneuvers, with the aid of a walker. The control methodology used for this maneuver is described in [19]. Since the knee torque required for this maneuver was always greater than the maximum torque generated by FES of the quadriceps (at the maximum stimulation amplitude), the gain and phase estimates consistently saturated at their respective allowable limits.

Stair ascent.

These experiments cooperatively utilized the quadriceps muscle groups and the knee joint motors to provide knee joint torques for stair ascent. Like the sit-to-stand maneuver, stair ascent required more extensive knee torque than could be provided by the FES alone, and therefore the adaptation consistently saturated at the maximum allowable limits.

Leg lifts while sitting

Figure 5-4. Experimental setup for cooperative control of weighted leg lifts.

Figure 5-4 shows the paraplegic subject performing the weighted leg lift experiments, while Figures. 5-5 and 5-6 show corresponding data from a series of leg lifts. Specifically, the top row in Figure 5-5, shows ten seconds of the knee angle tracking during cooperative control at the beginning and the end of a ten-minute leg lift experiment, while the bottom row shows the moving estimate of both the gain and phase characteristics of the muscle model. The convergence in parameter estimates, in addition to the effective tracking and (subsequently described) torque sharing, indicates the efficacy of the adaptation scheme. Figure 5-6 shows the overlaid knee joint trajectories and torques during the series of leg lifts. Specifically, the top row in Figure 5-6 shows the overlaid trajectories from 42 leg lifts without FES, along with the exoskeleton motor torque required to achieve these trajectories.
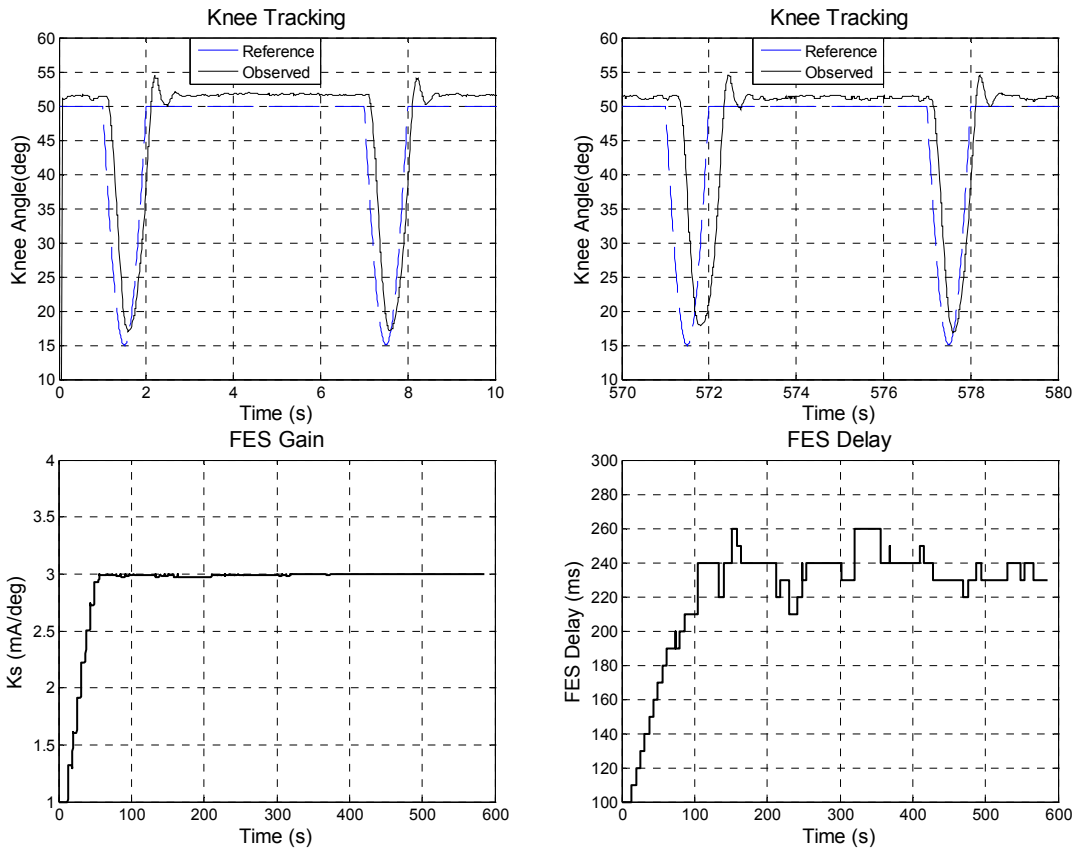
Figure 5-5. Knee joint tracking (top) and gain and phase parameter adaptation (bottom) during a series of weighted leg lifts with FES.
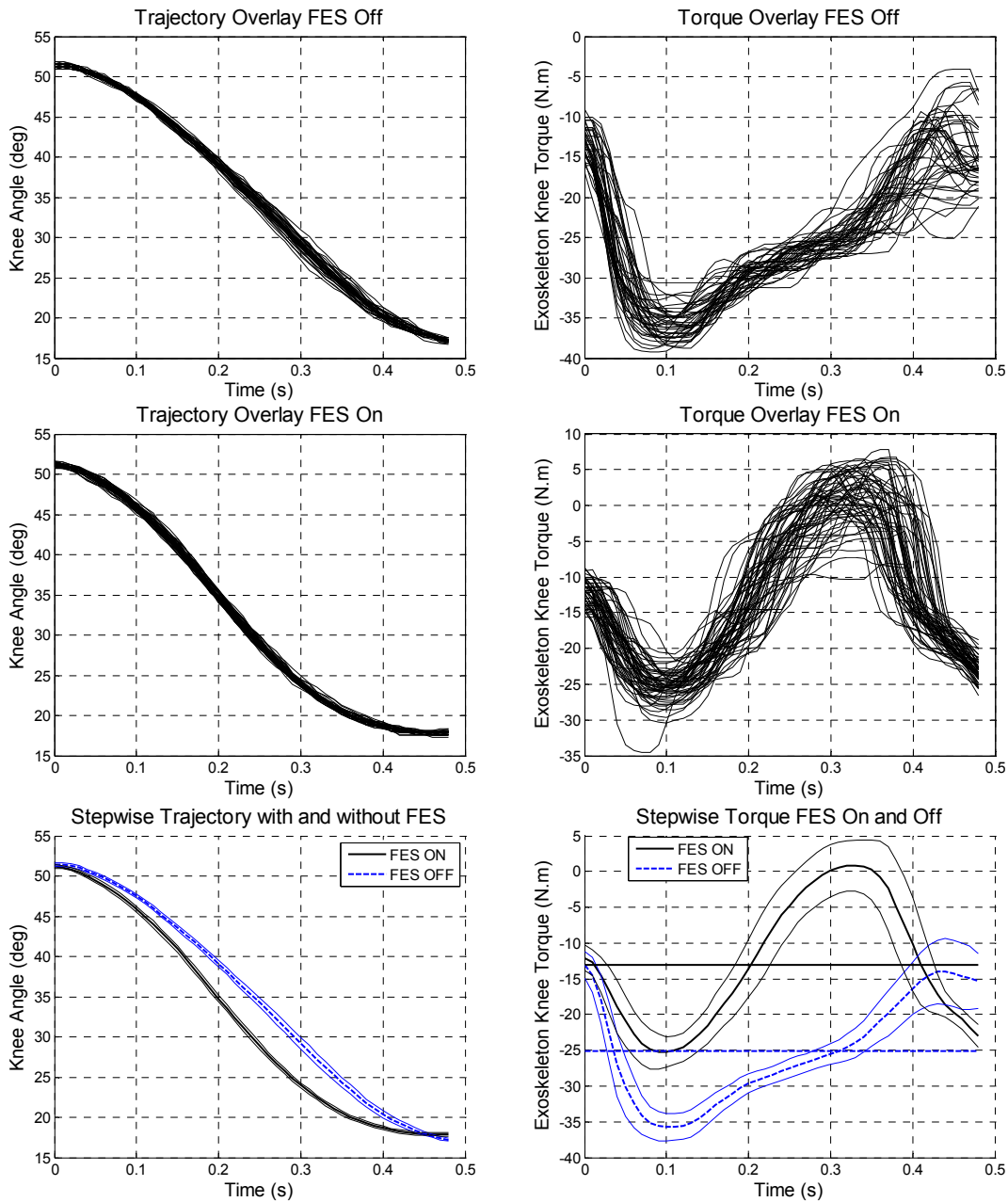
Figure 5-6. Top row: Series of 42 overlaid knee extension trajectories during leg lifts without FES, along with corresponding knee motor torque. Middle row: Series of 42 overlaid knee extension trajectories during leg lifts with FES, along with corresponding knee motor torque. Bottom row: Average trajectory tracking and average motor torque during a series of leg lifts, each bracketed by a standard deviation.

The middle row in Figure 5-6 shows the overlaid trajectories from 42 leg lifts with FES, along with the exoskeleton motor torque required to achieve these trajectories. Finally, the bottom row in Figure 5-6 shows the average of the overlaid trajectories and torques, along with a plus and minus one standard deviation in each. As indicated in the bottom left plot, the trajectory tracking is essentially the same in both cases, although somewhat better with FES than without. The bottom right plot shows the average torque required by the exoskeleton knee motors for the case with and without FES. The average motor torque required without FES was approximately 25 Nm, while the average motor torque required with FES was approximately 13 Nm. Computing the energy delivered by the knee motor (by computing power and integrating over the duration of the trajectory), one finds that on average, the knee motor provided 9.9 J of mechanical energy per movement without FES, and 4.4 J of mechanical energy per movement with FES. By implication, the stimulated quadriceps group provided approximately 12 Nm of average torque during the movement, and approximately 5.4 J of mechanical energy per knee extension. Thus, approximately half of the torque is being provided by the stimulated quadricep, and approximately 55% of the energy for movement was provided by the metabolic power supply when using the cooperative control approach.

Sit-to-stand maneuver

Figure 5-7 shows the subject performing the sit-to-stand maneuver experiment. Figure 5-8 shows the overlaid trajectories from 10 sit-to-stand maneuvers with and without FES, along with the exoskeleton motor torque required to achieve these maneuvers. Specifically, the top row in Figure 5-8 shows the overlaid trajectories from 10 sit-to-stand maneuvers without FES, along with the exoskeleton motor torque required to achieve these trajectories. The middle row in Figure 5-8 shows the overlaid trajectories from 10 sit-to-stand maneuvers with FES, along with the exoskeleton motor torque required to achieve these trajectories. Finally, the bottom row in Figure 5-8 shows the average of the overlaid trajectories and torques, along with a plus and minus one standard deviation in each.

Figure 5-7. Sit-to-stand cooperative control experiment.
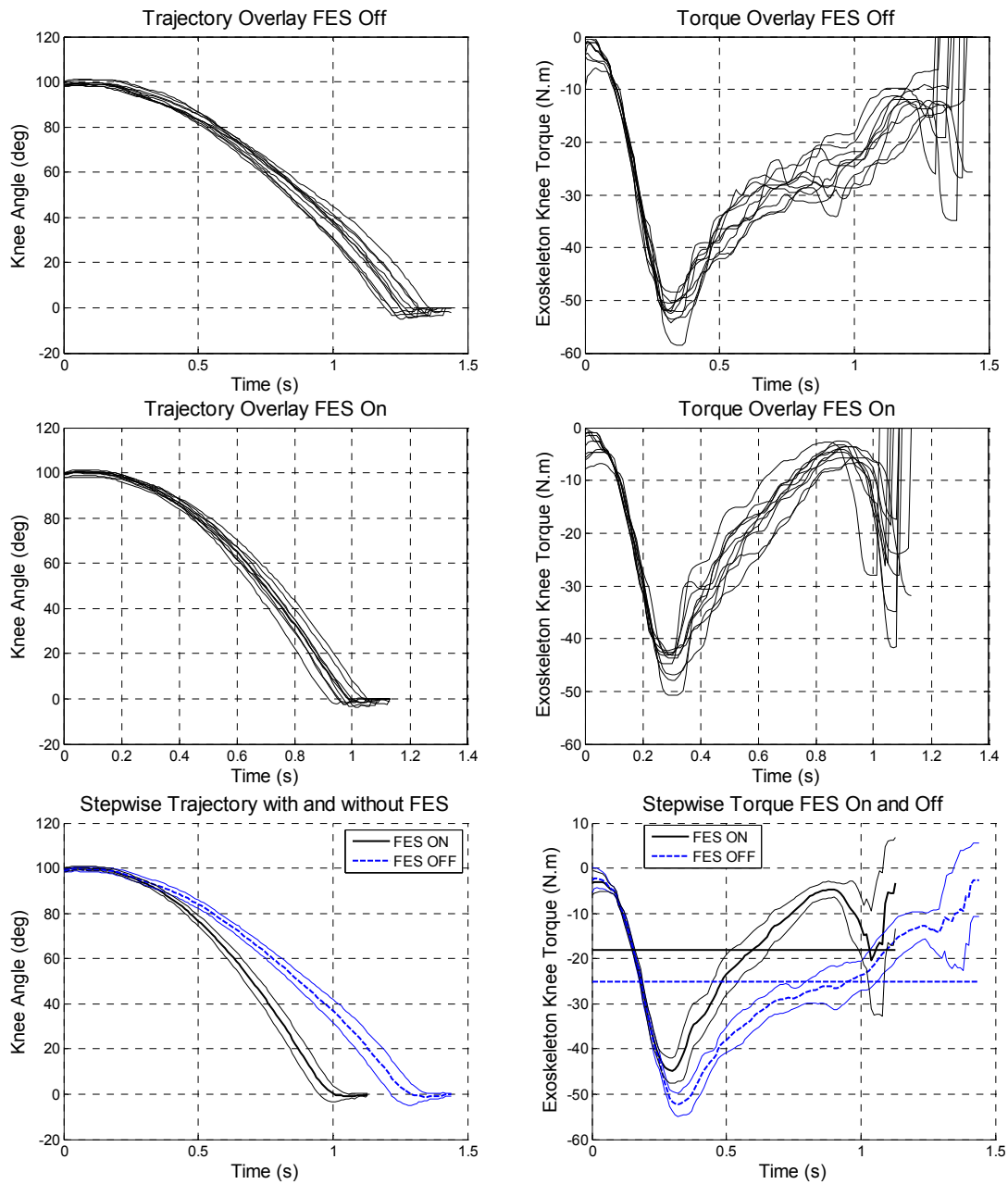
Figure 5-8. Top row: Series of 10 overlaid knee extension trajectories during sit-to-stand without FES, along with corresponding knee motor torque. Middle row: Series of 10 overlaid knee extension trajectories during sit-to-stand with FES, along with corresponding knee motor torque. Bottom row: Average trajectory tracking and average motor torque during a series of sit-to-stand maneuvers, each bracketed by a standard deviation.

As previously stated, the sit-to-stand controller utilizes a step command for the joint angle (rather than a continuous trajectory), and therefore the trajectories with FES, which are able to provide increased torque

relative to those without FES, are characterized by a faster step response, as indicated in the bottom left plot. The bottom right plot shows the average torque required by the exoskeleton knee motors for the case with and without FES. The average motor torque required without FES was approximately 25 Nm, while the average motor torque required with FES was approximately 18 Nm. Due to the increased available torque, the maneuver with FES requires only approximately 78% of the time relative to the maneuver without FES. As such, computing the energy delivered by the knee motor (by computing power and integrating over the duration of the trajectory), one finds that on average, the knee motor provided 37 J of mechanical energy per movement without FES, and provided 22 J of mechanical energy per movement with FES. By implication, the stimulated quadriceps group provided approximately 15 J of mechanical energy per knee extension. Thus, assuming the mechanical energy was invariant between the two cases (i.e., the same body mass was lifted the same distance), approximately 40% of the energy for movement was provided by the metabolic power supply when using the cooperative control approach.

Stair ascent

Figure 5-9 shows the subject performing the stair ascent maneuver experiment. Figure 5-10 shows the overlaid trajectories from 11 stair ascent maneuvers with and without FES, along with the exoskeleton motor torque required to achieve these maneuvers. Specifically, the top row in Figure 5-10 shows the overlaid trajectories from 11 stair ascent maneuvers without FES, along with the exoskeleton motor torque required to achieve these trajectories. The middle row in Figure 5-10 shows the overlaid trajectories from 11 stair ascent maneuvers with FES, along with the exoskeleton motor torque required to achieve these trajectories. Finally, the bottom row in Figure 5-10 shows the average of the overlaid trajectories and torques, along with a plus and minus one standard deviation in each.

Figure 5-9. Stair ascent cooperative control experiment.
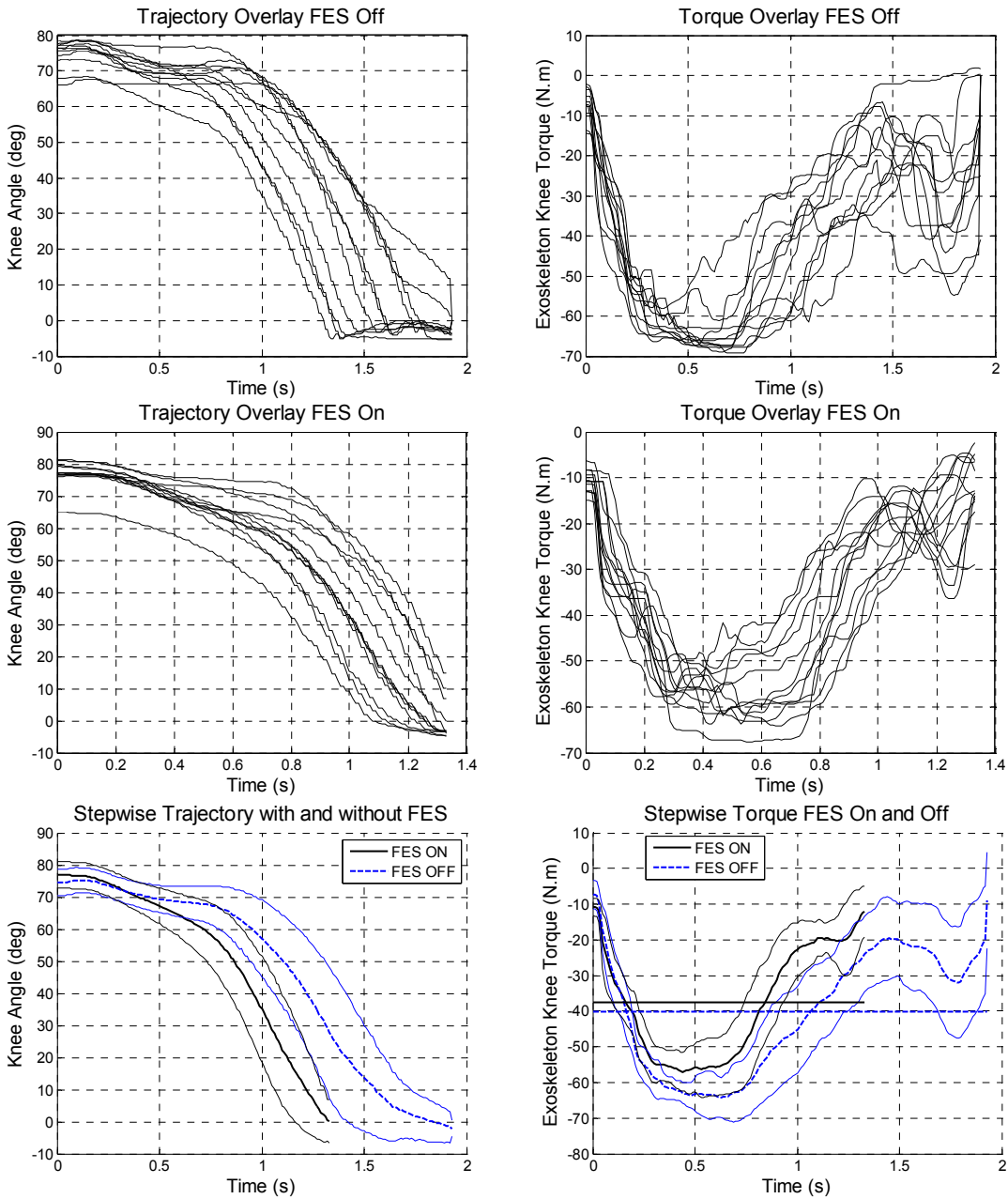
Figure 5-10. Top row: Series of 11 overlaid knee extension trajectories during stair ascent maneuvers without FES, along with corresponding knee motor torque. Middle row: Series of 11 overlaid knee extension trajectories during stair ascent maneuvers with FES, along with corresponding knee motor torque. Bottom row: Average trajectory tracking and average motor torque during a series of stair ascent maneuvers, each bracketed by a standard deviation.

As previously stated, the stair ascent controller utilizes a step command for the joint angle (rather than a continuous trajectory), and therefore (as in sit-to-stand) the trajectories with FES, which are able to

provide increased torque relative to those without FES, are characterized by a faster step response, as indicated in the bottom left plot. The bottom right plot shows the average torque required by the exoskeleton knee motors for the case with and without FES. The average motor torque required without FES was approximately 40 Nm, while the average motor torque required with FES was approximately 37 Nm. However, as previously mentioned, the maneuver with FES requires only approximately 75% of the time relative to the maneuver without FES. As such, computing the energy delivered by the knee motor (by computing power and integrating over the duration of the trajectory), one finds that on average, the knee motor provided 40 J of mechanical energy per movement without FES, and provided 32 J of mechanical energy per movement with FES. By implication, the stimulated quadriceps group provided approximately 8 J of mechanical energy per knee extension. Thus, assuming the mechanical energy was invariant between the two cases (i.e., the same body mass was lifted the same distance), approximately 25% of the energy for movement was provided by the metabolic power supply when using the cooperative control approach.

Results summary

A summary of the experiments, and the average torque and energy exerted by the knee motors in each case, is given in Table 5-1.Table 5-1. Summary of results for cooperative control experiments. The table also shows a summary of the energy reduction provided in each case when using the FES cooperative controller.

Experiment	FES	Repetitions	Av Motor Torque	Time of Movement	Mechanical Energy from Motor	Energy Reduction with FES
Leg Lifts	With FES	42	13.0	0.48	4.4	55.9%
	Without FES	42	25.1	0.48	9.9	
Sit-to-Stand	With FES	10	18.0	1.1	21.7	41.3%
	Without FES	10	25.1	1.4	36.9	
Stair Ascent	With FES	11	37.5	1.3	31.9	18.6%
	Without FES	11	40.2	1.9	39.2	

Table 5-1. Summary of results for cooperative control experiments.

Conclusion

This paper proposes an adaptive, cooperative controller that coordinates electrical stimulation of muscle with a powered exoskeleton. The efficacy of the controller was evaluated in a series of experiments on a T10 complete paraplegic, in which the controller was used at the knee joint of the powered exoskeleton, in conjunction with stimulation of the quadriceps muscle group. The respective experiments represented both lightly loaded and heavily loaded joint conditions. In these experiments, when cooperative using FES, the cooperative controller provided knee joint control that was in every case at least as good as without FES. In the lightly loaded experiment, approximately 55% of the energy for movement was provided by the metabolic power supply through the contribution of the stimulated quadriceps. In the heavily loaded experiments, the contribution of the quadriceps was essentially saturated at its maximum level; in these cases, the energy contribution was approximately 40% and 25%, respectively. In all cases, the controller appears to have effectively combined the respective efforts of the stimulated muscle and the externally powered actuator of the exoskeleton. It is hoped that effectively combining the human and robotic actuators will provide the robustness and control benefits of the powered exoskeleton, whilst also providing the physiological and energetic benefits of FES.

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CHAPTER VI

Conclusion

This document reports the design and implementation of a fully powered lower limb exoskeleton supplemented with FES. Through the different chapters, the effectiveness in allowing a paraplegic patient to stand and walk has been shown. Furthermore the work has shown the successful combination of electrical actuators with the action of the patient muscles by using FES. The main contributions of this dissertation have been stated in the conclusion section of each chapter and can be summarized as follows:

- A mechanical and electrical design for a light-weight, low-profile, electrically powered lower limb exoskeleton with onboard power, sensors, actuators and computational capabilities was developed.

- An embedded system architecture was designed to enable operation of the lower limb exoskeleton.

- The ability of the exoskeleton to provide over ground walking to a complete paraplegic subject was validated on a T10 complete paraplegic patient.

- An intuitive user control interface for sitting, standing, and walking movements, which are controlled by changes in the user's center of pressure, was developed.

- A two layer control architecture for a powered lower limb exoskeleton was developed. The low level controller is a closed loop PD position controller around each joint, which uses the commanded position, proportional, and derivative gains provided by the high level controller. The high level controller is an intent recognizer based on a state machine. This layer infers the user intent based on the user posture, and determines the appropriate trajectories and gains for each joint.

- The exoskeleton was shown to afford similar mobility and require a somewhat lower level of exertion relative to long-leg braces with a swing-through gait, and afford significantly improved mobility with significantly less exertion relative to long-leg braces with a reciprocal gait.

- A cooperative controller that combines FES with electrical motors was developed. The data obtained in the evaluation of the cooperative controller showed a significant torque contribution of the quadriceps muscle during knee extension, without negatively affecting the command trajectories.