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Control and Implementation 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, along with a user interface and control structure that enables control of the powered orthosis via upper-body influence. The orthosis and control structure was experimentally implemented on a paraplegic subject (T10 complete) in order to provide a preliminary characterization of its capability to provide basic walking. Data and video is presented from these initial trials, which indicates that the orthosis and controller are able to effectively provide walking within parallel bars at an average speed of 0.8 km/hr.**

Index Terms--lower limb exoskeleton, assistive technology, powered orthosis, SCI, paraplegia.

I. 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, at least 44% (at least 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 (average) 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 engineering 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 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 [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 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 [12-19].

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 [20-22]. More recently, Ruthenberg [23] developed a powered orthosis for evaluating design requirements for paraplegic gait assistance. In [24-26], 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 [27-30], 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

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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 (at this point) 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 aforementioned devices, is intended to provide gait assistance to paraplegics by providing assistive torques at both hip and knee joints. This paper specifically describes the powered orthosis and a control structure that enables a user to control the orthosis by using his or her upper body to influence his or her body posture. Data is presented from a preliminary experimental implementation, which indicates that the orthosis and controller are able to effectively provide basic walking within a set of parallel bars.

II. Powered Orthosis Prototype

The powered lower limb orthosis, shown in Fig. 1, 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. 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 Fig. 1, hook-and-loop straps on the hip segment, thigh segments, and shank segments secure the orthosis to the user. The total orthosis mass is 12 kg (26.5 lbs).

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. The DES is powered by a 29.6 V, 3.9 A·hr lithium polymer battery, and as indicated in Fig. 2, 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. 3.

Fig. 1. Powered orthosis prototype.

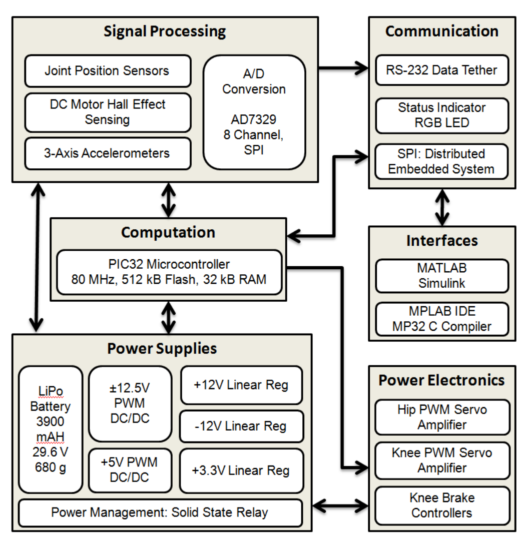

Fig. 2. Embedded system framework.

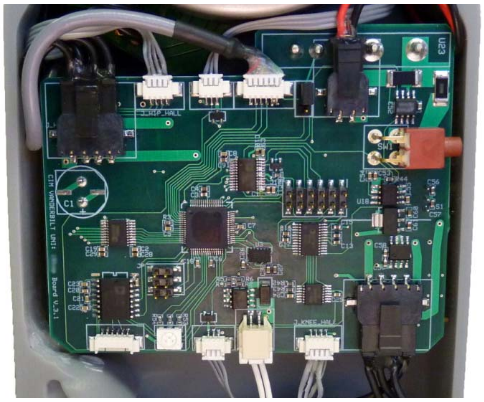

Fig. 3. Embedded system circuit board.

III. Orthosis Control

A. Joint-level Controllers

The orthosis controller consists of a set of low-level (i.e., joint-level) controllers, which are supervised by a high-level control structure that infers intent from the user, and based on the inferred intent, provides the appropriate joint functionality. The joint-level controllers consist of variable-gain proportional-derivate (PD) feedback controllers around each 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 backdrivable behavior of the joint actuator and transmission units, 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 can be used in finite states where it may be desirable to enforce a predetermined trajectory (e.g., during the swing phase of gait), while the latter can be used when it may be preferable not to enforce a pre-determined joint trajectory, but rather provide assistive torques that facilitate movement toward a given joint equilibrium point (e.g., providing joint damping during stand-to-sit maneuver).

B. Finite-state Control Structure

The joint-level controller receives trajectory commands, as well as PD gains, from the high-level-controller, which is a finite state machine (FSM) consisting of 12 states, as shown in Fig. 4. The FSM consists of two types of states – static states and transition states. The static states consist of sitting, standing, right-leg-forward (RLF) double support, and left-leg-forward (LLF) double support. The remaining 8 states, which transition between the four static states, include sit-to-stand, stand-to-sit, stand-to-walk with right half step, stand-to-walk with left half step, walk-to-stand with left half step, walk-to-stand with right half step, right step, and left step. Importantly, each state is fully defined by the combination of a set of trajectories, and a set of 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. 5. For all the trajectories shown in Fig. 5, the joint feedback gains are set high. The joint angles corresponding to the static states of RLF and LLF double support, and standing, correspond to the final angles of the trajectories shown respectively in Fig. 5. 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 is defined by zero gains, and therefore the joint angles are unimportant. The transition from stand-to-sit consists of a zero proportional gain and a high derivate gain (i.e., by damping without any stiffness). In this manner, the joint angles are also unimportant, assuming they are constant. Finally, the sit-to-stand state is defined by standing 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 I and Fig. 5 summarize the trajectories and nature of the feedback gains that together define completely the behavior in all states of the FSM shown in Fig. 4.

C. User/Orthosis Control Interface

The finite state controller defined by Fig. 4 and Table I requires a means of appropriately and safely switching from one to the next state, based on user commands. In order to provide a pathway for user control, an approach was developed based on the user's ability to affect his or her center of pressure via his or her upper body (in combination with his or her stability aid). Specifically transitioning between states is based on the location of the 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. 6, which indicates the estimation of CoP based on the sensors included within the powered orthosis. Specifically, the hip and knee angles are measured as indicated in the embedded system section, while the thigh absolute orientation (α) is obtained via the three axis accelerometer also described in the embedded system section. It is assumed that, with the use of the stability aid, the user can control the tilt of his or her upper body, and thus can control the location of the (estimated) CoP relative to the forward foot. Thus, this distance between the CoP and the location of the forward ankle joint is utilized as the primary control input, with which the user can command all the transitions entailed in the FSM, as indicated in Fig. 4. Specifically, the transition while walking to take the next step (right or left) is indicated by the user leaning forward (using the stability aid), such that the CoP lies within a predefined distance of the forward foot, at which point the FSM will enter either the right step or left step states, depending on which foot started forward. When transitioning from standing to walking, the user will additionally lean to one side, which indicates that the FSM should take a half step forward with the leg which is more unweighted (as detected by the 3-axis accelerometer, which is in this case utilized as a frontal plane tilt sensor). The transition from walking to standing is indicated by a pause (in double support) of sufficient time, which indicates to the FSM that the user wishes to take a half step forward into the standing static state. The transition from standing to sitting is based on a given amount of backward lean (i.e., the CoP reaches a threshold distance behind the ankle joints). Finally, transition from sitting to standing is enabled by a given amount of forward lean (i.e., the CoP reaches a threshold distance of proximity to the ankle joints). A summary of all switching conditions, governing the user interface with the FSM controller, is given in Table II.

TABLE I
STATE DESCRIPTION

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

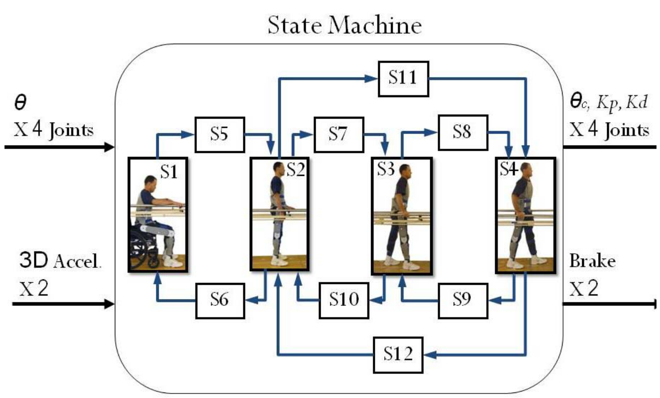

Fig 4. Finite state machine for sitting, standing, and walking.

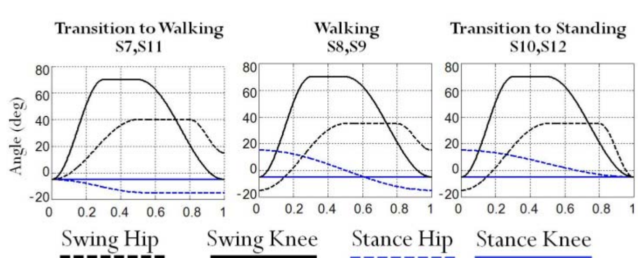

Fig 5. Walking trajectories.

IV. Experimental Implementation

The previously described orthosis prototype and control interface were implemented on a single paraplegic subject to provide a *preliminary* validation of the orthosis and controller.

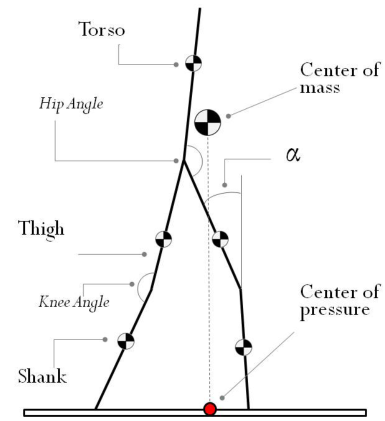

Fig 6. Center of Pressure.

TABLE II
FINITE STATE MACHINE SWITCHING CONDITIONS

Transition	Condition
S1 to S5	The user leans forward and pushes himself up.
S5 to S2	The four joints meet the Standing (S2) configuration.
S2 to S7	The user leans forward and to the left.
S7 to S3	The four joints meet the Right Forward (S3) configuration.
S3 to S8	The user leans forward.
S8 to S4	The four joints meet the Left Forward (S4) configuration.
S4 to S9	The user leans forward.
S9 to S3	The four joints meet the Right Forward (S3) configuration.
S3 to S10	The user keeps the torso vertical during four seconds, then leans forward.
S10 to S2	The four joints meet the Standing (S2) configuration.
S2 to S6	The user leans backward.
S6 to S1	A preset timer is over.
S2 to S11	The user leans forward and to the right.
S11 to S4	The four joints meet the Left Forward (S4) configuration.
S4 to S12	The user keeps the torso vertical during four seconds, then leans forward.
S12 to S2	The four joints meet the Standing (S2) configuration.

A. Efficacy of User Interface

The control interface defined by Fig. 4, Table I, and Table II, was implemented on the lower limb powered orthosis (previously described) and tested on a T10 complete paraplegic subject. The subject was a 35 year-old male, 9 years post-injury, 1.85 m tall, and with a body mass of 73 kg. The experiments were conducted in parallel bars with the subject starting in a sitting position from his wheelchair. Figure 7 shows the subject standing and walking, respectively, with the powered orthosis. During these trials, the subject was asked to rise from sitting to standing, then walk forward to the end of the parallel bars, then transition from double-support to standing, then return to sitting in the wheelchair (which was brought to the end of the bars). In this manner, the subject was required to move throughout the state machine. It should be noted that a couple states are not represented in Fig. 8, since the subject can at times choose to start and/or end with either a right or a left step, depending on preference, and depending on the length of the parallel bars. In the case shown, the subject choose to start with a right step (i.e., choose state 7 instead of 11), and choose to end with a right step (i.e., choose state 12 rather than 10). The sequence shown in Fig. 8, which is representative of all data taken during these trials, indicates that the CoP-based finite state control offers an effective means of user interface and control, without requiring (at least for these movements) explicit commands from the user. Additionally, this is qualitatively conveyed in the attached video.

B. *Efficacy of Powered Orthosis*

In order to assess the nature of gait provided by the orthosis, the subject was asked to walk repeated laps in the parallel bars. The measured hip and knee joint angles from 46 steps (23 right and 23 left steps) is shown (overlaid) in Fig. 9. 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 triggering the next step (i.e., for satisfying the conditions S8 or S9, as described in Table II. The data in Fig. 9 indicates that the powered orthosis is able to provide repeatable joint angle trajectories, which are reflective of the joint kinematics during healthy legged locomotion.

C. *Walking Speed*

The gait represented by the data in Figs. 8 and 9 is characterized by an average overground walking speed of 0.22 m/s (0.8 km/hr or 0.5 mi/hr). Note that approximately half of the stride time consisted of the time required for the subject to adjust his posture between commanded steps.

D. *Electrical Power Consumption*

The total average electrical power required by the system during walking was 117 W. As previously described, the high-level control (i.e., the finite state machine in Fig. 4) is currently implemented on a laptop computer. As such, the 117 W includes all power required by the orthosis, including the embedded system components, but does not include the computational power required to implement the finite state machine. Based on the presently unused computational capacity of the two PIC32 microcontrollers, in combination with the fact that the two controllers require less than one Watt of power, the (eventual) change from laptop host to embedded system implementation of the high-level control should not significantly increase the power consumption of the orthosis.

The battery pack presently embedded in the powered orthosis prototype is a 680 g lithium polymer battery with a 115 W-hr capacity. Based on the average measured electrical power consumption, this 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 increased, if desired, by increasing the size of the battery, since the battery mass (and volume) are currently quite small relative to the overall orthosis mass (i.e., the current battery is approximately 5% of the overall mass).

E. *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).

Fig. 7. Subject with T10 complete spinal cord injury walking wearing the powered orthosis.

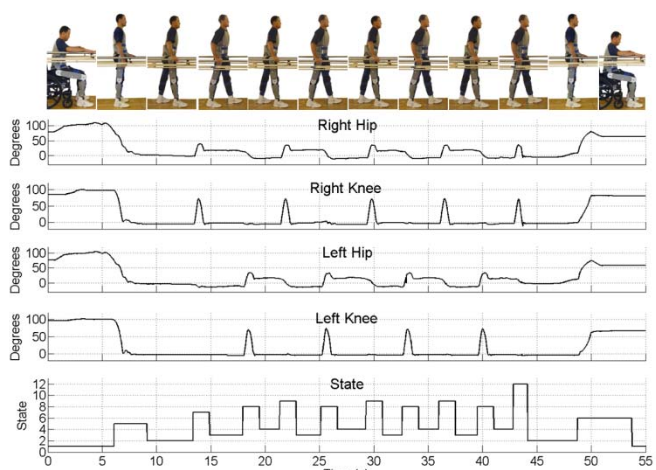

Fig. 8. Hip and knee joint angle data, in addition to state, corresponding to movement throughout the state machine.

V. Conclusion

This paper describes the implementation and control of a new powered lower limb orthosis prototype developed to assist gait in spinal cord injured individuals. *Preliminary* experimental results, conducted within a set of parallel bars on a single T10 complete paraplegic, indicate that the orthosis is capable of effectively providing user-controlled standing, sitting, and walking. Data indicates that the orthosis provides repeatable knee and hip joint angles during walking, which are reflective of the joint trajectory shapes and amplitudes that characterize healthy walking. Electrical power measurements indicate a battery life of approximately one hour, and at the average speed of 0.8 km/hr recorded during the trials, a corresponding walking range of approximately 0.8 km.

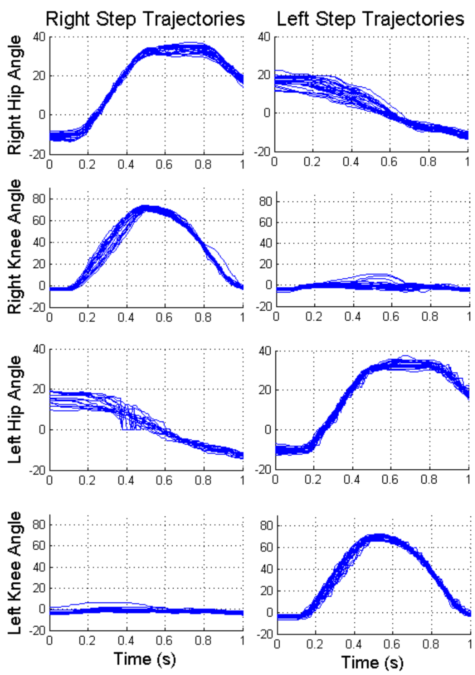

Fig. 9. Measured joint angles during 23 right and 23 left steps.

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