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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 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.

Index Terms

Assistive technology; lower limb exoskeleton; paraplegia; powered orthosis; spinal cord injured (SCI)

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, 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 mod-ulated 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.

II. Vanderbilt Powered Orthosis

The powered lower limb orthosis, shown in Figs. 1 and 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 105° in flexion and 30° in extension, while the range of motion for the knee joints is 105° in flexion and 10° 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. 1 and 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 I.

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. 3. The DES is powered by a 29.6 V, 3.9 A·h lithium polymer battery, and as indicated in Fig. 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 pulsewidth-modulated (PWM) power transistors. One of the two main (twin) DES boards is shown mounted within the thigh link in Fig. 4.

III. Orthosis Control

The orthosis controller consists of a state-flow system with four states, as shown in Fig. 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.

IV. 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, U.S.), 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. Fig. 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.¹ Data indicating hip and knee joint angles and electrical power consumption corresponding to over ground walking are given in the following subsections.

¹See <http://research.vuse.vanderbilt.edu/cim/FarrisTNSRE.wmv>

A. Gait Kinematics

Fig. 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. Fig. 8 shows the same data shown in Fig. 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 [33]) are also plotted in Fig. 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/h or 0.5 mi/h).

B. Electrical Power Consumption

Electrical power consumption was recorded during the walking represented by Figs. 7 and 8. The electrical power required by the servoamplifiers, corresponding to the data shown in Fig. 8, is shown averaged over all 46 steps (or 23 strides) in Fig. 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 II. With a one-second average pause between steps (corresponding to the 0.22 m/s walking data represented by Fig. 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. 9 and Table II, 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 I). 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.

C. 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 approximately 55 ± 2 dBA (with an ambient noise level of 38 dBA).

V. Discussion

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

VI. 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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Fig. 1.
Vanderbilt powered orthosis oblique view.

Fig. 2.
Vanderbilt powered orthosis frontal view.

Fig. 3.
Embedded system framework.

Fig. 4.
Embedded system circuit board.

Fig. 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.

Fig. 6.
Subject with T10 complete spinal cord injury, wearing the powered orthosis.

Fig. 7.
Measured joint angles during 23 right steps and 23 left steps.

Fig. 8. Overlaid joint angles for 23 gait cycles. Dashed vertical line represents an approximately 1-s pause taken by subject between steps. Dashed trajectories represent normal biomechanical data for level ground walking at medium cadence.

Fig. 9.

Average measured electrical power at the knee and hip joints over one gait cycle (average taken from 23 gait cycles). Dashed vertical line represents approximately 1-s pause taken by subject between steps. Dashed horizontal lines represent average electrical power within one gait cycle.

TABLE I**Mass Breakdown of Vanderbilt Orthosis**

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 II**Vanderbilt Orthosis Electrical Power Consumption**

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	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 with a one-second pause between steps (i.e., steps are being taken during 50% of the time during “walking”)