# A Powered Lower Limb Orthosis for Providing Legged Mobility in Paraplegic Individuals

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This article presents preliminary results on the development of a powered lower limb orthosis intended to provide legged mobility (with the use of a stability aid, such as forearm crutches) to paraplegic individuals. The orthosis contains electric motors at both hip and knee joints that, in conjunction with ankle-foot orthoses, provide appropriate joint kinematics for legged locomotion. The article describes the orthosis and the nature of the controller that enables the spinal cord–injured patient to command the device and presents data from preliminary trials that indicate the efficacy of the orthosis and controller in providing legged mobility. **Key words:** *mobility limitation, orthotic devices, rehabilitation, robotics, spinal cord injuries, walking*

One of the most significant impair-ments 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 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. Because 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 by Butler and colleagues.[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 individual with paraplegia leans forward against the stability aid, utilizing gravity to provide hip extension of the stance leg. Because 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 references 6 through 13.

To decrease the high level of exertion associated with passive orthoses, some researchers have investigated the use of powered orthoses that incorporate actuators to assist with locomotion. Historical efforts to develop powered orthoses to aid in paraplegic mobility are cited in references 14 through 16. More recently, Ruthenberg[17] developed a powered orthosis for evaluating design requirements for paraplegic gait assistance. A powered orthosis was developed by combining 3 electric motors with an RGO, 2 of which were located at the knee joints to enable knee flexion and extension during swing, and 1 of which assisted the hip coupling, which in essence assisted both stance hip extension and contralateral swing hip flexion.[18-20] The orthosis was shown to increase gait speed and decrease compensatory motions, relative to walking without powered assistance.

*Top Spinal Cord Inj Rehabil* 2011;17(1):25–33

www.thomasland.com

doi: 10.1310/sci1701-25

Researchers[21-24] 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 refs. 21-24) 6 electric motors (ie, powered sagittal plane hip, knee, and ankle joints). Like the powered lower limb orthosis HAL, 2 additional emerging commercial devices are the Re*Walk* powered orthosis (Argo Medical Technologies, Yokneam Ilit, Israel) and the eLEGS powered orthosis (Berkeley Bionics, Berkeley, California). At the time of writing of this article, studies have not been published regarding the design, performance characterization, or efficacy of these systems.

We have developed a powered lower limb orthosis to enable, with the use of a stability aid (eg, forearm crutches), legged locomotion in patients with paraplegia. This article describes the orthosis and its controller and presents data from preliminary trials on a single paraplegic subject (within parallel bars) that indicate the device can effectively provide legged mobility.

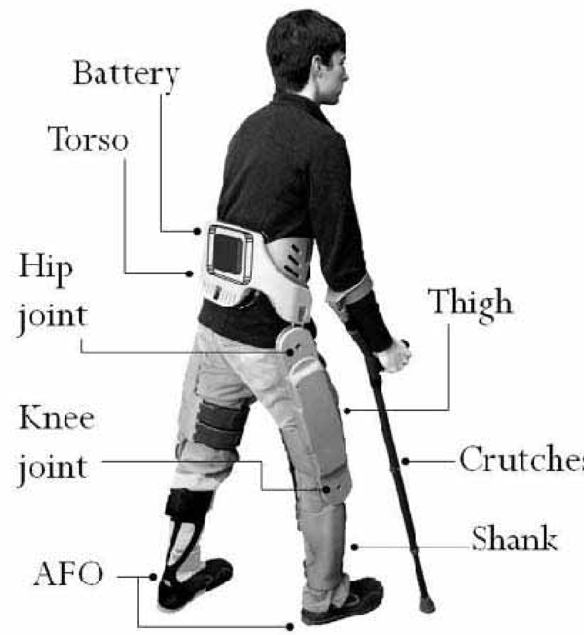


**Figure 1.** Powered orthosis prototype.

**Powered Orthosis Prototype**

The lower limb powered orthosis is shown in **Figure 1**. The user is expected to have sufficient upper body strength to maintain balance with a walker or forearm crutches. The orthosis incorporates powered hip and knee joints. A hip piece surrounds the lumbar and abdominal areas, enabling the orthosis to impose a hip joint torque and providing support to the upper body. The thigh pieces are connected to the hip piece by the hip joints, and each thigh is connected to the respective shank piece through the knee joint. The 4 articulations (right and left hip and knee joints) provide torque in the sagittal plane while restricting rotation in the coronal and transverse planes. The hip and knee joints are powered by 2 brushless DC motors through a transmission, which enables each joint to provide up to 12 Nm of torque continuously and 40 Nm of torque for shorter (ie, 2 s) durations. As a safety measure, both knee joints include normally locked brakes to preclude knee buckling in the event of a power failure. The system is designed to be used in conjunction with a standard AFO to provide stability at the ankle and to preclude footdrop during the swing phase of gait.

The orthosis is controlled by on-board electronics, with components distributed in the hip and both thigh pieces. To expedite control development and facilitate data collection and analysis, high-level control is implemented on a host computer (PC), as illustrated in **Figure 2**. A functional schematic of the control electronics (also called the distributed embedded system [DES]) is shown in **Figure 3**. The DES is powered by a lithium polymer battery, which is located in the lumbar area of the hip piece (see **Figure 1**). The primary functional elements of the DES include data processing (and control), sensor interface, power electronics, communication, and power management. Specifically, the data processing element consists primarily of a 32-bit microcontroller that operates at 80 MHz and contains 512 kB flash memory and 32 kB RAM. The sensor interface includes appropriate interface for the Hall effect sensors (for control of the brushless motors), potentiometer interfaces for measurement of joint angles in each of the 4 articulated joints, and 3-axis accelerometer interface. Note that battery voltage and motor currents are also monitored via a 12-bit A/D converter. The power electronics consist of a regenerative switching servoamplifier for each brushless motor and a spike and hold PWM driving circuit for each knee brake. The communication within the DES utilizes a serial peripheral interface (SPI) protocol, whereas an RS232 serial protocol is used to exchange data between a host PC and the DES. Finally, power management includes a combination of switching and linear regulators, with which the system provides various voltages (ie, 30V, 12V, -12V, 5V, 3.3V, and 2.5V) that are required by the DES.

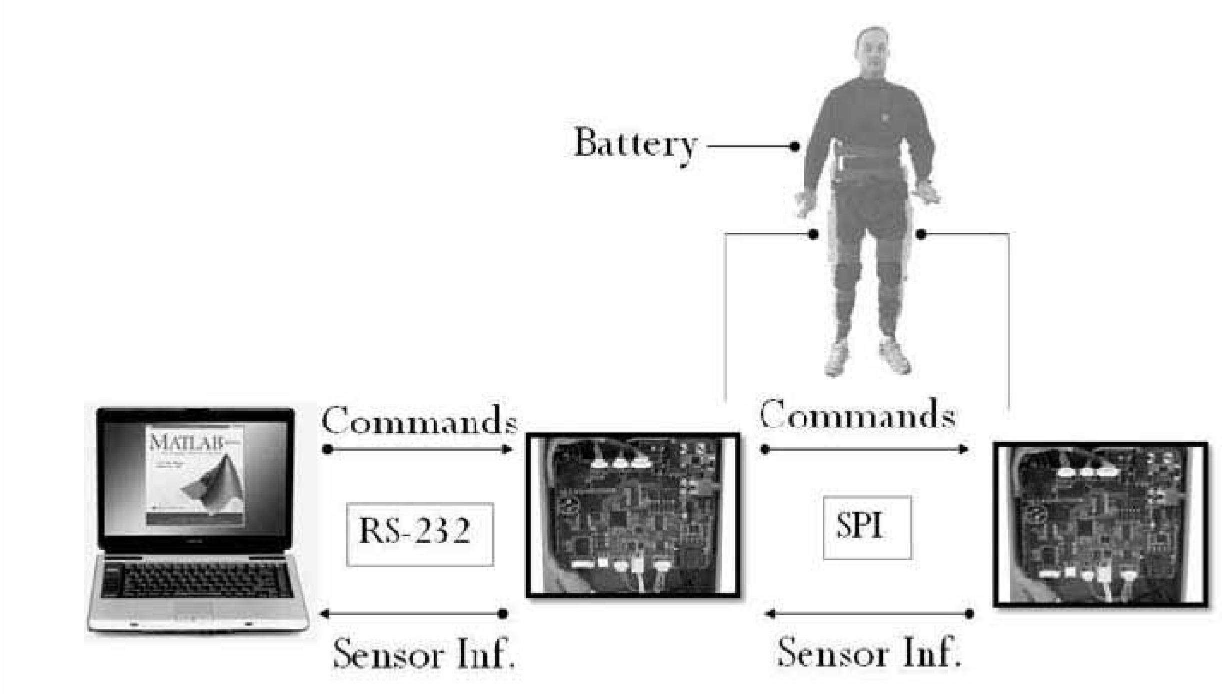


**Figure 2.** Layout of control hardware and distributed embedded system.

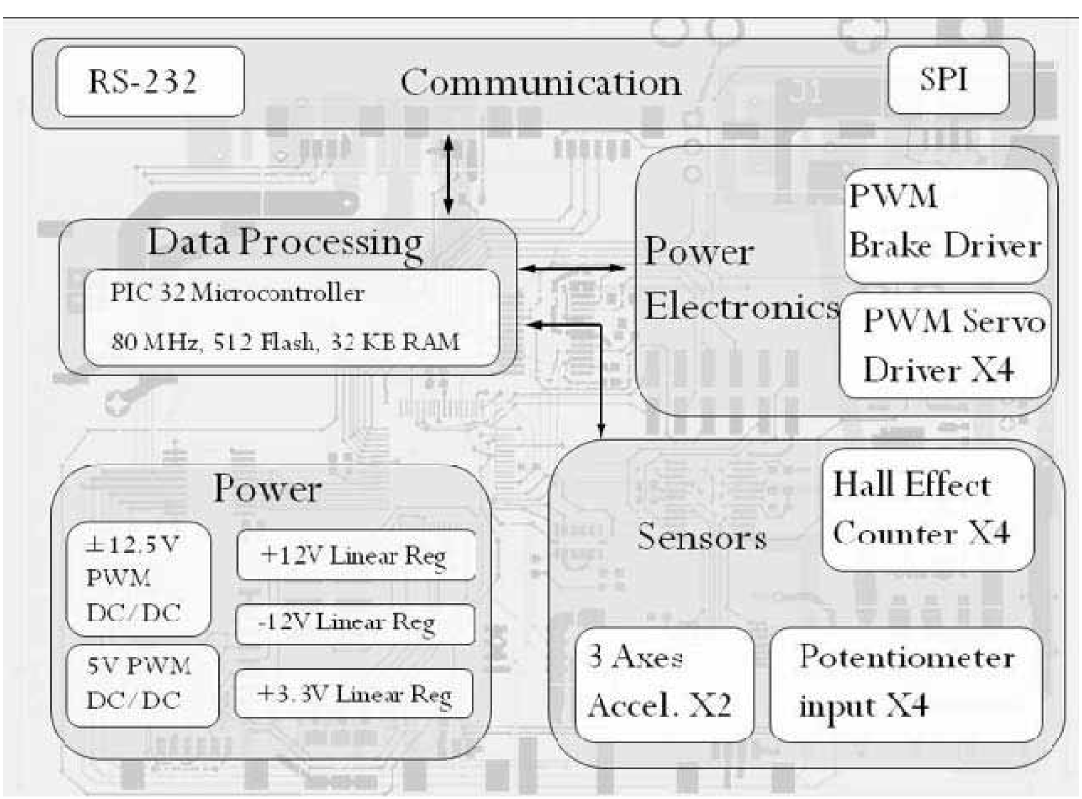


**Figure 3**. Functional schematic of distributed embedded system.

**Powered Orthosis Control Architecture**

The orthosis controller consists of a set of low-level (ie, joint-level) controllers, which are supervised by a high-level control structure; the structure 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 consist 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 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 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 (eg, during the swing phase of gait), whereas the latter can be used when it may be preferable not to enforce a predetermined joint trajectory, but rather to provide assistive torques that facilitate movement toward a given joint equilibrium point (eg, providing joint damping during stand-to-sit maneuver).

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 **Figure 4**. The FSM consists of 2 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 4 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. Each state 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 6 of the 8 transition states are shown in **Figure 5**. For all the trajectories shown in **Figure 5**, the joint feedback gains are set high. The joint angles maintained in the static states of RLF and LLF double support and standing correspond to the final angles of the trajectories shown, respectively, in **Figure 5**. Three states remain, which are the static state of sitting and the 2 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 (ie, 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 1** and **Figure 5** summarize the trajectories and nature of the feedback gains that together define completely the behavior in all states of the FSM shown in **Figure 4**.

The finite state controller defined by **Figure 4** and **Table 1** is not complete without a means of appropriately switching from one state to the next state, based on user commands. Such an interface should be intuitive, reliable, and controllable. The authors have developed a control interface that enables the user to control the FSM based on movements of the upper body (in conjunction with the stability aid, such as forearm crutches), as measured by sensors on the lower limb orthosis. Specifically, transitioning between states 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 **Figure 6**, which indicates the estimated computation of CoP based on the sensors included in the powered orthosis. Specifically, the hip and knee angles are measured as indicated in the embedded system section, while the thigh absolute orientation ($\alpha$) is obtained via the 3-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 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 **Figure 4**. Specifically, the transition 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 that 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

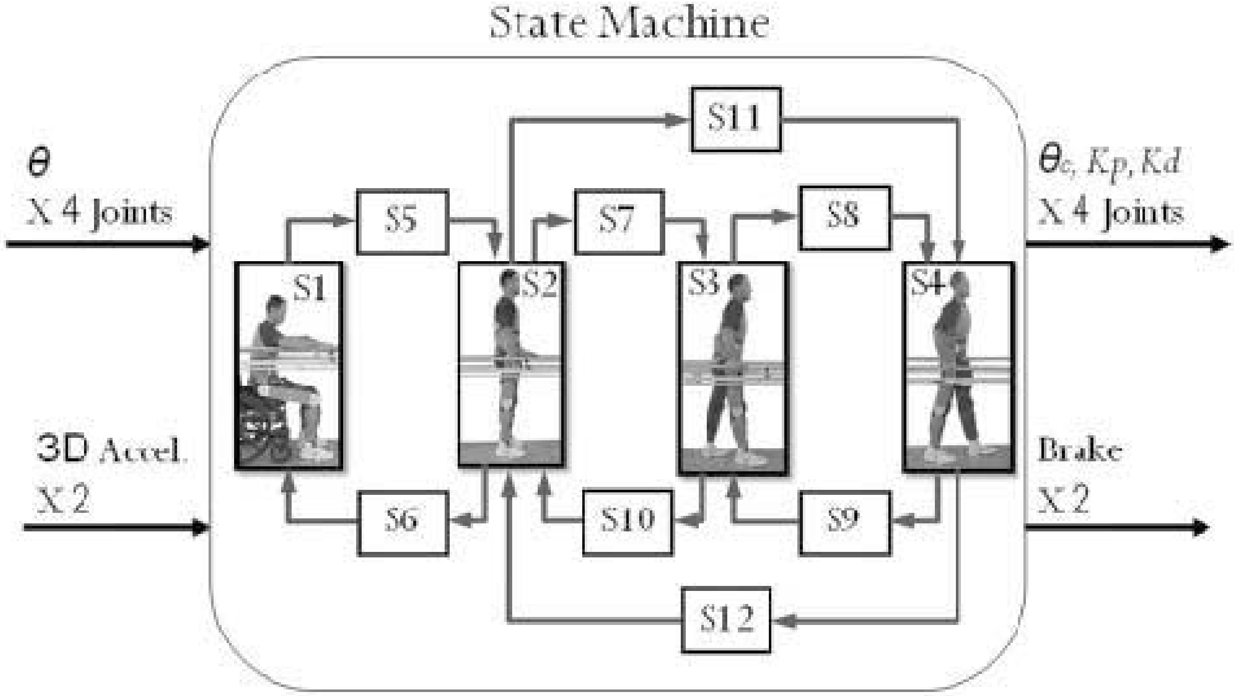


**Figure 4.** Finite state machine for sitting, standing, and walking.

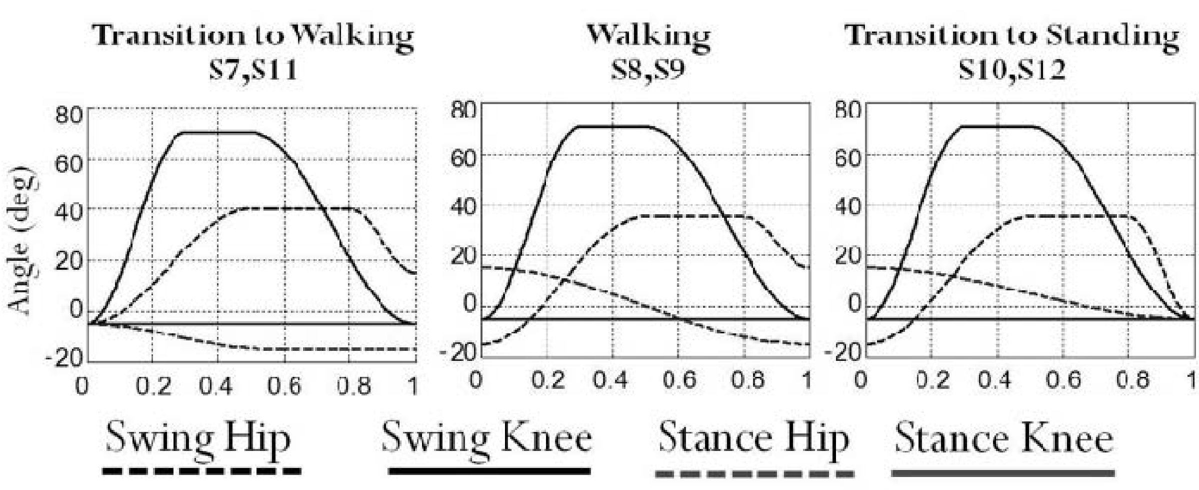


**Figure 5.** Hip and knee joint angle trajectories corresponding to various states.

**Table 1.** 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 |

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 (ie, 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 (ie, 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 2**.

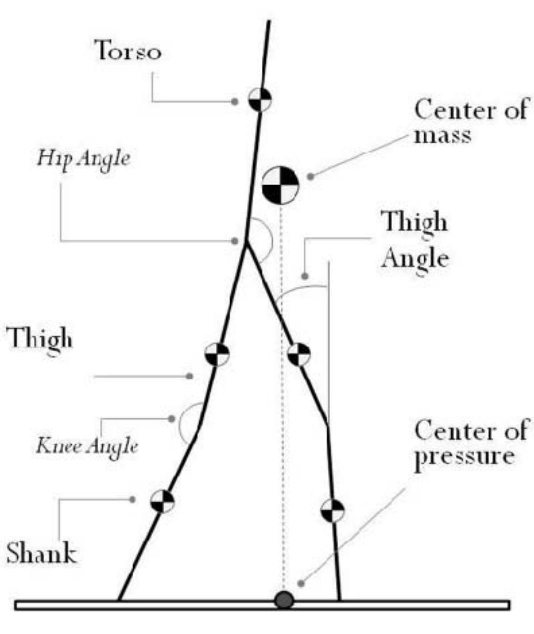


**Figure 6.** Estimation of center of pressure from orthosis sensors.

### Experimental Implementation

To provide a preliminary investigation of the ability of the orthosis and controller to provide legged mobility, the orthosis was tested on a T10 complete paraplegic subject. The subject was a 35-year-old male, 9 years post injury, 1.85 m (6 ft, 1 in) tall, and with a body mass of 73 kg (160.9 lbs). The experiments were conducted in parallel bars with the subject starting in a sitting position from his wheelchair. During these trials, the subject was asked to rise from sitting to standing, walk forward to the end of the parallel bars, transition from double support to standing, and then return to sitting in the wheelchair (which was brought to the end of the bars). **Figure 7** shows representative data corresponding to such a sequence. Specifically, the sequence shown in **Figure 7** entails first rising to a standing position (transition from S1 to S2 through S5), taking a right step (S7), alternatively taking left and right steps (S8 and S9), returning to standing through a right step (S12), and finally sitting (S6 to S1). Note that this sequence moves through the entire state chart, with the exception of S10 and S11, which are the left-side choices that correspond to the right-side choices S7 and S12. The subject, in other trials, utilizes either S10 and/or S11 in place of S7 and S12, respectively. As such, preliminary trials on this single subject indicate that measurement of the estimated CoP is sufficient to provide a control user control interface for sitting, standing, and walking movements.

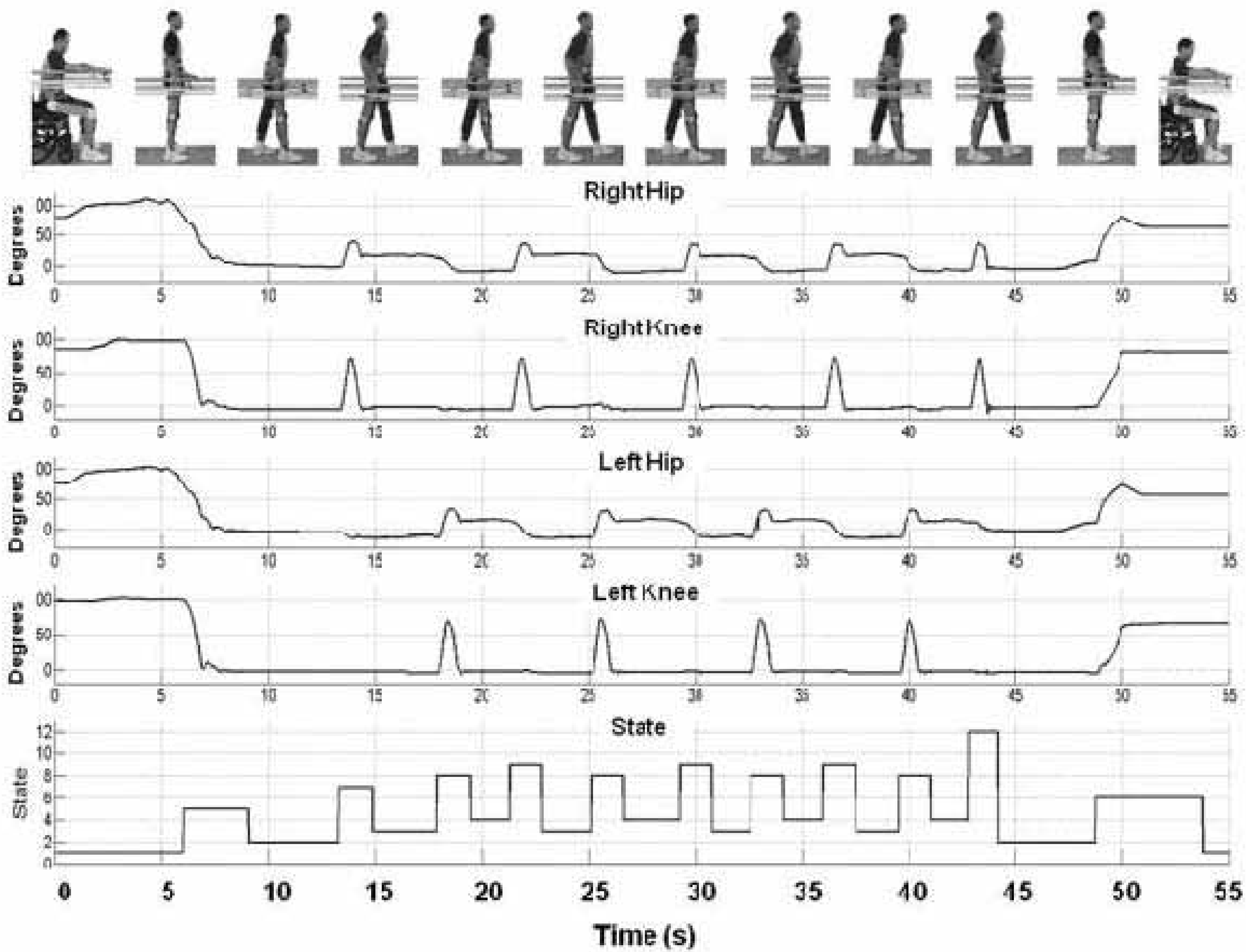


**Figure 7**. Right and left leg hip and knee angle data from a walking trial, along with the corresponding state within the finite state controller.

**Table 2.** Finite state machine switching conditions

| Transition | Condition |
|---|---|
| S1 to S5 | The user leans forward and pushes himself up. |
| S5 to S2 | The 4 joints meet the Standing (S2) configuration. |
| S2 to S7 | The user leans forward and to the left. |
| S7 to S3 | The 4 joints meet the Right Forward (S3) configuration. |
| S3 to S8 | The user leans forward. |
| S8 to S4 | The 4 joints meet the Left Forward (S4) configuration. |
| S4 to S9 | The user leans forward. |
| S9 to S3 | The 4 joints meet the Right Forward (S3) configuration. |
| S3 to S10 | The user keeps the torso vertical during 4 seconds, then leans forward. |
| S10 to S2 | The 4 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 4 joints meet the Left Forward (S4) configuration. |
| S4 to S12 | The user keeps the torso vertical during 4 seconds, then leans forward. |
| S12 to S2 | The 4 joints meet the Standing (S2) configuration. |

## Conclusion

We have developed a powered lower limb orthosis to provide legged mobility to individuals with paraplegia (with sufficient upper body strength to use a stability aid) and have described here a control structure to interface the orthosis with a paraplegic user. Specifically, by using his or her upper body control in conjunction with a stability aid, the user can affect his or her center of pressure, which is estimated by sensors on the orthosis. Preliminary testing on a single paraplegic subject (within parallel bars) indicates the orthosis and control interface offer an effective means of providing sitting, standing, and walking functionality.

## Acknowledgments

This work was supported in part by the US Department of Health and Human Services under National Institutes of Health grant 1R01HD059832-01A1.

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