Design and Simulation of a Joint-Coupled Orthosis for Regulating FES-Aided Gait

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Abstract—A hybrid functional electrical stimulation (FES)/orthosis system is being developed which combines two channels of (surface-electrode-based) electrical stimulation with a computer-controlled orthosis for the purpose of restoring gait to spinal cord injured (SCI) individuals (albeit with a stability aid, such as a walker). The orthosis is an energetically passive, controllable device which 1) unidirectionally couples hip to knee flexion; 2) aids hip and knee flexion with a spring assist; and 3) incorporates sensors and modulated friction brakes, which are used in conjunction with electrical stimulation for the feedback control of joint (and therefore limb) trajectories. This paper describes the hybrid FES approach and the design of the joint coupled orthosis. A dynamic simulation of an SCI individual using the hybrid approach is described, and results from the simulation are presented that indicate the promise of the JCO approach.

I. INTRODUCTION

Previous studies have demonstrated that functional electrical stimulation (FES) can effectively restore legged mobility to spinal cord injured (SCI) individuals (with the help of a stability aid), and that such legged mobility can provide significant physiological and psychological benefits to SCI users [1-18]. Despite this, two significant factors have hindered FES-aided gait systems from restoring gait to SCI individuals. The first is the rapid muscle fatigue that results from artificially stimulated muscle contraction [19], and the second is the inadequate control of joint torques necessary to produce reliable and repeatable limb motion and body support. The former (which significantly influences the latter), is due primarily to the synchronous nature in which artificial stimulation recruits motor units (i.e., is due to lack of neural specificity in the stimulation interface). The net effect is that, when stimulated at a high level of effort and high duty cycle, the muscle quickly (i.e., over tens of seconds) loses its ability to generate force [12]. Both issues (rapid onset of fatigue and poor controllability) can potentially result in collapse of the individual, a condition that is unacceptable in any viable gait restoration system. Additionally, FES-aided gait systems (and especially surface-based systems) generally provide stimulation for degrees of freedom in the sagittal plane, but do not provide control over several other degrees-of-freedom associated with gait, such as hip abduction and adduction in the frontal plane. A lack of control authority in this plane can result in one foot crossing in front of the other (i.e., scissoring), which is a condition that is not easily rectified by the user, and in fact often requires external assistance.

Due primarily to these challenges (i.e., the potential of collapse from muscle fatigue and the need to guide uncontrolled degrees of freedom), hybrid systems, which combine FES with an orthosis, appear to offer the greatest promise for commercially viable gait restoration systems. As such, recent efforts by various researchers have focused (and are focusing) on the development of hybrid systems (e.g., [20-23]). This paper describes a hybrid FES approach that addresses the shortcomings of muscle fatigue and limb trajectory control. The approach utilizes surface stimulation of only the quadriceps muscle group of each leg, along with an energetically passive, controllable orthosis which 1) unidirectionally couples hip to knee flexion; 2) aids hip and knee flexion with a spring assist; and 3) incorporates sensors and modulated friction brakes, which are used in conjunction with electrical stimulation for the feedback control of joint (and therefore limb) trajectories.

II. JOINT-COUPLED CONTROLLED-BRAKE ORTHOSIS (JCO)

The authors are developing a joint-coupled controlled brake orthosis (JCO), shown in Fig. 1, for regulating FES-aided gait. The JCO incorporates multiple features, which serve multiple functions. First, the JCO incorporates unidirectional joint coupling between knee and hip flexion, such that knee flexion generates hip flexion. Since the coupling is unidirectional, however, knee extension does not generate hip extension. The JCO also includes a biasing spring, such that the knee joint (and due to coupling, also the hip joint) is biased toward an equilibrium position in which both the knee joint and the hip joint are flexed. The combination of the (unidirectional) joint coupling and the biasing spring enables knee flexion, hip flexion, and knee extension, all from surface stimulation of only the quadriceps muscle group of each leg. The quadriceps muscle group is among the most powerful and easiest (in the lower
limb) to access via surface stimulation, and thus provides a convenient source of metabolic power for gait. In addition to the joint coupling and biasing spring, the JCO incorporates controllable friction brakes at both knees and hips, which can either independently lock these joints (i.e., to provide for “isometric” muscle contraction), or can modulate the resistive torque at each joint for purposes of controlling limb motion. The JCO also contains angle (and thus also angular velocity) sensing at both hips and knee, which provide essential information for purposes of feedback control of limb motion. Finally, the JCO constrains motion along uncontrolled degrees-of-freedom (e.g., ankle flexion and hip adduction) which enhances the controllability and stability of gait.

A. The JCO Gait Sequence

The gait control approach is described subsequently in the section on gait control and simulation, but is described briefly here to motivate the design of the JCO. Postural stability during gait is provided by a stability aid, such as a walker. The knee of each leg is locked by the controllable friction brakes during stance. Swing is initiated by unlocking the swing leg knee brake, which releases the energy in the biasing spring, which flexes the knee joint and (due to the joint coupling) also flexes the hip joint. During the second half of the swing phase, the hip is locked by the hip brake while the knee is extended by stimulating the quadriceps group. This knee extension (due to stimulation of the quadriceps) does not, however, generate ipsilateral hip extension, since the (cable-based) coupling is unidirectional. Once the knee is fully extended, it remains locked (by the knee brake) during the stance phase of gait.

B. Joint Coupling Design

The purpose of the joint coupling is to provide hip flexion necessary to generate forward leg motion, which is otherwise a challenge, due to the inaccessibility of the deep hip flexor muscles via surface stimulation. The JCO design incorporates a Bowden cable which spans the inside of the femur link and attaches to the hip and knee rotors on either end (see Fig. 2). Cable crimps are used in the hip brake as hard stops in only one direction of rotation, which provides unidirectional coupling of knee flexion to hip flexion. During knee extension, the distal end of the inner Bowden cable winds around the inner knee brake rotor, thus creating a torque in the direction of knee flexion as determined by the spring stiffness, equilibrium point, and preload (against a joint hard stop).

C. Wafer Disc Brakes

A key component of the JCO is the wafer disc brake, which serves a threefold purpose: 1) provide added safety via the normally “locked” design of the knee brake, which will prevent the wearer from falling should the device lose power; 2) increase muscle efficiency by locking joints during phases of gait when they are normally static, thus taking the burden of support off the leg muscles, reducing muscle fatigue and allowing longer walking times; and 3) smooth and control leg trajectories for a more natural and repeatable gait by utilizing the brakes as variable dampers controlled in relation to joint angle feedback. A previous effort to create a controlled brake orthosis [20, 24, 25] utilized magnetic particle brakes, which require electrical power to impose resistive torque. In the event of a power failure, the brakes (and thus the orthosis joints) remain unlocked, which could result in collapse and serious injury to the individual. The authors have developed a new type of brake, called a wafer disc brake (WDB), which provides approximately 45 times the torque-to-weight ratio of state-of-the-art magnetic particle brakes, and importantly, can be designed in either a “normally locked” mode or “normally unlocked” mode. Since the knee joints should fail in a locked mode, as previously mentioned, the knee brakes are thus of the normally locked type. Since the hip brakes are used primarily for trajectory control and are characterized by relatively low duty cycle operation, the hip brakes are of the normally unlocked type. Designing knee brakes to be normally locked and hip brakes to be normally unlocked
both minimizes electrical power consumption (based on data from [25], and importantly prevents collapse during an electrical power failure. The normally unlocked WDB, which was designed for the hip joint, consists of a stack of thin high-strength plastic wafers which are alternatively coupled (through splines) to the brake stator and rotor. A small brushless motor located inside the brake shaft transmits a compressive force through a ball screw to the stack. Assuming relatively low friction in the ball screw, the stack is subjected to a compressive force which is proportional to the motor current. Due to the series arrangement of discs, the resistive torque on the rotor is the product of the compressive force, the mean radius of contact, and the coefficient of friction, which is amplified by the number of interfaces between discs. Since the hip brake contains 61 discs, the effective hip torque is increased by a gain of 60. The net result is a proportional brake that provides a significantly greater torque-to-weight ratio than a state-of-the-art in magnetic particle brake. Since the ball screw is back-drivable, the brake torque remains in proportion to the motor current, and thus is proportional in nature. The normally locked type of WDB, which is used for the knee joint, is shown in cross-section in Fig. 3. A photo of the corresponding assembled prototype is shown in Fig. 4. The design is similar to the normally unlocked type, but the discs are preloaded with a compression spring. Applying current to the motor proportionally unloads the preload, such that full brake torque occurs at zero motor current, and minimum brake torque occurs at full motor current. Since the ball screw is back-drivable, the brake torque remains in inverse proportion to the motor current.

A first-generation prototype of the knee brake has been constructed and tested. The mass of this brake is 0.73 kg. The brake was experimentally measured to provide a maximum torque of 50.7 N-m, which provides a resistive torque-to-weight ratio of 69.4 N-m/kg. In comparison, a state-of-the-art magnetic particle brake (MPB) in a similar size range provides a torque of 1.7 N-m with a mass of 1.14 kg, and as such has a resistive torque-to-weight ratio of 1.5 N-m/kg (e.g., Placid Industries model no. B15). As such, the WDB has a torque-to-weight ratio approximately 45 times greater than the MPB. Experimental measurements further indicate a minimum torque of 0.16 N-m (i.e., the brake dynamic range is between approximately 0.16 and 50 N-m). For both brakes, the torque varies linearly (and inversely, for the knee brake) with input current.

D. Ankle Support

The JCO utilizes an ankle-foot-orthosis (AFO) at the ankle, which is sufficiently compliant to allow dorsiflexion during the stance phase of gait, but sufficiently stiff to prevent foot drop during the swing phase of gait. Current gait simulations indicate a stiffness of 15 Nm/rad (for a 75 kg user) provides an appropriate balance between these objectives.
E. Mass and Inertia

The total orthosis mass as shown in Fig. 1, based on the solid model and prototypes of the brakes, is approximately 6 kg (13 lbs). Approximately one half of the orthosis weight is located on the pelvis, and thus does not add significantly to the rotational inertia or gravitational loads of the lower limbs. The rotational inertia of the distal link of the orthosis about the knee joint is approximately 5% of a typical shank inertia, while the inertia of the proximal link about the hip joint is about 10% of a typical thigh inertia.

III. GAIT CONTROL AND SIMULATION

A dynamic simulation of an SCI individual walking with the JCO and a walker was conducted to 1) validate that the proposed approach can provide a safe and stable gait, 2) develop a robust gait controller, 3) explore variation of primary JCO design parameters, namely, the joint coupling transmission ratio and the spring stiffness, preload, and equilibrium point, 4) assess the magnitude of torque and duty cycle of stimulation required of the quadriceps group (which will help characterize the extent of expected muscle fatigue), and 5) assess the load borne by the arms (through the walker) relative to that carried by the legs.

The dynamic simulator has two main parts — the human body model and the gait controller. The human body model is based in classical rigid body dynamics and is composed of seven segments and six articulations in three dimensional space, as shown in Fig. 5. The associated geometric and inertial parameters normalized to a body height and mass, as given by [26], are listed in Table 1. The spatial model has 12 degrees of freedom, which is reduced to fewer when the feet are in contact with the ground. The model is influenced by 10 inputs, 6 of which are produced by the hybrid orthosis system (i.e., torques in the sagittal plane for both ankles, knees, and hips, applied by a combination of muscle stimulation and JCO input), and 4 of which result from the interaction between the user and the walker (i.e., one torque in the sagittal plane and three forces, all applied symmetrically at the shoulder joints, which are located atop the torso link). Note that the floor is modeled in the vertical direction as a unidirectional stiff spring and damper, such that the feet do not penetrate significantly into the floor (i.e., foot/floor penetration is on the order of a millimeter). In the horizontal direction, the foot/floor interaction is governed by Coulomb friction, where the coefficient of friction is assumed to be 0.3.

A. Walker Model

The user walks with the JCO on a flat and level surface with the aid of a regular walker. Note that the use of a walker assumes sufficient upper limb function to do so. The shoulder force in the frontal plane is modeled as a spring and damper with equilibrium point at the vertical orientation, which stabilizes the torso in the frontal plane (i.e., prevents “falling” in that plane). The remaining shoulder inputs, which are all sagittal plane inputs, are constrained by the nature of the walker. Specifically, the vertical force is restricted to act only in the upward direction (i.e., the user cannot pull up on the walker); the horizontal force is restricted by the coefficient of friction between the walker and floor (assumed to be 0.3) and the vertical force; and the horizontal force and shoulder moment are additionally constrained by the structural stability (i.e., tipping point) of the walker.

**Table 1, Simulation parameters.**

<table>
<thead>
<tr>
<th>Segment</th>
<th>Relative Length</th>
<th>Distance to Mass Center</th>
<th>Relative Mass</th>
<th>Relative Inertia</th>
</tr>
</thead>
<tbody>
<tr>
<td>Upper Body</td>
<td>0.29</td>
<td>0.64</td>
<td>0.65</td>
<td>0.50</td>
</tr>
<tr>
<td>Thigh</td>
<td>0.23</td>
<td>0.43</td>
<td>0.11</td>
<td>0.32</td>
</tr>
<tr>
<td>Shank</td>
<td>0.22</td>
<td>0.40</td>
<td>0.05</td>
<td>0.50</td>
</tr>
<tr>
<td>Foot</td>
<td>0.13</td>
<td>0.25</td>
<td>0.02</td>
<td>0.48</td>
</tr>
</tbody>
</table>

* L = High, M= Mass of subject

** The distance is a fraction of the segment longitude

All units are in the MKS system

B. Orthosis Model

Since the JCO utilizes an AFO at the ankles, ankle inversion/eversion is not considered, while flexion/extension torques are modeled as a torsional spring and damper about zero degrees of flexion. Thus, the ankle is not a “controlled” degree of freedom. The knee torques result from a combination of quadriceps stimulation, joint coupling, the knee joint spring, and the resistive torque of the brake. These are combined as described in the gait controller below. The hip torques result from the joint coupling and the hip brakes. Specifically, the hip joint coupling is modeled as a flexor torque that has the function to maintain the hip angle equal to the knee angle (in the case of a knee-angle/hip-angle coupling ratio of 1:1), although only in extension. In the model, the abduction-adduction movement is locked in the zero position. Note that the orthosis allows hip abduction/adduction with a limit stop on excessive adduction (to prevent crossing over). This degree of freedom was not modeled in the simulation, since frontal plane dynamics are assumed to be controlled by the user via the
walker, and thus are represented as a simplified single degree-of-freedom (i.e., rotation in the frontal plane).

C. Control Algorithm

In the proposed system, two controllers are simultaneously active, which are the JCO controller (i.e., brakes and electrical stimulation) and the user controller, which governs interaction with the walker through the arms (i.e., shoulder forces and torques). The control algorithm is a loop consisting of four states (Fig. 6):

D. S1 : State 1

In the initial condition, both knee brakes are locked in zero degrees of flexion. The support leg is in front of the swing leg. The upper body can rotate freely at the hips. The user rotates the upper body forward to an angle of 20 degrees, at which point the support hip brake is locked. At that moment the center of mass is in front of the support foot, which causes forward motion and a rotation around the front foot. The controller switches to the next state when the support leg is vertical and the user has raised the torso using the walker.

E. S2 : State 2

The swing knee brake is unlocked; this allows the spring to flex the knee, and the hip is concurrently flexed due to the joint coupling. During this state, the user maintains control of upper body orientation to approximately vertical. The stance hip brake allows hip extension (and not flexion) to zero degrees (at which point it will lock), while the stance knee brake remains locked. The controller switches to the next state when the difference between hip angles ceases to increase.

F. S3 : State 3

The swing hip is locked while the swing knee is extended via quadriceps stimulation. The stance knee remains locked while the stance hip does not allow extension beyond zero degrees. The controller switches to the next state when the swing knee is fully extended.

G. S4 : State 4

The swing knee is locked and the swing foot lands due to gravity and user arm control. The stance hip is free. The swing hip allows flexion such that the upper body leans forward 20 degrees from the vertical (as in state 1). The swing leg is now in front and becomes the stance leg; as such the system is returned to state 1 and ready for another step.

H. Simulation Results

A simulation of the JCO and gait controller was conducted using the parameters given in Table 1, for a user of height L=1.7m and mass M=65kg. Figure 7 shows the results of fifteen seconds of simulation, including position and joint angle data and ground reaction force data. A video of the simulated walk is included in the supporting material. Note that the user starts from rest, and remains at rest for the first second of the simulation. The cadence of the resulting gait was 34 steps per minute and the average velocity was 0.2 m/s. The upper body never leans backward and the maximum forward inclination is around 25 degrees. The simulation further indicates that nearly all the weight is carried by the legs, and thus the approach results in minimal weight bearing on the arms (i.e., the arms are used chiefly for stabilization and not for support purposes). Finally, Fig. 8 shows the evolution of states through the simulation. As
indicated in the figure, the duration of electrical stimulation (stage 3), is small compared with total cycle. The ratio is 0.1:3.5 seconds, which equates to a quadriceps duty cycle of approximately 3%.

IV. PRELIMINARY EXPERIMENTS

The authors have conducted preliminary experiments to test the joint coupling concept, and to assess the extent of fatigue imposed by the bias spring and joint coupling. A simple, one-legged version of the orthosis was created which included one-to-one joint coupling between the hip and knee, an adjustable extension spring for knee flexion, potentiometers on the hip and knee joints for angle measurement, and a locking knee joint with quick release pin. Experiments were conducted to 1) determine if the spring and joint coupling could provide sufficient knee and hip flexion in the context of stride, and 2) determine if the quadriceps could repeatably provide the power necessary to overcome the spring and extend the lower leg without significant fatigue. The first experiment (see Fig. 9) involved positioning an able-bodied subject in stance such that the leg wearing the orthosis was in the rear and ready to begin stride. The spring was loaded and the knee was locked in the extended position by means of a quick connect pin. As the pin was pulled by an assistant, the leg swung forward in knee and hip flexion as shown in Figs. 9 and 10. At the peak of hip flexion, the quadriceps was stimulated, which fully extended the knee and—in this test, due to the absence of a hip brake—allowed the hip to extend as well, dropping the foot to the ground. The results of this experiment showed that joint angles comparable to those typical in healthy subjects could be obtained involuntarily with the JCO system (see Fig. 10).

The second experiment involved conducting extended sets of pulsed quadriceps stimulation at a duty cycle of 15%—conservative according to simulations. Ten subjects underwent three five minute periods of stimulation with a rest period in between each trial of three minutes. Results from this preliminary fatigue testing showed that muscle output decreased 7% over five minutes for all the trials combined and averaged. As such, the quadriceps appears to be quite capable of providing sustained power for the proposed hybrid approach without significant degradation of performance.

V. CONCLUSION

A joint-coupled controlled brake orthosis (JCO) has been designed as part of a hybrid FES/orthosis system for restoring gait to spinal cord injured individuals. This device will 1) unidirectionally couple hip to knee flexion; 2) aid hip and knee flexion with a spring assist; and 3) incorporate sensors and modulated friction brakes, which are used in conjunction with electrical stimulation for the feedback control of joint (and therefore limb) trajectories. Dynamic simulations and a one-legged prototype of the orthosis were used to validate the design concepts and aid in the design development. The results of these efforts showed that robust walking can be achieved via the orthosis without significant muscle fatigue. Future work includes characterization of the latest brake prototype, development of a fully functional, two-legged JCO, control design and the addition of on-board electronics, and clinical trials of the JCO system.

Fig. 9. FES/JCO generated gait sequence experiment.

Frame 1: Right leg is locked in shown position. Assistant pulls pin to unlock the knee joint. Frames 2-5: Once pin is pulled, the spring pulls the knee into flexion and the joint coupling therefore pulls the hip into flexion as well. Frame 5 is the final resting state under no muscle contraction. Frames 6-8: The quadriceps is stimulated by a momentary push button switch on the walker handle. This causes the knee to extend and therefore relieve the joint coupling, allowing the hip to extend also. Frame 8 is the final resting position of the leg after it has rejoined the ground after stride.

Fig. 10. Joint angle data during gait experiment shown in Fig. 9.
REFERENCES


