Design of a Controlled-Brake Orthosis for FES-Aided Gait

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Abstract—Functional electrical stimulation (FES) is a means of restoring gait to individuals with spinal cord injury, but the performance of most FES-aided gait systems is hampered by the rapid muscle fatigue which results from stimulated muscle contraction and the inadequate control of joint torques necessary to produce desired limb trajectories. The controlled-brake orthosis (CBO) addresses these limitations by utilizing FES in combination with a long-leg brace that contains controllable friction brakes at the knees and hips. A laboratory version of the CBO utilizing computer-controlled magnetic particle brakes at the joints was designed and constructed, and preliminary results with a single spinal cord injury (SCI) subject have demonstrated reduced fatigue and more repeatable gait trajectories when compared to FES-aided gait without the brace. Significant work remains to demonstrate the efficacy of the concept across a wide range of SCI subjects and to design a system which meets appropriate user requirements of size, weight, cosmesis, ease of use and cost. The primary purpose of the paper is to detail the design of the CBO.

I. INTRODUCTION

One of the most disabling consequences of spinal cord injury (SCI) is the loss of mobility resulting from lower limb paralysis. Functional electrical stimulation (FES) is a means of restoring limited mobility to some individuals with SCI by using electrical stimulation of motor nerves to elicit muscular contractions [1]–[5]. Two fundamental problems severely restrict the ability of current FES systems to restore gait. The first is the inadequate control of joint torques necessary to produce desired limb trajectories, and the second is the rapid muscle fatigue which results from stimulated muscle contractions. Rapid muscle fatigue limits the standing time and walking distance of the SCI individual, and poor movement control results in abnormal gait trajectories with large step-to-step variations.

FES can be combined with a lower limb orthosis to resolve some of the control and fatigue problems. Several groups are working with hybrid systems for FES-aided gait, including combining stimulation with the Louisiana State University Reciprocating Gait Orthosis (RGO) [5]–[12], the Oswestry Parawalker system [13], [14], the hip guidance orthosis (HGO) [15], [16], the enhanced ankle-foot orthosis (AFO) system [17], [18] and the modular Hybrid Assistive System (HAS) [19]–[21]. All are designed to augment the limited capability of FES. For example, the RGO locks the knees to provide support and reciprocally couples the hip joints so that hip flexion can occur by extension of the contralateral hip. Because the knees are locked, RGO gait has considerable lateral motion. The enhanced AFO system is somewhat similar to combining stimulation and the Vannini-Rizzoli boot [22], [23] where the AFO fixes the ankle in a slightly dorsiflexed position to place the body in a narrow equilibrium range where both hip and knee joints can maintain standing loads without muscle stimulation. Small perturbations about this point require the muscles to be turned on to prevent collapse. The HAS uses a brace with joints containing both brake and motor where the brake can lock the joint during stance and the motor can provide energy during swing.

Orthoses reduce the complexity of gait by imposing kinematic constraints and enabling some degree of closed-loop control for electrical stimulation by providing a mounting platform for sensors. An orthosis can also help to reduce fatigue by mechanically locking the knee joint, thus reducing the duty cycle of the stimulated muscle. However, entirely passive hybrid systems are not able to improve the control over joint torques, and systems which lock the knee joint during swing result in a stiff-legged gait with poor stability [22]–[26].

Some research has been conducted on powered orthosis which contain dc electric motors at one or more joints and are designed for use either with [21], [27] or without [28], [29] supplemental electrical stimulation. Because the stimulated muscle power can be augmented with the motors, these systems need not rely solely on the stimulated muscle to provide the joint torques for gait and thus have the potential of being used by a larger set of the SCI population than can use unpowered devices. Due to the size and weight of current dc motors and their battery energy storage sources, self-contained, fully powered orthosis will not be feasible in the near future.

The system described in this paper addresses the limitations of FES-aided gait by utilizing FES in combination with a controllable passive (or "semi-active") orthosis [30], [31]. The controlled-brake orthosis (CBO) is a long-leg brace that contains controllable friction brakes at the knees and hips. The system achieves desired limb trajectories by utilizing the stimulated muscles as a source of coercely regulated power and regulating the power at each joint through computer control of the friction brakes. In contrast to uncontrolled passive orthoses such as the AFO and RGO, the CBO enables direct dynamic control of limb trajectories including the unlocked
knee motion which characterizes the swing-through kinematics of able-bodied gait [31]. The system reduces the effect of muscle fatigue by locking the controllable brakes at hip and knee during stance enabling a wide stability region with no muscle stimulation because the brace is rigid. Thus, with the CBO system, muscles are utilized only to provide limb motion, which greatly reduces the duty cycle of stimulation and diminishes the problem of muscle fatigue. The orthosis provides the requisite isometric joint torques for stance, but relies on the skeleton to support axial loads, and so need not be designed to withstand large compressive loads. Unlike the actuators on fully powered orthosis, the controlled brakes are small, lightweight, and energy efficient. Thus, a viable self-contained CBO is considerably more likely than a fully powered orthosis. Further, since the energy for gait is provided by the stimulated muscles, the SCI individual derives some of the physiological benefits of exercise [32]–[39].

The CBO concept does have limitations. The most important is that the user must don a long leg brace which covers both hip and knee joints. Despite the potential for improved gait performance, the orthosis may not be acceptable to many users. A second limitation is the need to provide energy for the controllable brakes. Practical versions of the CBO will need brakes with sufficiently low power requirements to enable an hour or more of use on a single battery charge.

The purpose of this paper is to present a detailed design description of the CBO hardware and control strategies. Preliminary results from use on a single SCI subject are presented to demonstrate that the hardware works as intended and that the CBO concept has sufficient validity to merit further development.

II. DESIGN REQUIREMENTS

The version of the CBO described here is being used strictly as a research tool and was designed to meet requirements of functionality, comfort, and sufficient adjustability to fit a wide range of body sizes. Many important product design issues were not and will not be addressed until laboratory evaluation of the CBO system warrants further product development. In particular, a product version of the CBO will require on-board battery energy sources and control computers and must meet additional design requirements of size, weight, ease of donning and doffing, and cosmesis.

III. A. Orthosis Degrees of Freedom

The controlled-brake orthosis accommodates four skeletal degrees of freedom: flexion and extension of the hip, knee, and ankle joints and abduction and adduction of both hip joints. Of the four degrees of freedom, brakes are used to control the two most important to gait: hip flex/extension and knee flex/extension. The hip ab/adduction degree of freedom was left free to achieve static equilibrium in the frontal plane when in single stance phase. However, by implementing an adjustable limit stop on hip adduction, the brace prevents the crossing of one leg in front of the other (scissoring) that often occurs during FES-aided gait.

B. Joint Kinematics

It is well known that the flexion and extension of the human knee joint consists not only of sagittal plane rotation, but also sagittal plane sliding and horizontal plane rotation [40], [41]. Although reasonably effective polycentric joint designs have been developed to accommodate these additional components of knee motion [42]–[45], incorporation of a polycentric joint in the CBO would complicate the design of a joint containing a brake. Further, polycentric joints have several degrees of freedom creating challenges for the sensing of joint position and torque that is required for control. Since a pin joint is a reasonably good approximation of knee motion [44], [46], [47], the CBO knee joints were designed as revolutes which not only simplifies control, but also reduces the size, weight, and cost of each joint.

The human hip is well approximated as a ball-and-socket joint. Hip flexion and extension can therefore be accommodated with a simple revolute joint which results in a design that is well characterized and controlled with simple measurements. Hip ab/adduction is also well characterized by a revolute joint, but its axis of rotation passes through the hip which is located internal to the pelvis. Since the hip joint of the orthosis is positioned on the lateral aspect of the pelvis, positioning a revolute joint “through” the hip would require a large gimbled joint, which would be both heavy and cumbersome. We addressed this difficulty by locating a multiple degree of freedom linkage off-axis that allows for skeletal hip ab/adduction. Since the joint is not controlled and thus need not impose external torques, a multiple degree of freedom solution does not complicate the orthosis, and in fact provides for a more compact design.

C. Joint Range of Motion

The ranges of motion for the joints were selected to accommodate normal gait and comfortable sitting. The hip joint allows 105° of flexion and 25° of extension, and in the frontal plane, accommodates 15° of abduction and 5° of adduction, with an adjustable adduction lock to prevent scissoring. The knee joint allows 105° of flexion and 0° of extension. The ankle plantar/dorsiflexion degree of freedom must impose a stiffness sufficient to prevent foot drop during the swing phase of gait, yet be compliant enough to allow dorsiflexion during the stance phase.

D. Maximum Brake Torques

The brace exerts modulated braking torques on the flex/extension degree of freedom of the hip and knee joints. We derived maximum joint torque specifications for the research version of the CBO based on three desired capabilities for the orthosis: 1) be able to provide the dissipative torques observed in normal gait, 2) be capable of locking a joint against a stimulated muscle contraction, and 3) be able to implement a controlled stand-to-sit maneuver without the aid of muscle stimulation. Calculations based on average-sized persons indicated that 50 Nm of knee torque and 30 Nm of hip torque would provide these capabilities. An important objective of the work with the research CBO is to establish
more specific, empirically-based design requirements for a practical version of the system.

E. Minimizing Parasitic Losses

Since stimulated muscles have significant power and energy limitations, and because the brace is assisting rather than replacing the existing limbs, minimizing orthosis weight, rotational inertia, and residual friction is paramount. Unlike an actively powered device, the added joint rotational inertia and residual friction of the brace cannot be negated by torque feedback from the actuators. Any increase in limb weight or rotational inertia, or any added joint friction, requires more muscle force for a given acceleration and more energy for a given motion. Intelligent distribution of the mechanical components on the brace minimized the effects of the added weight and inertia. For example, since the pelvis is not subjected to significant accelerations nor to large vertical displacements, it was advantageous to locate as much of the orthosis weight as possible above the hip joint.

IV. Design Description

Figs. 1 and 2 show the laboratory version of the CBO. The total orthosis weight configured for an averaged sized person is 6 kg (13 lbs) which is comparable to commercially available long leg braces. Because of careful component locations, the rotational inertia of the distal link of the orthosis about the knee joint is about 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.

A. Structure

The orthosis structure is fabricated from 2024-T3 aluminum alloy and 4130 (chromium-molybdenum) steel alloy, both chosen for their high strength-to-weight ratios. The aluminum alloy was utilized where a higher strength-to-weight was most important, and the steel alloy where greater strength-to-size was desired. The two alloys have roughly equal stiffness-to-weight ratios.

The orthosis utilizes unilateral uprights because bilateral uprights would obstruct gait by adding bulk to the medial aspect of the legs. Since the joint torques are applied only to the lateral upright, having medial uprights would also not offer any significant stiffness benefits. The brace links, which extend the length of each limb segment and connect the orthosis joints, were designed as tubular I-beams, a configuration which maximizes the orthosis stiffness and strength in the sagittal plane where most of the major bending stresses are generated. The I-beams are constructed from 4130 steel tubes. Chromoly steel was chosen over carbon composite tubes for safety since the chromoly exhibits ductile failure while carbon fiber has brittle failure and could present sharp edges if broken during a fall. The support beam tubes also function as convenient conduits for routing wires between joints. The orthosis design allows for small variations in the size of subject limbs and larger variations are achieved by interchanging the links which
enables fitting the brace to a wide range of male and female subjects.

B. Attachment Points

Applying controlled resistive torques to the musculoskeletal system requires a rigid coupling between the CBO and the body. The CBO applies resistive joint torques to the lateral side of the limb only. As illustrated in Fig. 3, this configuration leads to a lateral moment arm between the plane in which skeletal torques are applied and the plane of CBO joint torques. Brake torques acting across this moment arm subject the CBO to structural deformation, and any lateral compliance in the brace/body attachment results in the brace twisting around the leg. The structural deformation is reduced by minimizing the forces acting on the attachments and by making these attachments as rigid as possible. Since the CBO joint torques are transmitted to the limb segments through the attachment points by force couples, the attachment forces are decreased by increasing the spacing between them. Maximizing this spacing decreases the structural deformation of the CBO, effectively creating a more rigid coupling between the brace and body.

Given these considerations, the orthosis is attached to each leg segment with stiff cuffs fabricated from bent chromoly tubing and located at the proximal and distal ends of the respective segments. Velcro straps are used to tighten the leg against the cuffs. The orthosis is attached to the distal end of each shank by a standard, compliant AFO worn inside a conventional shoe which provides the appropriate stiffness to achieve the desired ankle motions.

Coupling the orthosis to the pelvis was more difficult. The only viable option was to “C-clamp” onto the iliac crests, since all other part of the pelvis lie deep in the body. This attachment is augmented by Velcro straps.

The brace/body attachment points are shown schematically in the diagram of Fig. 3, and on the orthosis in the photograph of Fig. 4. Tests with several subjects have shown they provide sufficient rigidity and support.

C. Controllable Brakes

The means for applying controlled braking loads is of great importance to the design of the CBO system. Considerations for selecting a brake technology included peak resistive torque, control bandwidth, size, weight and residual friction. The practical version of the CBO will have further requirements since it must be lightweight, battery-powered, and self-contained, and must minimize electrical power consumption.

We considered three technologies for controlled braking: hydraulic cylinders with variable orifices, dc motors and magnetic particle brakes. After a thorough investigation, we determined that magnetic particle brakes would provide the best performance for the laboratory version of the CBO. In this section, all three alternative braking technologies are described, along with their strengths and weaknesses.

1) Hydraulic Cylinders: Fig. 5 illustrates a hydraulic cylinder configured for use as a controlled brake. Hydraulic systems offer the great advantage of being able to support large static loads without any electrical power consumption once the flow valve is off. Controllable hydraulic brakes, however, generate a
TABLE I
COMPARISON BETWEEN DC MOTORS AND MAGNETIC PARTICLE BRAKES FOR PROVIDING CONTROLLED BRAKING LOADS

<table>
<thead>
<tr>
<th>Property</th>
<th>DC Motors</th>
<th>Magnetic Particle Brakes</th>
</tr>
</thead>
<tbody>
<tr>
<td>Holding Torque/Weight</td>
<td>0.15 Nm/kg</td>
<td>8.0 Nm/kg</td>
</tr>
<tr>
<td>Holding Torque/Size</td>
<td>About same as Torque/Weight (assuming equal density)</td>
<td></td>
</tr>
<tr>
<td>Holding Torque/Electrical Power</td>
<td>0.0060 Nm/watt</td>
<td>0.4 Nm/watt</td>
</tr>
</tbody>
</table>

Fig. 5. Concept for a controlled brake based on a hydraulic cylinder. By modulating the valve, resistance to motion of the cylinder piston can be controlled. A single-rod cylinder is shown for diagrammatic simplicity, although as described in the text, a double-rod cylinder is necessary if incompressible fluid is used.

host of design and implementation difficulties. The maximum joint torque and range of motion required result in large, heavy cylinders. Further, to maintain high stiffness the fluid must be incompressible and therefore the cylinder must be of the double rod type, which requires a linear space almost three times the length of the body of the cylinder. Hydraulic cylinders also exhibit considerable stiction from the high pressure fluid seals around the piston and piston rods. Motion of the hydraulic fluid through the circuit imposes additional friction and inertia, both of which will be increased by the rotary-to-linear transmission, since the inertia and friction of the hydraulic system will be multiplied by the square of the transmission ratio when reflected to the joint space.

Yet another difficulty in the development of a hydraulic system is the design and fabrication of a small, electrically controlled proportional hydraulic valve. Such a valve must have large bandwidth as well as the ability to operate under the extreme forces imposed by high pressure fluid. The combination of these capabilities favors the use of a high-power servovalve but most servovalves have substantial size and weight. Since the fluid is incompressible, the controllable valve cannot introduce volume to the system which excludes needle and other similar types of proportional flow valves.

Hydraulic systems would complicate the control problem, since the applied joint torque would not be independently controllable, but rather would be a function of the joint velocity as dictated by the behavior of fluid flow through an orifice. Imposing a constant resistive torque at the joint would necessitate measurement of joint torque and joint velocity, and real-time adjustment of the controlled valve.

2) DC Motors: DC motors offer the advantage of simple control over joint torque, high bandwidth, and the ability to add power to the system if desired. The relatively low torque output of motors, however, would necessitate a large, high ratio mechanical transmission. Since the inertia and residual friction of the motor would be multiplied by the square of the transmission ratio when reflected to the joint, a transmission of the size required would likely result in excess parasitic joint inertia and friction. These could be eliminated using torque feedback, but implementation of this type of control through a large transmission can result in instability.

Motors are also relatively inefficient in their conversion of electrical power to static holding torque, which would make a battery-powered CBO infeasible because of the time spent in single and double support phase with the joints locked by the brake. Finally, motors are active power sources which add energy to the system, and safety becomes an issue because the motors can interact with human limbs.

3) Magnetic Particle Brakes: Magnetic particle brakes offer the advantage of simple control of joint torque over a bandwidth suitable for controlling the events involved in human gait. DC current applied to the coil of the brake induces a magnetic field which links fine ferrite particles to the rotating brake shaft. The amount of current determines the strength of the magnetic field, which in turn determines the resistive torque imposed on the brake shaft. The relation between torque and input current is linear and the engagement smooth and quiet. These devices provide a relatively compact and low weight means of exerting dissipative mechanical torques using an electrical control signal.

Unlike dc motors, particle brakes are passive devices with no possibility of human injury resulting from unstable behavior, and thus are well suited for applications involving human interaction. Additionally, magnetic particle brakes offer a resistive torque to weight ratio more than an order of magnitude greater than a typical high performance dc torque motor (Table I). A particle brake of roughly the same size and weight as a dc motor requires a transmission ratio about one tenth that of the motor, which not only reduces the size and weight of the transmission, but also greatly reduces the inertia and friction (which scale with the square of transmission ratio) that is reflected from the brake to the joint. Finally, particle brakes consume less electrical power than comparable dc torque motors with the holding torque to power ratio of a magnetic particle brake at least 50 times greater than that for a high performance dc torque motor (Table I). For these reasons, magnetic particle brakes were chosen as the controlled brakes for the CBO.

Particle brakes with custom windings were used (Force Limited, Santa Monica, CA). The brakes for the knee joint (Model B20SF15) weigh 335 g, are 51.7 mm in diameter, 27.9 mm long (excluding the shaft) and can resist a continuous maximum torque of 2.8 Nm. The brakes for the hip joint (Model B20SF14) weigh 235 g, are 48.7 mm in diameter, 26.0 mm long and can resist 1.8 Nm.
The primary dynamic behavior associated with a magnetic particle brake involves first-order electrical energy domain dynamics of the brake inductance and second-order mechanical energy domain dynamics of the ferrite particle motion both of which are characterized by time constants faster than those seen in human gait. The brakes can therefore be controlled through a standard voltage-to-current servoamplifier which simplifies torque control. The servoamp power supply rails are at $+54$ V and $-28$ V to increase the bandwidth of the controller. The purpose of the negative rail is enable fast turn-off of the brakes.

In steady state, when the hip brakes are resisting 1.8 Nm, they consume 350 mA at 10 V (3.5 W). When the knee brakes are resisting 2.8 Nm, they consume 350 mA at 28 V (9.8 W).

D. Transmission

Magnetic particle brakes work best in high angular velocity, low torque applications; however, human gait consists primarily of low angular velocity, high torque motions. Using the full mechanical power range of the brakes requires a transmission with at least a 16:1 speed reduction from the brake to the joint to provide the specified maximum joint torques from the selected particle brakes. Since power flow is from the joint to the brakes, the transmission must be easily back-driveable; that is the joint can be driven to move the brake. Further, to minimize parasitic dynamics, the transmission should be stiff, light weight, and compact.

We selected an Evoloid gear set (ASI Technologies, Horsham PA) for the transmission. Evoloid gears utilize a geometric variation on the conventional involute-tooth helical gear profile that allows the use of a four tooth-pinion without the associated undercut exhibited by conventional gearing. These gears provide a compact, back-driveable, high ratio transmission in a single stage that is capable of carrying large static loads. Evoloids are much smaller and lighter than comparable single stage spur or helical transmissions and, unlike hypoid and Spiroid gears, offer parallel rather than perpendicular shaft arrangement. When compared to multiple stage gear trains or a planetary gear set, the single stage Evoloids only require the housing and bearings necessary to support two shafts. An Evoloid set is also far stiffer than a cable drive transmission.

The gears were machined from 4140 (chromium-molybdenum) steel alloy, and the matching pinions from 6100 series (chromium-vanadium) alloy. Both the hip and knee joint transmissions have a 16:1 ratio.

E. Joints

As shown in Fig. 6, each orthosis joint consists of a magnetic particle brake, an Evoloid gear set, and a gear housing. The gear housing is fabricated from two aluminum plates separated by the gear shaft and two smaller structural shafts. Mounting the Evoloid gear onto its shaft with a bearing results in a rigid connection between the housing plates and the shaft, and the shaft functions as the primary structural component of the housing. Because the gear is mounted directly on the bearing, there is less lateral stiffness than fixing the gear to the shaft and supporting each end of the shaft on bearings. The single bearing configuration, however, offers the advantage of a smaller and lighter gear housing.

The mechanical transmission is among the heaviest of the brace components so we attempted to minimize its size and weight. The Evoloid gears were cut so that only a webbed section necessary for engagement over the total joint range of motion remained, reducing the size and weight of the components. The combined weight of the spiked gear and pinion is 115 g for the hip and 182 g for the knee joint.

The linkage designed for the hip joint ab/adduction motion (Figs. 7 and 8) is a three degree-of-freedom mechanism that utilizes revolute joints to accommodate the skeletal degree of freedom. The limit stop which blocks the motion of the
middle joint of the linkage results in a structurally over-determined brace/skeletal system, which provides the adduction lock necessary to prevent scissoring during gait.

F. Sensors

The hip and knee joints are instrumented with angle and torque sensors. Potentiometers (CP-2FK, Midori America, Corona, CA) to measure joint position are located inside the tubular gear shafts, and joint velocity is derived by off-board analog differentiation of the position signal. Strain gages (EA-13-060PB-350, Micro-Measurements, Raleigh NC) are fixed to a load bearing member of the brace to measure joint torque. The electronic circuitry for strain gage and potentiometer signal conditioning is also located inside the gear shaft to save space and minimize signal noise.

V. ORTHOSIS CONTROL

A. Control Strategy

One of the primary functions of the CBO is to improve the trajectory control of the lower limbs during gait. A typical control system affects control commands through an actuator which can both add and remove power to and from the plant. The CBO, however, effects control commands through a dissipator, and can therefore only control the amount of power removed from the plant. Thus, the control strategy used for the CBO differs from that of a typical control system containing active actuators.

Fig. 9 depicts the modulated dissipation control strategy for a single joint. The controller is closed loop and modulates the brake to regulate the desired position and velocity of the joint during gait. If the muscle torque is sufficient to provide the desired accelerations, this strategy provides an effective means for controlling limb motion. Control performance, however, is affected by the rigidity of the coupling between the brace joint and the skeletal joint and the bandwidth of the modulated dissipator, because a poor coupling or slow dissipator will introduce dynamics in the feedback loop that will restrict the controller bandwidth and severely limit performance.

An advantage of the controller is that it does not require a mathematical model of the muscle or system. The controller treats the muscle as an unregulated power source and utilizes control of the brake to achieve the desired joint trajectories. The only requirement for the muscle is that it provide sufficient power to make the desired motion possible. All other behavior and variation due to gravity, inertial coupling, spasticity, and fatigue is treated as a disturbance which the controller is designed to reject.

Because control over the joints is effected solely by the magnetic particle brakes, and because these brakes can only dissipate energy, there is no possibility of unstable mechanical behavior. In the worse case, there will simply be poor tracking performance. However, we have found through both simulation and experiment that the electrical subsystem of the magnetic particle brakes can be driven into dynamic instability which results in chatter at the output of the brake and poor tracking at the limb. This has motivated an ongoing effort to develop a detailed mathematical model of magnetic particle brakes and to create improved controllers.

B. Joint Angle Trajectory Control

A block diagram of the controlled-brake trajectory controller is shown in Fig. 10. In the figure, $\theta$ is joint angle, $\dot{\theta}$ is joint angular velocity, $\epsilon$ is the tracking error (a weighted sum of position and velocity errors), $T_c$ the control command, $T_b$ the brake torque, $T_m$ the muscle torque, and $T$ the net torque acting on the joint. The controller is tuned by adjusting the weights $\mu$ and $\gamma$. The controller is somewhat analogous to a traditional PD controller, except that the “actuator” cannot add energy to the system, a difference which is represented by the rectifier in Fig. 10. When the stimulated muscle causes the limb to overshoot the desired trajectory (a positive tracking error in Fig. 10), the controller responds by removing energy from the moving limb, so that the desired trajectory can catch up. Normal PD control action occurs only in this region. When the limb has fallen behind the desired trajectory (a negative tracking error), the best the controller can do is shut off completely (represented by the rectifier), allowing the stimulated muscle torque to accelerate the limb back toward the desired path.

Using both position and velocity error information yields smoother and more efficient control. For example, when the limb is behind the desired position but moving faster than the desired velocity, the velocity error information enables the controller to start dissipating energy before the limb reaches the desired position trajectory, slowing it down so it joins the desired path smoothly.

Due to the nonlinear nature of the controller, the sign conventions associated with the tracking error depend on
whether the joint is in flexion or extension. Fig. 10 shows the proper sign convention for flexion, given that $\theta$ and $\omega$ are positive in this direction. The tracking error during flexion is thus represented as

$$e = -\mu(\dot{\theta}_d - \theta) - \gamma(\omega_d - \omega).$$

Since the direction of muscle torque changes in extension, the sign of the tracking error must be reversed (positive feedback in the block diagram), so that

$$e = \mu(\dot{\theta}_d - \theta) + \gamma(\omega_d - \omega).$$

The desired swing phase joint trajectories were generated from piecewise sinusoidal fits to “typical” trajectories for slow speed able-bodied gait. It should be noted that the control structure permits tracking of any reasonable desired trajectory and the choice of matching able-bodied gait was more for convenience than the result of an optimization process.

C. Muscle Stimulation Level Control

Although the controlled-brake system obviates the need for fine regulation of muscle stimulation, some level of stimulation control is important for effective gait. If stimulation levels were too high, there would be unnecessary power dissipation in the brakes and excessive muscle fatigue. If too low, the muscle would not produce sufficient torque to provide the acceleration necessary to follow the desired trajectory. The appropriate level of stimulation depends to a large extent on the state of muscle fatigue which is an unknown function of time and other variables. The purpose of the stimulation level controller is to adjust the level of stimulation during gait so that the muscle can provide adequate limb acceleration without excessive fatigue.

A block diagram of the stimulation controller is shown in Fig. 11. The controller receives feedback from the position tracking error $\theta_e$ and the torque command error $T_e$. The former is used to detect when the stimulation level is too low and the latter to detect when the level is too high.

The performance of the joint angle trajectory controller is much more sensitive to errors caused by the stimulation level being too low than it is to errors of excessive stimulation. When the stimulation level is too high, the system can maintain the desired trajectory by control of the brakes. When the stimulation level is too low, the best the trajectory control system can do is turn off completely and let the muscle torque accelerate the limb toward the desired trajectory.

For the error region in which the stimulation level is too low, the position tracking error is half-wave rectified to produce a negative error signal $\theta_{ne}$. This error is averaged (the integration block in Fig. 11) and discretized on a step-by-step basis because it is ineffective to regulate stimulation within a single swing phase. Corrections to this error occur on the next step.

Maintaining the system in the region where the brake can exert its control requires that the brake maintain a nonzero torque for the duration of motion. The difference between the torque command set point $T_d$ (which is fixed at a small positive value) and the actual torque command $T_e$ is the torque command error $T_e$. A negative value of $T_e$ indicates that the brake torques are high which means the muscles are working harder than needed and the stimulation is too strong. The positive value of $T_e$ saturates at $T_d$ (since $T_e$ cannot be
negative), thus this section of the controller cannot compensate for stimulation levels that are too low. The torque error term is also averaged and discretized step-by-step.

The total stimulation level error $\lambda$ is a weighted difference of the position and torque error terms as given by

$$\lambda = \epsilon \theta_{nc} - \beta T_e.$$  

One limitation of the stimulation controller is that it cannot detect disturbances. A foot scraping the ground or one leg rubbing against the other during swing phase cannot be differentiated from a state of inadequate muscle torque. The disturbance adds an external torque about the joint which results in an increase in negative position error and a corresponding decrease in brake command. Inadequate muscle torque due to insufficient muscle stimulation would provide a similar error condition. Since disturbances occur relatively infrequently and at arbitrary intervals, their effect can be diminished by low-pass filtering the stimulation level error to determine the desired change in stimulation level. The discrete-time low pass filter $H(z)$ weights the error terms from $n$ steps as described by

$$H(z) = \frac{\Delta I(z)}{\lambda(z)} = A\left[\frac{nz^n + (n-1)z^{n-1} + (n-2)z^{n-2} + \cdots + z}{0.5n(n+1)z^n}\right],$$

which represents the difference equation

$$\Delta i_k = \left[A\left[\frac{1}{0.5n(n+1)}\right]\right]n\lambda_k + (n-1)\lambda_{k-1} + \cdots + (n-k)\lambda_{k-n+1}.$$  

The full control system which combines joint angle trajectory control and muscle stimulation level control is shown in Fig. 12.
VI. PRELIMINARY HUMAN SUBJECT TESTING

Clinical evaluation of the laboratory-based CBO using SCI subjects is underway. In these tests, performance of the four channel FES-aided gait with the CBO is compared to conventional four channel FES-aided gait without the CBO. Characterization of muscle fatigue and energy expenditure for the two methods is based on measuring: 1) total walking distance, 2) total standing time, 3) stimulation duty cycle, 4) isometric recruitment curves (IRC) of the quadriceps, 5) step trajectory control, 6) heart rate and blood pressure, and 7) metabolic energy requirements. The IRC is taken at regular intervals during the gait session and represents the nonlinear static gain between stimulation level input and muscle force output. It is measured using the sensors on the CBO and the cross-correlation techniques developed by Beck and Durfee [48], [49]. Step trajectory control is characterized primarily by step-to-step variations in joint angle trajectory and stride length.

For the laboratory CBO, the subject issues the vocal commands of stand, sit, right step, or left step to the operator who implements them using the keyboard of the controlling computer. A practical system could use any of the autonomous subject command paradigms seen in existing systems, with the most common being finger and thumb switches.

Initial tests to verify equipment reliability and system operation were conducted on a single subject who is a T6-complete paraplegic and who walked between parallel bars [50]. A subset of the fatigue and energy measures listed above were measured for this subject. Preliminary results indicated improvement in step trajectory control and reduction in muscle fatigue when compared with conventional four channel FES-aided gait. Also, the hip adduction lock prevented scissoring during gait.

Fig. 13 shows a typical stimulation record for this subject for several steps taken with conventional four channel FES-aided gait compared to similar steps taken with the CBO. For FES gait, each quadriceps muscle group was stimulated for approximately 85% of the time, while for CBO gait the muscles were stimulated for about 10% of the time because they were active only during swing. This greatly reduced muscle fatigue and enabled the subject to walk further when using the CBO. The pace of this subject was slow (approximately 0.12 m/s) because at that time, we were not attempting to optimize speed.

For the same subject, Fig. 14 illustrates the variation in knee trajectories over 50 strides with conventional FES-aided gait compared to trajectories when using the CBO. Significant reductions can be seen in stride-to-stride variation when using the CBO. Continued testing indicated that the 10 steps with the largest flexion angles shown on the left side of Fig. 14 were unusual for this subject whose typical knee flexion ranged from 40 to 80°. However, the data shown does verify the need to control the reflex flexion response which can demonstrate considerable step-to-step variation.
For these tests, both hip and knee trajectories were controlled during swing and locked during stance. Swing phase hip control was less successful than knee control because the hip flexors were activated indirectly through the withdrawal reflex and the hip extensors were not stimulated at all so that gravity was the only force working to extend the hip. Direct activation of hip extensor and flexor muscles would lead to improved control.

VII. DISCUSSION

Preliminary testing of the laboratory version of the CBO indicated that the system functioned as intended and is a viable means for further evaluation of the CBO concept. Full assessment of the concept requires testing in a large number of subjects with a variety of spinal injuries, at higher gait speeds and for longer distances. The indications for the CBO are the same as for any FES-aided gait system; however, the particular target population is those who lack the ability to stand for long periods with stimulation alone and those who exhibit considerable variation in the reflex activation of swing phase.

Even if the CBO concept does prove feasible, considerable technical hurdles remain before a viable product could be produced. The ultimate success of this approach will depend on customer acceptance which in turn depends on practical product design issues such as size, weight, ease of donning and doffing, and cosmesis of the brace. The current laboratory version clearly falls short on all of these measures. Further, a lightweight, energy efficient controllable brake needs to be developed. When fully locked at their maximum holding torques, the four brakes on the laboratory CBO consume 27 W of electrical power, far too much for existing battery power sources of reasonable weight. A more appropriate design will require brakes which consume no power in the locked state.

Although these technical challenges are indeed daunting, they can be overcome if the CBO proves to provide sufficient benefit to a large population of SCI subjects. Ongoing clinical testing with the laboratory CBO is designed to estimate just what this benefit will be, and further design work will then target the development of a practical version of the CBO. With this information, a more accurate cost benefit tradeoff can be determined for CBO approach.

REFERENCES


[43] Innovation Sports, 7 Chrysler Street, Irvine, CA, MVP Knee Orthosis.


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