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Feasibility of a Hybrid-FES System for Gait Restoration in Paraplegics

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Abstract

This paper proposes a new configuration for a hybrid-FES gait restoration system, and presents a combination of simulation and experiment that support the feasibility of the proposed approach. Gait simulation results are presented that indicate the majority of load bearing and the majority of power for gait is provided by the legs (i.e., quadriceps muscle stimulation). Based on these simulations, experiments on healthy subjects indicate that the gait restoration approach should be capable of providing long periods of locomotion unimpeded by quadriceps muscle fatigue.

I. Introduction

PREVIOUS studies have demonstrated that functional electrical stimulation (FES) can be used for restoring gait to people with spinal cord injuries (SCI). Standing and walking can provide significant physiological and psychological benefits to those with SCI [1–9]. However, electrically stimulated muscles suffer from rapid fatigue [10], relatively poor controllability of stimulated muscles, and the inability to control degrees-of-freedom outside the sagittal plane. Hybrid systems, which combine FES with an orthosis, can address several of the shortcomings of FES-aided gait. Use of an orthosis reduces the duty cycle of muscle stimulation during stance, reduce the complexity of gait by imposing kinematic constraints, and enables closed-loop control of electrically stimulated muscle by providing a mounting platform for joint angle sensors. Recent research in hybrid FES gait restoration systems using surface and implanted electrodes include [11–15] and [16–18], respectively.

This paper assesses the feasibility of a hybrid FES system that is different from others previously reported ([11–18]). The approach uses modulated brakes at the hip and knee joints (like [11–15]), unidirectional joint coupling between knee and hip flexion, and a biasing spring that biases the knee joint, and via coupling the hip joint, toward a flexed position. The device is called the Joint Couple Orthosis (JCO). Due to the joint coupling and biasing, a swing-through gait can be generated utilizing only surface stimulation of the quadriceps group of each leg (i.e., one stimulation channel per leg). The orthosis is an energetically passive device because all the power for gait comes from the user's metabolic power supply, which enhances the physiological benefit for the user and also minimizes the battery requirements for the orthosis. In addition to joint angle sensors, the orthosis uses controllable brakes at the hip and knee joints, a unidirectional coupling between knee and hip flexion, and joint biasing springs at the knee. The purpose of the brakes, as in [11–15], is to lock to provide isometric joint torques at the knee during stance and thus decrease the duty cycle of muscle stimulation, and also to modulate joint torques to obtain improved control of limb trajectories. The purpose of the unidirectional joint coupling and biasing

spring is to enable a swing through gait cycle via stimulation of only the quadriceps group, which is a large muscle group that is easy to access and provides significant joint torque. The unidirectional joint coupling between knee and hip flexion is illustrated in Fig. 1. Note that stimulating the quadriceps only, and the use of joint biasing and coupling is similar to the approach described by Durfee et al. [14–15] (the energy storage orthosis, or ESO). The ESO utilizes springs that separately bias the hip and knee joints in flexion, and adds a controlled energy transfer mechanism consisting of the combination of pneumatic actuators, valves, and an accumulator to store energy from knee extension during swing, and subsequently transfer that energy to hip extension during stance. The JCO incorporates a single biasing spring (at the knee) and a (non-controlled) kinematic, unidirectional coupling between knee and hip flexion, such that knee flexion generates hip flexion directly, but knee extension does not result in hip extension. Thus the coupling in the JCO is simpler (and likely more efficient) and hip extension need not overcome a bias spring, but the approach sacrifices the propulsive capability of active hip extension during stance offered by the ESO. To assess the viability of gait with the JCO, a simulation was conducted to determine the sufficiency of propulsion and the required exertion from the upper body.

II. Simulation

A dynamic simulation of an SCI user walking with the JCO and using a walker for stability was conducted to assess the viability and physiological requirements of the proposed system. The human was modeled as a spatial seven-link, rigid body system, where the head, arms, and trunk are lumped into a single link. Normalized inertial parameters and link lengths were based on standard biomechanical data as given in [19], with an overall body mass of 65 kg and a height of 1.7 m. The interaction between the user and the walker was modeled by the combination of a moment in the frontal plane and two forces (a vertical and horizontal force) in the sagittal plane. All forces and moments were constrained by the stability requirements of the walker (i.e., the forces and moments must be such that all four legs of the walker remain in contact with the ground). Since the purpose of the simulation was to assess the sufficiency of propulsion, speed, and stability of gait; to measure the required exertion from the upper body; and to assess the associated duty cycle of muscle stimulation, the simulation did not consider the low-level control issues associated with combining muscle stimulation and controllable brakes for desired trajectory control, issues that have been treated in previous work [11, 12, 20]. Instead, the quadriceps were assumed to generate a constant torque when stimulated with muscle dynamics represented by a low pass filter, and the brakes were controlled to be either locked or unlocked. In the proposed approach, the quadriceps are only stimulated to extend the knee during swing.

The controller implemented in the simulation was a finite-state machine, with the four states described below (Fig. 6).

1) State 1(double support)

In the initial condition, both knee brakes are locked in full extension, with one leg forward and the other back. The hip brakes remain unlocked, enabling the upper body to rotate freely about the hips. Using the arms, the user pulls the upper body forward until the forward hip is flexed to 15 degrees, at which point the forward hip brake is locked, and the center of mass of the user is located vertically above or near the forward foot. In this configuration, relatively little arm force is required to move the center of mass forward, a motion which is further aided by gravity once the center of mass passes over the forward foot. For purposes of the simulation, the initiation of swing (State 2) is triggered when the support leg is in a vertical orientation. Since the forward stance hip is locked at 15 degrees of flexion and since the support leg is in a vertical orientation, the center of mass is clearly located between the forward foot and the walker.

2) State 2 (swing flexion)

At the end of State 1, the center of mass of the user is located between the forward foot and the walker, so that the backward leg is ready for swing. Swing is initiated by unlocking the knee brake, which allows the bias spring to flex the knee, and also, due to the joint coupling, the hip. During this state, the user is assumed to maintain a vertical upper body orientation. This is implemented in the simulation by modeling the horizontal arm forces as a spring and damper that maintains the torso in a vertical position. The same arm model is used in States 3 and 4. The stance hip brake allows hip extension of the stance leg to zero degrees, at which point it locks, and the stance knee brake remains locked. The prevention of hip hyper-extension generally maintains the center of mass in a vertical plane between the stance leg and the walker, which enhances postural stability, although it also results in smaller step lengths. The controller switches to the next state, swing knee extension, when the swing knee reaches a maximum flexion angle and the knee angular velocity is zero.

3) State 3 (swing extension)

The quadriceps is stimulated, which extends the swing knee. 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.

4) State 4 (heel strike)

The swing knee is locked and the swing foot lands due to gravity and user arm control. The hip brakes remain unlocked. The controller switches to the next state when the swing leg, now in front, supports thirty percent of the weight. The swing leg becomes the stance leg, and the system is switched to state 1 and is ready for another step. Note that one control cycle represents one step (i.e., one half stride), and therefore, two control cycles are required for each gait cycle.

III. Simulation Results

Simulations of the JCO and gait controller were conducted in Matlab/Simulink. In the simulation, the user starts from rest and at steady state maintains a cadence of 22 steps/min, which results in an average forward velocity of about 0.1 m/s. Figure 2 shows the foot-floor forces, arm forces, and frontal plane moments, all normalized to the body weight of 637 N. The top two plots in Fig. 2 (body weight normalized foot/floor and arm/walker forces) indicate that more than 90% of body weight is supported by the legs, and thus the arms act primarily to stabilize the body, rather than counteract gravity. Fig. 3 shows the gait kinematics for twenty seconds of simulation. The angles and general gait pattern are similar to those exhibited during healthy biomechanical gait although the forward velocity is considerably slower. Fig. 4 shows the power expended by the quadriceps and arms. The quadriceps delivers power during State 3 when knee is extending, although this power is also used during swing when the biasing spring flexes the knee and hip. The arms are active through the entire cycle, but are characterized by a considerable low mean value. The ratio of root-mean-square arm to quadriceps power is approximately 0.26. Negative arm power, which means the arm is absorbing power, does not require as much metabolic energy as generating power. Computing the ratio of arm to leg power using non-rectified average power, the ratio is 0.04. Therefore the arms generate somewhere between 4% and 26% of the total power required for gait. In either case, the majority of power required for gait is generated by the quadriceps muscles. As shown in Fig. 5 (by the control state diagram), the duty cycle of quadriceps stimulation (which occurs in State 3) over a full cycle is approximately 10.2%. Finally, Fig. 6 illustrates the animated output of the dynamic gait simulation, which provides a qualitative perspective on the resulting gait.

IV. Experiments

Rapid muscle fatigue has been a significant issue in surface FES-based gait restoration [10]. Based on the simulation data shown in Figs. 2–5, the quadriceps are used only 10.2% of the time, and specifically during knee extension in late swing. Despite this, when the quadriceps are active, they are performing significantly more work than they would in biomechanically healthy gait, since they are storing energy in the biasing spring, which later flexes both the knee and hip joints. Experiments were conducted on ten healthy subjects that emulated the conditions described by the simulation. The experimental setup consisted of a knee joint orthosis biased in flexion by a gas spring with properties representative of those used in the simulation. In the experiments, each subject donned a pair of quadriceps electrodes, strapped their right leg into a test orthosis, and stood atop a block on their left leg, such that the right knee joint was free to flex and extend without contact with the ground. For each subject, the experiments consisted of five trials, where in each trial, the quadriceps was stimulated for 0.6 sec every 4.0 sec (duty cycle of 15%, representative of gait simulations) for a period of five minutes (i.e., a total of 25 min for all trials). Subjects were given a three-minute rest period between each five-minute trial. Muscle stimulation for the experiments was provided with a custom-fabricated computer-controlled stimulator set to a 45 Hz pulse frequency and a 300 microsecond pulse width. The subjects self-selected the stimulation amplitude (between 0 and 150 milliamps) prior to the experimental trials, which once selected remained at a constant level for the five trials. Because the test orthosis did not include a locking knee brake, the knee joint returned to the flexed position immediately following the quadriceps stimulation unlike in the proposed gait sequence where the knee joint is locked at full extension following quadriceps stimulation, and unlocks only during the swing phase of gait). Representative knee angle data for the full set of five, five-minute trials is shown in Fig. 7(a), which shows the envelope of the knee movement during the 25 minutes of quadriceps stimulation along with the amplitude of knee motion, which is approximately one radian for the trial shown. This amplitude of motion was computed for each subject, and the result averaged across all subjects Fig. 7(b). As shown in the data, after an initial warm-up period, the amplitude of knee motion stabilizes in time, indicating that quadriceps fatigue would not have a significant effect on the proposed gait system. This means the proposed JCO approach should be capable of providing long periods of locomotion unimpeded by quadriceps muscle fatigue.

V. Conclusion

This study investigated the viability of a hybrid gait system that combines unidirectional knee and hip coupling with a knee flexion bias spring, and surface stimulation of the quadriceps. The simulated gait speed was 0.1 m/s. Simulation indicated that 90% of the body weight is carried by the legs, and thus the arms act primarily to stabilize the body, and to ensure the center of pressure remains between the stance leg and the stability aid. The simulation further showed that the arms supply somewhere between 4% and 26%, depending on whether or not the user is considered energetically conservative, of the power for gait, relative to the legs. The simulation also indicated the quadriceps should be stimulated at a duty cycle of approximately 10%. To assess the extent of quadriceps fatigue in the proposed system, experiments were conducted on healthy subjects, which imposed a duty cycle and workload representative of that required in the simulation. The results of these experiments indicated that the gait restoration approach should be capable of providing long periods of locomotion unimpeded by quadriceps muscle fatigue.

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Fig. 1.
Schematic of unidirectional joint coupling in JCO.

Fig. 2.
Leg and arm forces and torques during walking.

Fig. 3.
Forward progression and joint angles during walking.

Fig. 4.
Body mass normalized leg and arm power while walking.

Fig. 5.
Control state and hip and knee brake locking during walking.

Fig. 6.
Video sequence of walking with JCO, corresponding to the simulation data shown in Figs. 2 through 5. (State X is labeled as S_X)

Fig. 7.

(a) Knee angle for a single subject, for all five trials (rest periods, which occur every 300 sec, are not shown), (b) average knee amplitude across all subjects for the 25 minute period.