FES-assisted Locomotion on an Inclined Treadmill: Use of Passive Dynamics

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Introduction

Functional electrical stimulation (FES) using surface electrodes has been used to facilitate walking in spinal cord injured (SCI) patients for decades [1,2]. Although established as an assistive technology, neuroprostheses for walking are not widely accepted by SCI patients and are seldom used in activities of daily living (ADL). Two limitations of surface FES are: 1) muscle fatigue occurs rapidly, within minutes after walking is initiated; and 2) there is a need to assist hip flexion by swinging the upper body.

Our intention is to use surface stimulation as a form of therapy to help restore voluntary walking function through repetitive locomotion training. In the last five years, a new phenomenon has been reported from many sources that regular use of FES sometimes results in the recovery of voluntary function. One plausible explanation for this is that electrical stimulation of motor axons sends antidromic impulses into the spinal cord. When these impulses coincide with voluntary muscle effort, there is a possibility that synaptic modifications are made resulting in restored neural conduction. According to this hypothesis, direct muscle stimulation would have greater rehabilitative potential than the stimulation of reflexes, e.g. peroneal nerve stimulation, which is still the most popular mode of FES-assisted walking.

One limitation of surface FES is that the hip flexors cannot be stimulated directly; they are innervated too profoundly. The main sources of hip flexion moment, then, are voluntary contractions and gravity. When voluntary iliopsoas contractions are too weak or absent, gravity must be relied upon. To maximize the gravitational component of hip flexion, the hip should be fully extended and the foot placed as far as possible behind the hip joint. This can be accomplished using a treadmill with an overhead harness and a downward incline. The belt of the treadmill will literally drag the swing foot into this ideal position. Such a system is basically an “active” alternative to body-weight-support treadmill training, which is proving to be an effective procedure in SCI rehabilitation.

To overcome fatigue and lack of hip flexion moment, we propose the following innovations. Rapid fatigue may be eliminated by: 1) stimulating muscles only when needed instead of stimulating them during the entire gait phase in which they contribute to the locomotion; and 2) exploiting the passive dynamics of the subject’s legs to facilitate locomotion.

Methods

A computer model of the swing phase of human gait was developed in order to evaluate the torque requirements during swing phase when an able-bodied subject walks on a treadmill. The main concern of this study was to minimize the torques and exploit passive leg dynamics to assist the swing phase. Special focus was placed on reducing intrinsic hip torque to zero, since hip flexors are not accessible by surface FES. Downhill walking was used as a means to generate passive hip flexion caused by gravity.

1.1 Leg Model
The kinematic and dynamic models were developed using Newton-Euler and Lagrangian representations. The Lagrangian model was developed to provide inverse dynamics calculations. The Newton-Euler model was used as a forward dynamic model to validate the Langrangian model. The models described the human leg as three
segments: foot, shank and thigh. The segments were modelled as rigid bodies. The ankle, knee and hip joints were modelled as pin joints with axes orthogonal to the sagittal plane. The model parameters were obtained from Winter [3]. Matlab 6.1 (Mathworks Inc.) was used to develop the models.

Model validation consisted of:

Step-1: A set of kinematic trajectories was provided to the forward dynamic model (Newton-Euler model) to obtain joint torques as functions of time. Then, the calculated joint torques were input to the inverse dynamic model (Langrangian model) to obtain kinematic trajectories. The original and calculated kinematic trajectories were compared, and they matched.

Step-2: Kinematic data from an able-bodied subject during locomotion (taken from Winter [3]) were provided to the Newton-Euler model and the joint torques were calculated. The calculated torques were compared to Winter’s calculations. The correlations found were 0.9915, 0.9985 and 0.9791 for hip, knee and ankle joint torques, respectively. Based on an ANOVA analysis, each of the torque curves fitted within a 95% confidence interval ($F_{0.05,2,27}$ of .2815, .0008, and .1550 for the hip, knee and ankle joint torques, respectively).

1.2 Simulation of Downhill Locomotion

Winter’s kinematic data for normal locomotion [3] were modified for downhill walking. Under normal conditions, the hip flexors are required to advance the swing foot forward, then the quadriceps must extend the knee before heel contact. As expected, downhill walking was found to reduce the torque requirements of the hip flexors as the angle of inclination increased. It was also found that the quadriceps muscles did not have to contract during the swing phase, since the foot was landing just ahead of the hip, i.e. gravity pulled the lower leg into heel contact position just before the end of swing phase. However, the intrinsic hip torque needed to perform swing phase could not be reduced to zero. Therefore, a new torque profile was hand-crafted in which the hip torque was set to zero throughout the entire swing phase, while knee flexion and ankle dorsiflexion were increased slightly. Simulations showed that this profile caused the foot to lift off the ground and the leg to swing forward. The end result was a walking pattern without any intrinsic hip torque. The torque profiles obtained from these simulations were optimized to reduce the overall energy requirements of the muscles while maintaining an acceptable gait pattern.

1.3 Development of the Stimulation Protocol for FES-assisted Locomotion

An experiment was conducted to determine if the joint torque profiles developed in the previous section could be produced in a particular SCI subject (female, T4 complete, 12 months post injury). Since the purpose of this project was to prove the concept that a neuromuscular prosthesis can be developed for treadmill training that does not stimulate hip flexion, one SCI subject was sufficient. This subject was chosen because she did not have denervated muscles that actuated knee and ankle joints. This research program was approved by the Toronto Rehabilitation Institute Ethics Board and the subject signed a letter of consent.

FES was applied using a Compex Motion stimulator (Compex SA, Switzerland) [4]. Since, the simulations in 1.2 provided torque requirements for each joint as a function of time, it was needed to determine intensity of the stimulation pulses as a function of time (stimulation protocol) that can generate the required joint torques. To calculate the stimulation protocol, the torques produced by the hamstrings would have to be known for a variety of speeds and currents throughout the range of motion. The similar calculations were not as necessary for the tibialis anterior muscle due to the low torque requirements and small range of motion (simple on-off strategy was sufficient).

First, a muscle strengthening program was administered since the subject had suffered significant muscle atrophy since her injury. She underwent 15 training sessions over four weeks. Stimulation was applied to hamstrings and tibialis anterior. After the muscle strengthening program, the torque produced by the hamstrings when undergoing FES was measured using a Biodex-2®. Isokinetic trials were conducted at fixed angular
velocities ranging from 30-105 deg/s in 15 deg/s increments; and fixed stimulation currents from 80-110 mA in 10 mA increments. Biphasic stimulation pulses were used at a constant pulse width of 250 µs and frequency of 40 Hz. Knee torque was plotted as a function of knee angular velocity and stimulation current. The torque profiles generated by the model were then matched up with the profiles measured by the Biodex-2® resulting in a plot of current versus time that generates the swing phase of gait. The sequences were then programmed into the Compex Motion stimulator.

Results

Figure 1 shows the torque produced in the hamstrings when 110 mA of pulse amplitude was applied under varying isokinetic conditions. This surface overlaps the proposed torque necessary to produce downhill walking. The most significant observation here was that this level of current far exceeds what would be required, where the difference can be viewed as inefficiency leading to early fatigue.

The knee joint torques produced by FES were measured at a variety of angular velocities and currents to create torque profiles versus current, speed, and position (see Figure 1). These profiles were then used to calculate the stimulation protocol for the hamstrings muscle group as a function of time that would facilitate downhill locomotion (see Figure 2). As mentioned earlier the tibialis anterior muscle only needed to be contracted for a very short time period during lift off phase of the gate to provide proper clearance to initiate toe off. Toe clearance was provided solely by knee flexion.

Discussion

The purpose of this is study was to test the hypothesis that by applying the inclined treadmill, one can facilitate FES supported locomotion without intrinsic hip flexion moment. Also, by having the participant fully supported in the harness, the quadriceps and gastrocnemius muscles are not required to facilitate swing phase. Our current efforts are aimed at calculating minimum torques that need to be generated by quadriceps and gastrocnemius muscles to facilitate full load bearing during downhill treadmill locomotion. The theoretical analysis presented here demonstrates that a stimulation protocol is feasible to generate locomotion without stimulating hip flexors as long as the treadmill is inclined at 10 deg. The proposed protocol utilizes passive leg dynamics to produce hip flexion moment via gravity. These theoretical results need to be tested and validated in a clinical environment. Once tested, this stimulation protocol will be used to facilitate FES-assisted locomotion training as an alternative method for current treadmill training in SCI subjects [5]. It is well established that locomotion training in incomplete SCI subject promotes recovery of walking function. Our intention is to develop a neuroprosthesis system that can facilitate locomotion training using FES and apply it in a clinical environment.

References


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Figure 1; Surface represents the knee joint torque as a function of angular position and angular velocity, produced by hamstrings m. using FES for constant stimulation pulses that have 110 mA amplitude. Curve represents required knee joint torque as a function of angular position and angular velocity that facilitates swing during downhill locomotion.

Figure 2; a) Knee and ankle joint torques as a function of time needed to facilitate swing during downhill locomotion; b) Stimulation protocols applied to hamstrings m. and tibialis anterior m. that generate torques in a) as a function of time.