The feasibility of a FNS powered mechanical gait orthosis with coordinated joint locking

To CS [1,2], Kirsch RF [1,2,3], Kobetic R [2], Triolo RJ [1,2,3]

[1] Department of Biomedical Engineering, Case Western Reserve University, Cleveland, Ohio
[2] Motion Study Laboratory, Louis Stokes Cleveland Department of Veterans Affairs Medical Center, Cleveland, Ohio
[3] Department of Orthopaedics, Case Western Reserve University

Email: ronald.triolo@cwru.edu

Abstract

The purpose of this study was to examine the feasibility of a hybrid orthosis for walking after spinal cord injury (SCI) that coordinates the locking and unlocking of knee and ankle joints of a reciprocating gait orthosis (RGO), while injecting propulsive forces and controlling unlocked joints with functional neuromuscular stimulation (FNS). The effectiveness of the hybrid system relative to gait stability and posture were determined in this simulation study. A three-dimensional computer model of a hybrid orthosis system (HOS) combining FNS with a RGO incorporating feedback control of muscle activation and coordinated joint locking was developed in Working Model 3D. The simulated hybrid orthosis system achieved gait speeds, stride lengths, and cadences of 0.51 ± 0.03 m/s, 0.85 ± 0.04 m, and 72 ± 4 steps/min respectively, exceeding the performance of other hybrid systems. Forward trunk tilt was found to be necessary during initial step from standing and pre-swing, but posture and stability were significantly improved over FNS-only systems. The results of the model shows that a HOS that coordinates knee and ankle joint locking with electrical stimulation to the paralyzed muscles holds significant advantages over brace- and FNS-only walking systems in terms of enhanced trunk stability and posture.

1 Introduction

The reciprocating gait orthosis (RGO) is a trunk-hip-knee-ankle-foot orthosis (THKAFO) for assistive walking after spinal cord injury (SCI) that reciprocally couples hip movement allowing the moment generated during hip extension to assist with contralateral hip flexion. However, brace-only ambulation is slow, energy intensive and dynamically unstable. Walking without external bracing by means of functional neuromuscular stimulation (FNS) [1-3] incurs high rates of muscle fatigue, leading to instability, poor control, limited walking distances [2], and large anterior trunk tilt of 10-48^{o} [3]. Freeing one or more joints of the lower extremity from the RGO constraints can result in an enhanced gait speed and cadence over brace-only walking [4]. Hybrid orthosis systems (HOS) exploit the mechanical stability of bracing while allowing a degree of joint mobility that can be controlled by FNS [5,6]. Since reciprocal coupling of the RGO prevents bilateral hip flexion, walking with an HOS can reduce trunk tilt to 5-18^{o}, extend walking distances, and delay the onset of fatigue. The HOS has been applied only to a limited extent to provide hip or knee mobility under FNS control without consideration of the ankle [5,6]. The objectives of this simulation study was to examine the effect of locking and unlocking the knee and ankle joints of the RGO during gait while maintaining control of the freely moving limb by FNS and to investigate the possibility of maintaining a vertical posture throughout gait.

2 Methods

A three-dimensional computer HOS model was developed for this study. The model and all associated simulations were implemented in Working Model 3D v4.0 (Knowledge Revolution: San Mateo, California).

1) *Human model*: A kinematic model of an anthropomorphic human body [7] was created to approximate a long-time user of FNS-only walking systems [5]. The trunk was assumed to be one rigid body. Each of the lower extremities possessed thirteen degrees of freedom. The foot-ground contact model was adapted from [8]. The passive joint moments were applied to the human model using equations adapted from [9] and [10].

2) *Modeling FNS*: The actions of FNS on the paralyzed muscle were approximated by applying torque constraints to the lower extremity joints. Torque constraints applied to the joints (1) were modelled as pure torque generators, (2) activated as pure step functions, (3) produce movements only in the sagittal plane, and (4) have maximum values less than moments produced by FNS of paralyzed muscle for the corresponding joint (Figure 1). Derived from FNS-only experimental gait trials [1] and electromyographic (EMG) studies of normal [11] and RGO gait [4], the activation of a torque depends on the angle of joints, the locked state of a joint, and/or the activation of other torques in a closed-loop fashion.

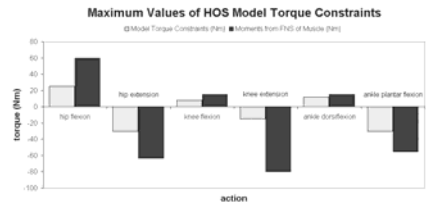

Figure 1: *Maximum Values of HOS Model Torque Constraints – the maximum torque magnitudes used in the HOS model are lower than the average moments [1] measured with FNS activation of paralyzed muscle.*

3) *RGO model*: The isocentric reciprocating gait orthosis (IRGO) utilizes a reciprocating bar that extends across the hip joints to couple the hips. The IRGO was specified to set the following constraints on the human model: (1) limit the hip and ankle joint rotation to the sagittal plane, (2) unlock/lock the knees and ankles on demand during the gait cycle, and (3) provide a 1:1 flexion/extension coupling ratio between the hips. The closed-loop algorithms for locking or unlocking a knee or ankle joint are dictated by the joint angle and the activity of torque constraints.

4) *Customized walking aid/arm supports*: The walking aid incorporated into the HOS model was designed to maintain lateral and anterior/posterior trunk stability by applying forces and moments to the shoulders. The change in length of each support is proportional to the vertical velocity of the contralateral hip. Each support allows forward trunk tilt from heel strike to the end of initial swing.

Three-dimensional computer simulations, utilizing Kutta-Merson integration, of walking on a level surface were performed with the HOS model. Initially, the model is in a quiet standing posture with both arm supports in contact with the ground. The duration of the simulation was the time to complete ten steps.

3 Results

Simulations with the HOS model exhibited a walking velocity of 0.51 ± 0.03 m/s, a stride length of 0.85 ± 0.04 m, and a cadence of 72 ± 4 steps/min. The top of Figure 2 depicts a typical gait cycle. The plots of Figure 2 show the locked state of the knee and ankle joints and the orientation of the joints relative to the percentage gait cycle. The knee is locked at the end of terminal swing when it is fully extended and remains locked during the stance phase. The knee is unlocked after the pre-swing phase to permit knee flexion during swing. The ankle is locked at initial contact and remains locked to the end of terminal stance. The ankle is unlocked during pre-swing to allow the ankle to plantar flex for push-off and remains unlocked after pre-swing to allow for ankle dorsiflexion. The hip angle ranges from 15^o flexion to 15^o extension, since the reciprocator confines the user to a 1:1 flexion/extension ratio.

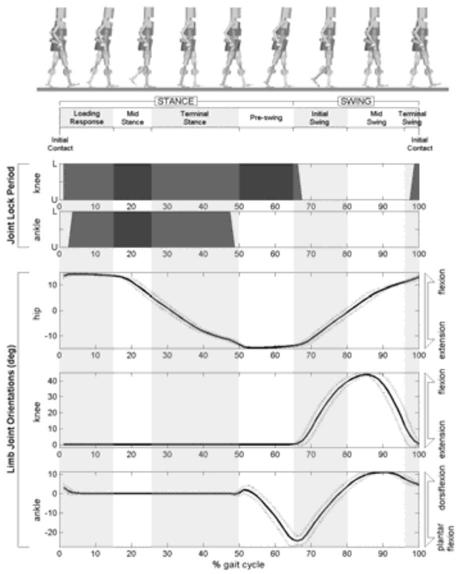

Figure 2: *A right gait cycle of the HOS model (the arm supports are not shown for clarity). The alternating shaded and clear regions separate the phases of the gait cycle. Plots: Joint Locking Period – L = locked; U = unlocked; Limb Joint Orientations (deg) – mean hip, knee, and ankle orientations with one standard deviation.*

The lateral trunk tilt is minimal ($\pm 1^o$) due to the balance provided by the arm supports. Also, there is no significant backward trunk tilt. Minimal forward trunk tilt (7^o) occurs during the loading response and again in pre-swing. Upright posture is restored during the late stance and swing phases.

The majority of the shoulder forces act vertically, occurring sporadically in the form of transient spikes (50-70% of the total model weight) during swing. The shoulder torques suggest that the posterior trunk orientation depends minimally on the arm supports except during the initial step from standing at which, the supports supply a declining backward moment with an onset peak value of 124 Nm.

4 Discussion and Conclusions

In a study conducted by Kobetic *et al.* (2003), a subject with paraplegia using an HOS with the ankle locked achieved an average gait speed of 0.32 m/s, stride length of 0.64 m, and cadence of 58 steps/min [5]. Jefferson and Whittle (1990) reported speed and cadence values of 0.30 m/s and 35 steps/min respectively from assisted gait with a RGO-only system [12]. In the HOS model, ankle plantar flexion during pre-swing assists in forward propulsion and the transfer of body weight during double stance. During the swing phase of HOS gait, knee flexion and ankle dorsiflexion allow for successful foot-floor clearance without the lateral trunk tilt needed in stiff-legged brace-only gait. The gait speed reported for a 26 channel FNS-only system was 0.55 m/s, cadence was 58 steps/min, and stride length was 1.14 m [5]. The HOS model was able to use lower active joint moments, yet still perform similarly to the FNS-only system due to the enhanced stability and the reciprocal coupling of the hips provided by the IRGO. The HOS further minimizes fatigue by allowing muscles to rest when the joint is locked.

As a result of knee flexion in the swing phase of HOS gait, lateral trunk tilt is not required for foot clearance and is minimal. Also, the design of the RGO causes hip flexion by FNS control to indirectly force the trunk to tilt backward behind the stance foot. Because the upper body is upright when standing, the shoulder moments indicates that during the initial step, back support from the walking aid is needed to prevent posterior trunk tilt. Later steps do not require such support because the anterior trunk tilt occurs at pre-swing. Ensuing hip flexion during swing will only restore upright posture rather than force the trunk behind the stance foot. To prevent backward trunk tilt beyond the stance foot during the first step, a modest degree of anterior trunk tilt could be maintained before taking the first step.

The HOS model provides evidence that a RGO with coordinated joint locking powered by FNS has benefits over existing assistive gait systems. First, coordinated joint locking allows for a natural swing phase, removing the instabilities caused by lateral trunk movements found in stiff-legged walking. Second, longer walking distances might be achieved with the HOS as opposed to the FNS-only system because less muscle activation is needed to drive freed joints and the muscles controlling a particular joint are allowed to rest when that joint is locked. Third, an upright posture can be maintained with minimal trunk tilt necessary only intermittently during some phases of gait.

References

[1] Kobetic, Synthesis of paraplegic gait with multichannel FES, *IEEE TRE*, 2(2): 66-79, 1994.

[2] Kobetic, Implanted FES system for mobility in paraplegia: A follow-up case report, *IEEE TRE*, 7(4): 390-398, 1999.

[3] Kobetic, Muscle selection and walking performance of multichannel FES systems for ambulation in paraplegia, *IEEE TRE*, 5: 23-29, 1997.

[4] Yang, Effects of joint motion constraints on the gait of normal subjects and their implications on the further development of hybrid FES orthosis for paraplegic persons, *J. Biomech.*, 29(2):217-226, 1996.

[5] Kobetic, Development of a hybrid gait orthosis, *J. Spinal Cord Med.*, 26(3):254-258, 2003.

[6] Solomonow, Standing and walking after spinal cord injury: Experience with the reciprocating gait orthosis powered by electrical muscle stimulation," *Top Spinal Cord Inj Rehabil*, 5(4):29-53, 2000.

[7] Winter, *Biomechanics and Motor Control of Human Movement*, 2nd ed. NY: John Wiley Sons, Inc., pp. 57, 1990.

[8] Wright, Passive regulation of impact forces in heel-toe running, *Clin Biomch*, 13:521-531, 1998.

[9] Edrich, Analysis of passive elastic joint moments in paraplegics, *IEEE Trans BME*, 47(8):1058-1065, 2000.

[10] Esteki, An experimentally based nonlinear viscoelastic model of joint passive moment *J Biomech*, 29(4): 443-450, 1996.

[11] Perry, *Gait Analysis: Normal and Pathological Function* Thorofare, NJ: SLACK Inc., 1992.

[12] Jefferson, Performance of three walking orthosis for the paralyzed *Prost Ortho Int*, 14:103-110, 1990.

Acknowledgements

This work was supported by the NSF-IGERT Neuro-mechanical Systems Program. To sincerely thanks Davy, Quinn, Mansour, Audu, and Amankwah for their advice.