TRUNK MOMENTUM CAN REDUCE UPPER LIMB FORCES IN FES AIDED PARAPLEGIC LOCOMOTION

Brian J. Andrews, Adam Thrasher, Arthur Chan, Rahman Davoodi

National Spinal Injuries Center, Stoke Mandeville Hospital, O.O.E.C. University of Oxford, U.K.
Department of Biomedical Engineering, University of Alberta, Canada
E-mail: b.andrews@ualberta.ca

Abstract – *The use of trunk momentum to assist FES aided paraplegic locomotion was examined via computer simulation and clinical trials. While standing in a pair of medially linked knee-ankle-foot orthoses, paraplegic subject (T12) was able to lift his feet in turn by oscillating his trunk. He was able to do so at frequencies ranging from 0.38 to 0.55 Hz, comparable to swing frequencies of passive gait. An able-bodied subject also performed these movements, and was able to walk at speeds up to 0.61 m/s. Two computer models demonstrated rocking as well, but stable walking patterns could not be found. Another model demonstrated that the Coriolis effect of trunk momentum did not contribute significantly to sit-to-stand transfers.*

Keywords: Trunk momentum, sit-to-stand, gait, hybrid FES orthosis

1. Introduction

Clinical FES systems for assisting paraplegic locomotion have limited application because they require excessive physical effort and large upper limb forces. In man the shoulder girdle is ill-adapted for locomotion. Furthermore, many paraplegics suffer from overuse syndromes of the shoulder [5] [2]. The motivation for the present studies was to explore the possible role of the trunk as a "spinal engine" [4] that may help unload the upper limbs during FES sit-to-stand and gait.

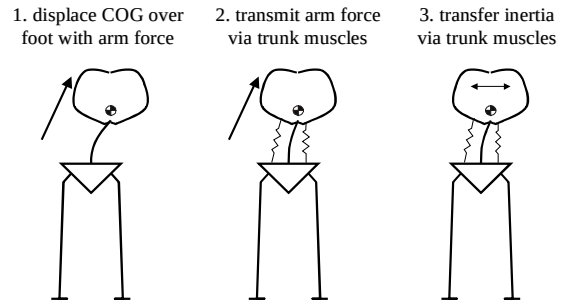
FES sit-to-stand typically begins with the subject moving to the edge of the seat and assuming a static posture with the ankles, knees and hips almost fully flexed [7]. This way, the lifting forces must all come from the arms and the lower limb joints. Since the knee extensors under FES are weak, the arms must bear 30-45% of the body weight [11] [6]. In a computer simulation study, Davoodi and Andrews [3] theorized that the arms could be unloaded completely using a trunk momentum strategy, even with the reduced strength of stimulated knee extensors.

In normal sit-to-stand, there is a phase prior to lift-off when the trunk rotates forward. It is generally accepted that the purpose of this is to position the center of mass closer to the new base of support and to produce momentum that will carry the upper body forward after lift-off. The movement of the trunk is typically more vigorous when the arms are not used; one will thrust one's trunk forward further and faster. In one of the present studies, we examine the proposition that trunk rotation generates a Coriolis force that contributes vertical force at lift-off.

We used a hybrid system based on the medially linked knee-ankle-foot orthosis (mLKAFO) [1, 14] and depicted in figure 1. Two KAFOs are linked by a medial joint below the perineum to provide better medio-lateral stability during standing than lateral hip joints aligned over the greater trochanters. The ankle-foot sections of the mLKAFO are of the floor reaction (FRO) type and can be rapidly detached for convenience in activities of daily living. The FROs can be used alone with FES, for example, to prolong standing as described in [13]. Typically, the above-knee link joint component can be rapidly attached and used to further improve locomotion e.g. resist scissoring of the legs and increase M/L stability when standing.

Paraplegic mLKAFO ambulation is inefficient One reason is that volitional upper body force actions, transmitted through walking aids such as crutches, are ineffectively coupled to the lower limbs due to paralysis of the lumbar spine musculature. This is illustrated below as case (1). The crutch force cannot laterally tip the mLKAFO about the opposite foot because there is no mechanical coupling. Instead, the upper body must be supported by the arms whilst the COG is laterally displaced over the pivot foot, causing the mLKAFO to tip to provide ground clearance for the swing leg. Case (2) illustrates how FES can be usefully applied to the lumbar spine muscles to effectively stiffen the trunk in the M/L plane to allow the crutch force to generate the required lateral tipping moment.

Achieving foot clearance in mLKAFO



The mLKAFO also allows the trunk to articulate unrestrained. In case (3) properly controlled, lateral inertia of the trunk could conceivably produce useful forces on the lower limbs, providing foot clearance and stability during gait. This aspect will be further explored in this paper.

Figure 1: *Medially linked knee-ankle-foot orthosis designed for FES-assisted paraplegic gait. (Inset) close up view of uniaxial medial joint.*

Some insight was found in the robotics literature. Using anthropometrically scaled robots, McGeer [10] demonstrated that gait patterns could be sustained simply by the interaction of gravity and inertia, suggesting that humans are capable of walking with little to no motor control. Very recently, Kuo [8] simulated passive stiff-legged gait, but concluded it was intrinsically unstable in the frontal plane, much like an inverted pendulum. Among Kuo's proposed methods to stabilize lateral motion, or "roll," was a trunk that oscillated like a metronome. Six years earlier, Li et al. [9] built a biped robot with an oscillating trunk. They found that periodic trunk motion could stabilize the robot.

We further investigated the foot-clearing and stabilizing effects of an active trunk via computer simulations and clinical trials of the mLKAFO brace. If trunk momentum is useful in this way, FES applied to the paralyzed trunk may reduce arm forces and make hybrid orthoses possible for a greater population of spinal cord injured persons.

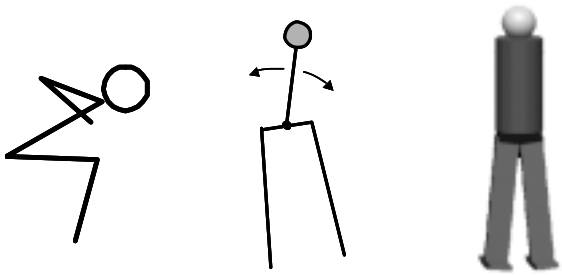
2. Methods

Two human subjects were used in our clinical studies: one paraplegic male (age 67, height 168 cm, weight 86 kg) with a complete spinal cord injury at T12, and one able-bodied male (age 22, height 168 cm, weight 62 kg). Both subjects were asked to perform certain tasks while in mLKAFO brace with knees locked in full extension. Starting from rest, the able-bodied subject was instructed to rock from side to side, lifting each foot in turn, using only trunk inertia. His arms were folded across his chest. Trials were recorded for slow, fast, and "most comfortable" rocking frequencies. The subject was then instructed to progress forward at his most comfortable step frequency. Then, he was asked to take longer strides. Two additional trials were recorded after the subject had 30 minutes of experience walking in the brace. The paraplegic subject was instructed to perform the same rocking trials at the able-bodied subject, only with arms poised to grab handrails in case of accidents. Due to hip contractures and spasticity, the paraplegic subject was not able to safely perform forward progression in the brace.

All movements were recorded using reflective markers and a four-camera Vicon motion analysis system [Vicon95]. Microswitches were attached to the shoes of the paraplegic subject to more accurately detect foot contact. The data was low-pass filtered (second-order Butterworth) at a cutoff frequency of 3 Hz. Residual analysis showed that the movements were slow, smooth and contained virtually no signal above 3 Hz.

Illustrated in Figure 2 are the three computer models of the human body employed in these studies. The first (Figure 2a), a 5-segment sagittal plane model, simulated the sit-to-stand transfer. It was scaled to the mass and dimensions of a 180 cm tall male weighing 75 kg. Although the hands rested on a fixed point in space, no moment was transmitted through the elbow or shoulder joints. Maximal hip and knee moments were computed for various trunk velocities. This model was programmed in MATLAB using equations of motion derived via Newton-Euler analysis.

Figure 2: *Biomechanical models. (a) sit-to-stand in sagittal place; (b) lateral rocking in mLKAFO brace, frontal planar model; (c) walking in mLKAFO brace, three-dimensional.*



(a) (b) (c)

The second model (Fig. 2b) was of a two-segment robot in the frontal plane, analogous to a human rocking laterally in the mLKAFO brace without hands. This model was scaled to the mass and dimensions of the paraplegic subject. The model was derived via Newton-Euler analysis and programmed in MATLAB. The lumbosacral joint was modeled as a semi-stiff joint with a motor that applies constant torque in the direction of the raised foot.

The third model was a 3-dimensional extension of the second model. The medial joint was added to enable the hips to articulate in flexion and extension. Prescribed torque at both hips, combined with the lateral motion of the trunk attempted to create gait patterns. This model was developed using WorkingModel 3D software. The trunk and hip motion were controlled using handcrafted algorithms based only on which foot was raised.

3. Results

The paraplegic subject used his arms and required assistance from operators to stand up in the brace. Due to contractures, he could not stand freely. A strap attached to a safety frame applied a horizontal force to his buttocks. Only then could he stand without holding on to the frame. For short periods of time (15-45s), he was able to lift his feet alternately by rocking his trunk side to side without using his arms. Figure 3 indicates the trunk motion during a typical rocking trial. Table 1 summarizes the frequency and amplitude of the trunk oscillations during the paraplegic trials.

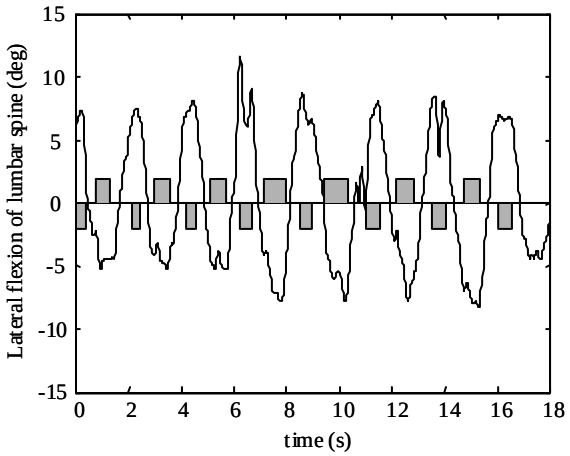


Figure 3: *Paraplegic (T12) subject in mLKAFO rocking from one foot to the other using trunk momentum. Negative (-) flexion is to the right. Shaded regions indicate when a foot lost contact with the ground; left foot - above zero; right foot – below zero.*

Trial	Frequency (Hz)	Mean amplitude (deg)
1	0.38	8.7
2	0.53	7.5
3	0.55	9.0
4	0.39	6.7
5	0.46	10.4
6	0.42	11.1

Table 1: *Summary of trunk motion for T12 paraplegic (168 cm, 86 kg) rocking from foot to foot in mLKAFO. Frequency was calculated by dividing number of oscillations by elapsed time. Amplitude reported here was the average peak lateral flexion of the lumbar spine.*

The able-bodied subject was able to stand up in the brace with arm support. He could then rock and walk without any arm support. Table 2 summarizes the trunk motion of the rocking trials, and Table 3 summarizes the walking trials. The subject was given 30 minutes to practice walking prior to trials 9 and 10.

Trial	Frequency (Hz)	Amplitude (deg)	Walking speed (m/s)
1	0.65	5.8	n/a
2	0.69	6.3	n/a
3	0.48	14.5	n/a
4	0.56	15.2	n/a
5	0.80	5.8	n/a
6	0.63	6.1	0.213
7	0.67	8.4	0.244
8	0.51	12.7	0.376
9	0.91	10.5	0.589
10	0.89	12.0	0.609

Table 2: *Summary of trunk motion for able-bodied subject (168 cm, 62 kg) in mLKAFO. Trials 1 through 5 involve rocking in place. Trials 6 to 10 involve walking. Amplitude reported here was the average peak lateral flexion of the lumbar spine.*

Using the sit-to-stand model, we found that increased trunk momentum at lift-off reduces required hip and knee moments via Coriolis effect, but very slightly. With trunk rotational velocities up to 180 degrees per second, the required knee moment was reduced by less than 10%.

The two-dimensional mLKAFO rocking model was also able to lift its feet one at a time by oscillating its trunk. Figure 4 shows a typical trial. The rocking frequencies varied from 0.4 to 1.2 Hz, with amplitudes not exceeding 6 degrees.

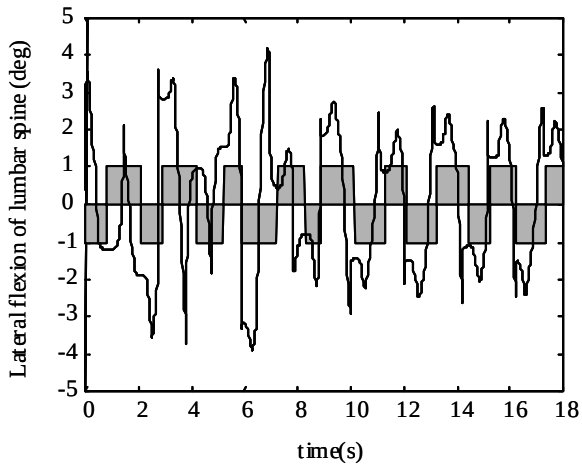


Figure 4: *Computer model of mLKAFO rocking from one foot to the other using trunk momentum. Negative (-) flexion is to the right. Shaded regions indicate when a foot lost contact with the ground; left foot - above zero; right foot – below zero.*

The three-dimensional mLKAFO gait model was also able to rock, but could only take one or two steps before falling over.

4. Discussion

The results from both the clinical trials and computer simulations of mLKAFO rocking and walking are promising. They suggest that trunk momentum can be a useful drive for gait, if properly harnessed. The paraplegic subject was able to articulate his trunk sufficiently to lift his feet from the ground, one at a time, at frequencies comparable to slow passive gait [10]. The able-bodied subject was able to oscillate his trunk at slightly higher frequencies than the paraplegic subject. This was not surprising, as the able-bodied subject has superior control of the trunk musculature. The paraplegic subject had a very low level injury (T12), therefore most of his trunk muscles were under voluntary control.

The two-dimensional "rocking model" was also able to lift its feet using oscillating trunk momentum at frequencies similar to both human subjects, however the amplitude was much lower. This was likely due to inaccurate assumptions made of the lumbosacral joint. Our able-bodied subject proved that coordinating hip flexion with trunk oscillations is possible, but our failed attempts using the three-dimensional model indicate that it is a difficult control problem. The able-bodied subject had sensory feedback that our model did not use.

As for the role of trunk momentum in sit-to-stand, our study concluded that the Coriolis component of trunk rotation contributed less than 10% of the vertical force required for lift off. This does not contradict the previous study by Davoodi and Andrews [3], which showed that hands free sit-to-stand could theoretically be possible with the reduced knee extensor strength of FES. Clearly, there is more at work than the Coriolis effect.

In conclusion, our results suggest that there is potential for the trunk to play an active roll in paraplegic locomotion. The question remains, is it practicable to activate and control the muscles of the lumbar spine via FES?

Acknowledgments: *This research was supported by grants from the Natural Sciences and Engineering Research Council (NSERC). The author Adam Thrasher was supported by a scholarship from the Alberta Heritage Foundation for Medical Research (AHFMR).*

References:

[1] Cliquet A., Baxendale R.H., Andrews B.J.(1989) Paraplegic locomotion and its metabolic energy expenditure, In: Comprehensive Neurologic Rehabilitation vol 3, Ed Rose, Jones & Vrbova, p 139-146, Demos USA.

[2] Bayley JS, Cochran TP, Sledge CB. (1987) The weight-bearing shoulder: The impingement syndrome in paraplegics. *J Bone Joint Surg*, 69A:676-678

[3] Davoodi R., Andrews B. J. (1997) Modeling and control of FES standing in paraplegia: possibility of unassisted standing-up, Proc. IFESS'97 & NP'97, pp. 93-94, Simon Fraser University, Vancouver BC, ISBN 0-86491-173-4.

[4] Gracovetsky S. (1988) The Spinal Engine, Springer-Verlag, Wein-New York, ISBN 3-211-82030-2

[5] Goldstein B, Young J, Escobedo EM. (1997) Rotator cuff repairs in individuals with paraplegia, *Am J Phys Med & Rehab*, 76:316-322

[6] Kamnik R, Bajd T, Kralj A. (1999) Functional electrical stimulation and arm supported sit-to-stand transfer after paraplegia: a study of kinetic parameters, *Artificial Organs*, 23:413-417

[7] Kralj A, Bajd T. (1989) Functional electrical stimulation: standing and walking after spinal cord injury. 1st edition CRC Press, pp. 97-102

[8] Kuo AD. (1999) Stabilization of lateral motion in passive dynamic walking, *Int J Robot Res*, 18(9):917-930

[9] Li Q, Takanishi A, Kato I. (1993) Learning control for a biped walking robot with a trunk, In Proc '93 Intern Conf on Intelligent Robots and Systems, Yokohama, Japan, July 1993, pp. 1771-1777

[10] McGeer T. (1990) Passive dynamic walking, *Int J Robot Res*, 9(2):62-82

[11] Riener R, Fuhr T. (1998) Patient-driven control of FES-supported standing up: a stimulation study, *IEEE Trans Rehab Eng*, 6:113-124

[12] Vicon 140 version 1.0 (1996) Movement analysis hardware & software, Oxford Metrics Ltd, Oxford

[13] Andrews et al.(1988) Hybrid FES Orthosis Incorporating Closed Loop Control and Sensory Feedback., J.Biomed Eng., 10,pp 189-195.

[14] Andrews BJ, Cliquet A, Olin M, Burnham R, Thrasher TA (1997) Paraplegic locomotion: a linked knee-ankle-foot hybrid FES system, Proc. IFESS'97, pp. 165-167, Simon Fraser Univ. Vancouver BC.