Walking with Functional Electrical Stimulation and Unlocking Braces in Thoracic-level Paraplegia

Jacques Bobet, SuLing Chong, Robert Rolf and Richard B. Stein

Abstract—Walking was tested in 4 people with thoracic-level paraplegia using stimulation of quadriceps muscles, the flexor reflex and unlocking knee-ankle-foot orthoses (KAFO). Heart rate, speed, distance, kinematics and ground reaction forces were measured while subjects walked using a walker. None of the subjects could walk without the system; all could walk continuously for at least 4 minutes with it. Joint angles and some other kinematic features resembled normal walking, but the walking was too slow (average speed: 3.8 m/min.) and too demanding (heart rate: 128 b/min; physiological cost index: 15 b/m) to be practical. Subjects supported about 1/3 of their weight with their arms during stance and about 2/3 during swing. Our results suggest that the braces reduced the effort needed and that the low speeds were due to both a lack of power at push-off and the time needed to stabilize the hip and trunk. The high heart rates arose from excessive contraction of the arm and trunk muscles for balance and propulsion.

I. INTRODUCTION

Walking-like movements can be restored in persons with complete thoracic spinal cord injury using functional electrical stimulation (FES) of the lower limb. However, the resulting walking is slower than normal and demands more effort. Perhaps the best FES-only system is the "Parastep" (Sigmedics Inc., Chicago IL). It stimulates the quadriceps, flexor reflex, and gluteal muscles. Using it, subjects walk at about 5 m/min, with a heart rate of about 170 beats per minute (b/min) [1]. For comparison, uninjured men walking with their customary normal cadence walk at 82 m/min, with a cadence of 108 steps/min (i.e., a stride time of just over 1 s) and a heart rate of 96 b/min [2]. In principle, walking performance can be improved by combining FES and orthoses for the lower limb [3]-[5]. These "hybrid" systems are an improvement over FES alone, but still do not approach normal walking speeds and levels of effort. The best results have been obtained by combining FES with a reciprocating gait orthosis (RGO). Solomonow et al. [6] reported speeds of 3-22 m/min, heart rates of 119 b/min, and walking distances of 400 m to 2 km with practice. The ‘ParaWalker’ hybrid system also uses surface FES and a rigid orthotic exoskeleton. In 16 complete thoracic paraplegics, this system gave mean speeds of 17 m/min, a mean physiological cost index (PCI)[7] of 3.1 b/m, and mean active heart rates of 134 b/min.[8]. Other combinations have been tested on a limited number of subjects[9]. These systems lock the knee and ankle, so the walking can appear unnatural. Others [6] [10-12] have added an unlocking knee to variants of an RGO, but performance was not dramatically improved.

Combining FES with an unlocking knee-ankle-foot orthosis ("KAFO", or "brace") may improve speed and reduce effort [13]-[14]. In an unlocking KAFO, the knee can be alternately locked and unlocked. Locking the knee during stance prevents it from flexing. The KAFO can then support the body without the need for stimulation. Unlocking the knee during swing allows it to flex, so the lower limb can be swung forward more easily [10]. Allowing both knee flexion and ankle dorsiflexion during swing is most beneficial [11]. However, there is little experimental data to support the use of this combination. Kagaya et al. [13] combined percutaneous FES with a KAFO containing a novel knee-locking device in one T8 complete paraplegic. Much less stimulation was needed with the knee-locking device than without. Walking speed was 6 m/min.; heart rate and energy cost were not reported. Stein et al. [14] used a custom KAFO with unlocking knee in one complete T4/5 paraplegic. With practice, he could walk at 5 m/min with a PCI just under 10 b/m.

In this study, we attempted to determine if KAFOs with FES and an unlocking knee can provide functionally useful walking. We did this by recording speed, effort, kinematics and ground reaction forces in 4 subjects with thoracic-level paraplegia.

II. METHODS

A. Hybrid System

The system we used [14] consisted of a modified KAFO and up to 4 channels of FES per leg. The knee joint can be locked against flexion at any angle, while allowing knee extension. When unlocked, the knee joint rotates freely, although a mechanical stop prevents hyperextension. The unlocking knee joints were obtained either from Horton Technologies (stance control KAFO), Little Rock, AK or Otto Bock (Sensor Walk), Minneapolis, MN and modified for push button control. Depressing a push button on one side elicited a flexor reflex and unlocked the knee on that leg. Releasing it ended the nerve stimulation, locked the knee, and turned on quadriceps stimulation long enough for the knee to extend and the foot to contact the ground. No stimulation was needed in double support because both knees

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were locked. The buttons were mounted on a conventional 4-wheeled walker. For some subjects, the 2 rear wheels had one-way locks such that they could roll forward, but not backward.

B. Subjects

Four male subjects aged 23-27 years gave informed consent, and the study was approved by the University of Alberta’s human ethics review board. All had essentially complete lesions (ASIA A) of the thoracic cord, with sparing of the motor neurons of interest. Injury levels were T4/5, T7/8, T9/11, and T8. Subjects were enrolled 8-60 months after injury. None could walk or stand unaided. All had enough arm strength and control to support their weight with a walker when the braces were locked and to operate the push buttons. Upon enrolment, each subject was provided with an 8-channel stimulator and instructions on how to use it. Subjects then independently completed a period of muscle stimulation intended to strengthen muscles and bones. Following this, subjects were fitted with custom carbon-fiber braces, and taught to walk with the braces and FES. Once they could use it safely, they took the system for home use.

C. Free walking sessions

Subjects returned about every 3 months for assessment of their walking. Subjects walked for 4 minutes at a self-selected speed along a flat, straight, tiled hallway using their usual walker. Resting heart rate (HR) while seated was measured for 2 minutes before walking with a portable heart rate monitor (Polar Electro, Finland). “Active” heart rate was measured during the last 2 minutes of walking. The total distance travelled was measured and used to calculate the average speed of the walk. This speed, in m/min, was then divided by the difference between active and resting heart rate (in b/min) to obtain the physiological cost index (PCI), a measure of walking efficiency [7]. In some subjects, we also examined other forms of walking (such as walking with locked braces but no FES – see below). We also occasionally asked subjects to walk as far as they could. These “endurance walks” were at a self-selected speed, and continued until the subject voluntarily stopped walking.

D. Motion Analysis Sessions

Detailed analyses of each subject’s walking were done at 6-month intervals using a Vicon (Oxford UK) MX recording system. Subjects walked on a flat, tiled 7-m walkway at a self-selected speed using their walker. Three Bertec (Columbus OH) force plates embedded in the walkway measured ground reaction forces and moments. Eight infrared cameras recorded the position of the reflective markers. Markers were placed over the toes, heels, ankles, knees, hips, and shoulders, and on both sides of the walker. For 2 subjects the walker handles were instrumented with 2 strain gauges (AMTI, Waterton MA) that recorded forces and torques in 3 dimensions. Subjects began the trials while seated in their wheelchair. They then stood up using quadriceps stimulation. Stimulation was turned off and the braces were locked. Once stable, subjects walked one length of the walkway. The same sequence was then repeated walking in the opposite direction.

A standard two-dimensional biomechanical analysis (temporal and kinematic variables) [15] was done using programs written in Matlab (MathWorks, Natick, MA). We also estimated the vertical force borne by the hands, either directly from the instrumented walker force plate or indirectly from the body weight and the force plate data; both methods gave similar results. Foot-floor contacts were identified from the video record. Between 5 and 60 strides were analyzed for each subject.

III. Results

A. Physiological Measures

In three of the subjects, we measured the physiological variables periodically for several months (7 to 29 months). In all three, speed increased to a plateau before declining slightly. PCI and HR did not show any consistent pattern with time. For these subjects, we used their most recent results for speed, PCI, and HR. For the fourth, who was studied for the longest period, we used his average values.

Overall, speed was much less than that of uninjured controls, while effort was higher. The mean and standard deviation of heart rate, PCI, and speed across all subjects were 128 ± 18 b/min, 15 ±11 b/m, and 3.8 ± 1.6 m/min. Three subjects completed 5 endurance walks. On average, they walked for 9.5±1.3 minutes and travelled 32±6.3 m.

B. Body Weight Support

The mean and standard deviation of vertical force taken by the feet was 424 ± 125 N (56 ± 13% of body weight) over the entire stride, 464 ± 116 N (61.6 ± 11.5%) in double support, and 295 ± 90 N (38.9 ± 9.1%) in single support. Subjects increased the amount of weight supported by the walker just prior to initiating swing.

C. Kinematics

Subjects used an alternating gait as in normal walking, showed good left and right symmetry, and exhibited a fairly normal swing phase. They moved the limbs in the sagittal plane, with only occasional external rotation of the legs. They did not elevate (‘hike’) their hips or circumduct the leg to help the swing foot clear the ground. In stance, they maintained the knee near full extension. However, their gait differed in important ways from normal walking. They walked very slowly, leaned forward excessively (about 20o of trunk flexion), and dropped the lower limb to the floor at the end of swing. They also advanced the walking frame and upper body in a series of short ‘pushes’ after foot contact, so that the trunk advanced discontinuously. The ankle angle was about 90o at heel strike, then gradually decreased (dorsiflexed) throughout stance as the lower limb rotated about the ankle. The ankle angle abruptly decreased further early in swing due to the flexor reflex. The ankle angle then increased (plantarflexed) late in swing; this was expected since the weight of the foot will cause the forefoot to drop once the reflex is over. The knee remained almost fully extended throughout stance. This too was expected since the knee joint on the prosthesis is locked during this phase. The

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knee angle decreased (knee flexion) as the stimulation was turned on and the knee joint unlocked; it then increased again (knee extension) when stimulation ceased and the quadriceps stimulation was initiated. The hip showed a similar pattern to the knee, remaining fully extended throughout stance before flexing, then extending in swing. On average for the 4 subjects, each stride took 15.6 ± 2.2 s of which about 75% was spent in double support. Stride length was 0.61 ± 0.11 m; this was largely dictated by the placement of the force plates. Walking speed was slower (2.4 ± 0.3 m/s) because of the need to place the feet on the force plates.

D. Additional Observations

In two subjects, we obtained the physiological measures with different combinations of bracing and FES. These were: 1) ankle-foot-orthosis (AFO) and FES, 2) brace with FES or 3) brace without FES. For condition 3), the knees of the brace were locked in full extension and the subject used a ‘swing-to’ gait. Results from one subject, who did repeated trials, are shown in Figure 1. Condition 3 produced significantly greater speeds and significantly lower PCIs. The other subject showed similar results, though only 1 trial was available.

![Figure 1: Physiological measures in 1 subject walking with either 1) AFO and FES, 2) brace with FES, or 3) brace without FES. He used an alternating gait in conditions (1) and (2), and a swing-to gait in (3). Mean and 1 standard deviation. * denotes significant difference (p<.05).](image)

In one subject, we stimulated the hip extensors (gluteus maximus) as well as the flexor reflex and quadriceps. We added the gluteal stimulation after the subject had been using the system for over a year. The subject then used this new system for 6 months, during which we measured his physiological variables 5 times. Comparing to the 6 month period immediately prior to adding the hip extensors, PCI improved significantly from 11.7 to 8.9 b/m. Speed and heart rate did not change significantly.

III. DISCUSSION

This combination of bracing and FES produced an alternating gait with knee flexion and extension. The gait produced some features of normal walking. However, walking speeds and effort did not approach normal. Using this system, subjects could stand at a counter, or walk short distances around their home. There may also be secondary health benefits for people with paraplegia simply from standing and walking regularly [18]. But the system provided only limited functional walking.

The heart rates we obtained were lower than those obtained with the Parastep system [1]. Since the main difference between these two systems is the KAFO, this suggests that the KAFO reduced the effort required to walk. This is presumably because it, rather than muscle, supported the body weight during stance.

Although the effort was improved somewhat compared to FES alone, walking speeds were still too low and energy costs too high. Speeds increased with time in some subjects, but were still well below normal even after many months of use. Indeed, the speed and efficiency were still lower than the swing-to gait commonly used with locked KAFOs.

The speeds and heart rates we obtained were not as good as those obtained with the RGO+FES or the ParaWalker. The main difference between our KAFO-based system and these other systems control the hip joint. This suggests that control of the hip joint is necessary for good results. Presumably, the hip control helps to stabilize the trunk. It may also improve the coupling between arms and legs, allowing energy generated by the arms or trunk to be transferred more readily to the legs.

The low walking speeds may have been caused in part by a lack of power at push-off. In control subjects, a burst of power from the plantarflexors at push-off propels the body forward and upward [16]. In our subjects, the brace generates no net power. The flexor reflex gives only limited power, and only during swing. This lack of power at push-off would also explain the repeated small pushes we observed to move the walker and body forward. Lacking plantarflexor power, the subjects had to use the arms to move the body forward and upward at the conclusion of stance. Stimulating more muscles or using a powered orthosis may provide better results. Time spent balancing and positioning the trunk also contributed to low walking speeds. Casual observation suggested that subjects spent 5-7 s per stride in this activity.

Given that the brace eliminated muscular work during stance, and that our subjects spent the majority of the stride in double support, we might expect much lower heart rates. While the heart rates we obtained were somewhat lower than with the Parastep system, they were still high (128 b/min, about 47 b/min above resting). High heart rates have also been reported with the RGO II, which combines supportive bracing and FES, and with various bracing systems that don’t use FES [6]. We attribute these high rates to co-contraction and excessive work of the muscles of the upper trunk and arms to balance the otherwise unstable hips and trunk. Our observation that subjects supported 44% of their body weight with their arms suggests that the KAFO reduced the effort required to walk. This is presumably because it, rather than muscle, supported the body weight during stance.

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presence of FES, since they are not observed when paraplegics use an FES-powered wheelchair [14] or perform body-weight-supported FES-driven cycling or stepping [17]. They are not due to high rates of mechanical work, as kinetic energies in our subjects were very small. This explanation suggests that efforts to reduce the energy demands of paraplegic walking should focus on the arms and trunk and on the need for balance and support, rather than on the legs.

REFERENCES


