Myoelectric Control of an Implanted Neuroprosthesis to Restore Gait in Incomplete Spinal Cord Injury

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Abstract—Functional electrical stimulation can be used to restore gait after incomplete spinal cord injury but needs to be coordinated with the user's retained volitional control. This paper outlines the case study of an implanted neuroprosthesis comparing an open-loop, pre-set pattern of stimulation with use of intramuscular electromyogram (EMG) recordings to trigger changes in stimulation. The user could modulate walking speed with the EMG control between 0.20 and 0.48 m/s by altering the amount of time spent in double support, while velocity was maintained at 0.31 m/s without the EMG control. Further investigation is warranted, but this preliminary finding demonstrates that the EMG control should allow users to modulate speed to walk faster in open spaces or slow down to navigate confined areas.

I. INTRODUCTION

There are approximately 12,000 new spinal cord injuries (SCI) in the United States each year, the majority of which result in incomplete paralysis [1]. Implanted neuroprostheses in the lower extremities have been shown to be beneficial to many aspects of the SCI population [2, 3] by using functional electrical stimulation (FES) to apply electrical impulses to peripheral nerves and elicit muscle contractions. One of the main problems with using FES to control walking in incomplete spinal cord injury (iSCI) is the inability to trigger stimulation at the appropriate time accurately and reliably. Since many people with iSCI have some lower extremity volitional control, it could be useful to synchronize their volitional movement with the stimulation cycle, which can lead to more natural or comfortable gait than if the two remain separate. Foot switches or force sensitive resistors can be used as a triggering system. However, their reliability can vary based on the placement of the devices in the shoe, the type of footwear worn, the terrain walked, and the type and severity of impairment. In long-term use, foot switches have been shown to deform and malfunction from mechanical breakage of solder joints and sticking contacts [4]. These devices also require extra equipment that must be donned, which can be difficult for some patients with upper extremity impairments.

Using the EMG of muscles previously associated with walking that are still under voluntary control can be an accurate way to control stimulation to the paralyzed muscles of the lower extremities. Research has been done to integrate surface EMG activity to initiate electrical stimulation of the dorsiflexors to control footdrop in hemiplegia [5-8]. However, fully implanted EMG and FES systems for gait in iSCI are not very well explored. This single-subject case study explores an iSCI subject with an implanted walking assist neuroprosthesis and compares two control strategies for applying pre-determined patterns of FES to restore gait: one in which the left step and right step patterns cycle automatically after each other and one in which the subject can independently trigger initiation of both the left and right step patterns using retained volitional muscle contractions detected by the implanted myoelectric signal (MES) recording electrodes. Using the subject's own volitional muscles as a control source should permit a more dynamic gait pattern with less time spent in double support and allow the user to more naturally vary walking speed.

II. METHODS

A. Subject

This male subject sustained an incomplete, ASIA Impairment Scale D SCI at the C6 level due to a motorcycle accident at the age of 34. In accordance with protocols approved by the United States Food and Drug Administration and the Institutional Review Board of the Louis Stokes Cleveland Veterans Affairs Medical Center, consent was acquired and he was enrolled into the project. Nineteen months post-injury, he received an 8-channel implanted neuroprosthesis for restoration of gait [9]. After a period of study, the 8-channel implanted pulse generator was replaced with a 12-channel system capable of telemetering data from voluntary muscle contractions to the system's external control unit. Because the fully implemented system could record from only two channels, surface EMG signals from a variety of muscles were collected while walking with the 8-channel neuroprosthesis to determine which muscles to implant with recording electrodes. Since the subject's iSCI resulted in a greater degree of paralysis on his left side and superior retained volitional control on his right, the right medial gastrocnemius (RGS) and right vastus lateralis (RVL) signals were found to be the most repeatable and synchronized with the gait cycle so were chosen as the

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control sources. The final system consisted of a 12-channel implanted stimulator-telemeter (IST-12 [10]); intramuscular (IM) MES recording electrodes in RGS and RVL; IM stimulating electrodes in bilateral iliopsoas and gluteus maximus as well as left hamstring, gluteus medius, quadriceps, tensor fasciae latae, and posterior portion of adductor magnus; and four-contact spiral nerve cuff electrodes around bilateral fibular nerves to elicit ankle dorsiflexion (Figure 1). A Universal External Control Unit (UECU) received the processed EMG signals and supplied both power and commands to the implanted system via an inductive link established by a radio frequency transmitting/receiving coil taped to the skin over the IST-12 (Figure 2).

B. Algorithm Development

Over the course of the subject's rehabilitation and training to use the FES system, temporal patterns of pulse-width modulated stimulation were determined for each channel to optimize bilateral stepping motions and coordinate them with his retained volitional control. Stimulus pulses were constant-current and charged-balanced. A stimulation frequency of 16 Hz was found to be high enough to obtain fused muscle contractions and also low enough to avoid fatigue. The IM muscle-based electrodes were stimulated at 20 mA, and the spiral cuff nerve-based electrodes were stimulated at 2.1 mA.

First, gait with automatic cycling between left and right steps at a cadence the subject determined to be comfortable and functional (approximately 40 steps per minute) was considered (Figure 3). The only control he exerted over the system was to start and stop the cycle of the stepping patterns through a pushbutton-driven menu.

Data were then collected from the implanted EMG-recording electrodes while he was walking with the cyclic stimulation. The IST-12 circuitry recorded bipolar voluntary EMG signals after blanking the stimulation recharge waveform and M-wave. Signals were amplified (gain=8000), filtered (bandpass, 100-1000 Hz), rectified, integrated and digitized (12-bit analog-to-digital converter) over a 20 ms window during each stimulus period. The result was one EMG data point telemetered from the IST-12 to the UECU every 62 ms. The acquired EMG signals were then filtered (lowpass, 1 Hz cutoff frequency, 2nd order) in the UECU and recorded for analysis and controller development.

Data collected during several walking trials showed that the RGS signal exhibited a distinct and repeatable rise in magnitude before each right step (indicative of push off) and that the RVL signal demonstrated a distinct and repeatable rise in magnitude before each left step (indicative of terminal knee extension and acceptance of weight back on the right leg after swing). These events were subsequently utilized to switch between the same left and right stimulation patterns as described above for automatically cycling between steps.
A depiction of the algorithm can be seen in Figure 4 and is described here. Starting from double support with bilateral extensor muscles activated, the left step involved turning the left extensors off and flexors on to swing the left leg, then the left flexors off and extensors on to return to double support. Stimulation remained at the double-support levels following the left leg's swing (indicated by "..." in the figure) until the RGS signal exceeded a threshold of 200 units (which remained consistent between testing sessions on different days). When the RGS exceeded this threshold, which could occur immediately following the left swing or anytime afterwards when the subject chose to activate RGS to push off, stimulation proceeded to the right step pattern. After the right leg's swing, achieved by turning the right extensors off and flexors on for a brief interval, stimulation remained at the double-support levels (indicated by "..." in the figure) until the RVL signal exhibited a peak. At this point, which could occur immediately following the right swing or after some delay if the subject volitionally slowed his actions, stimulation returned to the left step pattern.

C. Experimental Protocol

The subject used the both the automatic- and EMG-triggered algorithms to alternate left and right steps while walking on a 5 m GAITRite Electronic Walkway (CIR Systems, Inc., Sparta NJ). The GAITRite collected spatiotemporal gait parameters using pressure sensors embedded in the walkway that detected the foot contact with that mat. Data were collected with the GAITRite System while attempting to walk at both slow and fast speeds with the EMG controller. Two trials at each speed with the EMG triggering algorithm and one trial with the automatic triggering pattern were collected and analyzed.

III. RESULTS

A. EMG Recordings

The EMG signals from the implanted muscles were collected in the UECU and used as described by the algorithm depicted in Figure 4. They were also available to be exported to a lab computer for consideration. From an example section of collected EMG data with respect to time (Figure 5), we can observe the following:

a. When the left step stimulation pattern began, no EMG was monitored while completing the swing portion of the stimulation pattern.

b. After left swing, the program started monitoring RGS EMG (black line) and held stimulation at the end of the left step pattern until the subject voluntarily caused the RGS EMG signal to exceed threshold by pushing off during preswing.

c. At that point, the right step stimulation pattern began to assist his volitional movements, and no EMG was monitored while completing the swing portion of the stimulation pattern.

d. After right swing, the program started monitoring RVL EMG (gray line) and held stimulation at the end of the right step pattern until the subject voluntarily caused the RVL EMG value to peak by accepting weight onto the limb.

e. At that point, one full left and right step were completed and the cycle repeated back to the beginning of the left step stimulation pattern. As in (a) above, no EMG was monitored while completing the swing portion of the stimulation pattern.

f. After left swing, the program started monitoring RGS EMG (black line) in preparation to trigger the next right step as in (b) and (c) above. However in the example from which these data were recorded, RGS did not exceed threshold due to the subject's desire to cease stepping, so stimulation was maintained at the double support phase of the end of the left step stimulation pattern. In a more prolonged walking trial, the plot would repeat similar to (a-d) above for many iterations of left and right steps.

B. GAITRite Results

The spatiotemporal elements of the GAITRite data showed the subject could go slower with the EMG-triggered control algorithm ($0.20 \pm 0.02$ m/s, 25 steps recorded) than
with the automatic-triggered control algorithm (0.31 ± 0.00 m/s, 9 steps recorded) and also faster with the EMG-triggered control algorithm (0.48 ± 0.03 m/s, 8 steps recorded) (Figure 6). This speed modulation was due to the subject having direct control over when the next step was initiated, as seen by the time spent in double support. The subject spent the least amount of time in double support during the EMG fast trials (1.03 ± 0.08 s) and the most time in double support during the EMG slow trials (1.84 ± 0.06 s), while the automatic triggering double support time assumed an intermediate value (1.64 ± 0.01 s, Figure 6). For both velocity and double support time, all differences (between slow and auto, auto and fast, slow and fast) were statistically significant (p < 0.01).

IV. CONCLUSION

In this small sample, a walking algorithm based on the EMG of an iSCI subject's volitional control of lower extremities was developed and tested. The subject was able to effectively vary his speed when using EMG-based control, walking faster and slower than his pre-programmed automatic cycling pattern by varying his double support time.

While this EMG-based control has been shown to allow the subject to modulate his speed, this feasibility study had a limited number of trials and further investigation needs to be performed. A motion capture system that can allow for quantitative kinematic and kinetic measurements, in addition to simple spatiotemporal measures, will be employed in the future to further elucidate the mechanisms and other potential benefits of volitional control.

Having stimulation coordinated with the volitional activities of muscles associated with the stepping movement can allow for a better controlled gait that is more natural and intuitive for the subjects. The ability to modulate speed is important when a subject wishes to walk faster in the community or needs to slow down in confined spaces such as their homes or large crowds. This may lead to a better acceptance of the system which will allow for improving community ambulation and providing the physiological benefits of gait.

REFERENCES