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Ambulation after Incomplete Spinal Cord Injury with Electromyogram-triggered Functional Electrical Stimulation

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Abstract— **Ambulation after spinal cord injury is possible with the aid of functional electrical stimulation (FES). Individuals with incomplete spinal cord injury (iSCI) retain partial volitional control of muscles below the level of injury, necessitating careful integration of FES with intact voluntary motor function for efficient walking. In this study, the surface electromyogram (sEMG) of the volitionally controlled Erector Spinae was used to detect the intent to step and trigger FES-assisted walking in a volunteer with iSCI via 8-channel implanted stimulation system. The inference system was able to trigger the FES-assisted swing-phase of gait with a false positive rate of 1% during over ground ambulation on a level surface. The performance of the sEMG inference system highlights its potential as a natural command interface to better coordinate stimulated and volitional muscle activities than conventional manual switches and facilitate FES-assisted community ambulation.**

I. Introduction

FUNCTIONAL electrical stimulation (FES) provides an opportunity for brace-free ambulation to wheelchair dependent individuals with incomplete spinal cord injuries (iSCI). Pacemaker-like implanted FES systems can electrically activate a customized set of muscles selected to address individual gait deficits with pre-programmed patterns of stimulation to produce cyclic movement of the lower extremities for ambulation [1]. Implant recipients normally use a ring-mounted thumb switch connected to the external control unit (ECU) to manually trigger each step and progress through the customized pattern of stimulation to achieve walking function. Some individuals with limited finger and hand function find it difficult to actuate thumb switch, more so while trying to maintain balance during ambulation. This particular function of the thumb-switch can be replaced by a gait event detector.

Our preliminary results indicate that the EMG from partially paralyzed muscles can provide significant information related to the volitional activity of the muscle and can be used for gait-event detection [2]. This paper presents a novel inference system for foot-off intent detection using EMG from two partially paralyzed muscles below the level of injury. The inference system detects the intent to initiate swing-phase and integrates that information in the FES controller to trigger FES-assisted swing of the limb.

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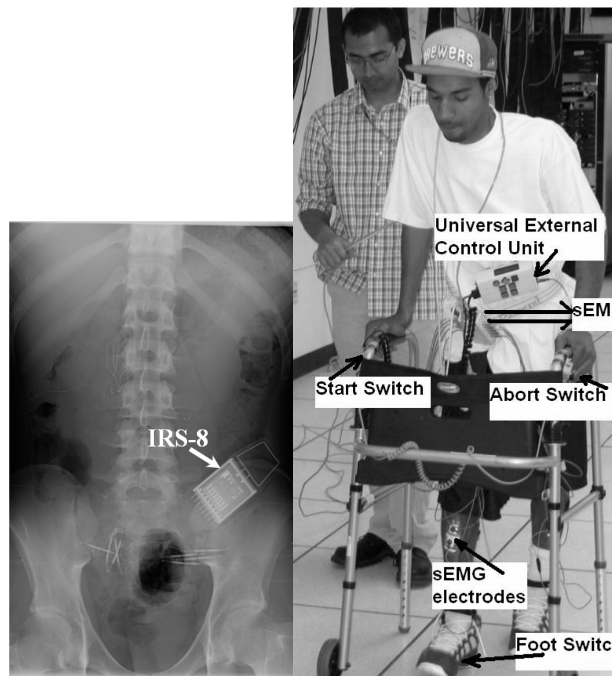

Fig. 1. Experimental setup for testing sEMG-triggered FES-assisted walking

II. Methods

A. Subject

The subject was a 23 years old male with C7 motor and C6 sensory incomplete spinal cord injury (ASIA C) who could stand but could not initiate a step without the assistance of FES. He received an Implantable Receiver Stimulator (IRS-8) and eight surgically implanted intramuscular electrodes in a related study designed to facilitate household and community ambulation with the implanted neuroprosthesis [3]. Temporal patterns of stimulation to activate the muscles were customized for his particular gait deficits according to our established tuning procedures [4] in order to achieve forward stepping in a rolling walker. The limited range of motion of his fingers made it difficult for the subject to operate the standard manual ring-mounted thumb switch. After discharge from rehabilitation as an independent community ambulator with the neuroprosthesis, the subject volunteered for studies related to the EMG control of the FES system.

Informed consent was obtained and all study related procedures were approved by the Institutional Review Board of the Louis Stokes Cleveland Department of Veterans Affairs Medical Center.

B. Experimental Setup and Data Processing

The experimental set-up for evaluating surface EMG-triggered FES-assisted walking is shown in Figure 1. Surface EMG (sEMG) signals were collected from lower body musculature during manually-triggered FES-assisted gait according to SENIAM guidelines using Ag/AgCl electrodes with 2 cm. inter-electrode distance [5]. Retro-reflective markers were placed on the body segments according to the 'plug-in' gait marker set in the Vicon Workstation^{TM} (Vicon Peak, USA) software to acquire lower-body kinematics data using a seven camera Vicon^{TM} (Vicon Peak, USA) motion capture system. Gait events (foot-strike and foot-off) were derived from foot-floor contact patterns obtained from insole-mounted foot switches placed bilaterally at medial and lateral heel, first and fifth metatarsal, and big toe, and confirmed with the kinematics data acquired.

The subject was asked to make multiple passes across an 8m straight level walkway. Each pass constituted one trial and multiple trials were collected during a session. The sEMG data collection was evenly spread over multiple sessions over a month to capture the day-to-day variability. 150 steps (each side) were captured over multiple sessions during this period.

Stimulation artifact and M-wave were prevented by blanking for 30ms following the start of the stimulation pulse. The blanked portion of the sEMG was reconstructed with the average value of the sEMG in the pre- and post-blocks [6]. The rectified sEMG pattern was low pass filtered (5th order zero-lag Butterworth, f_{cutoff}=3 Hz) to get the linear envelope of each signal. The sEMG linear envelopes during a gait cycle were then classified into double-support and swing phases of left and the right side based on the occurrence of foot-strike and foot-off as determined from the insole foot switch data.

C. Development of the EMG Inference System

The gait was divided into two tasks, the swing-phase and the stance-phase of each side. Task-specific features were identified from the sEMG patterns that were used by the inference system to identify the task (i.e. swing-phase or stance-phase). The schematic of the EMG inference system with the FES-controller is shown in Figure 2. The processed linear envelopes (LE) for each muscle pair (bilateral muscle set) were divided into two classes: the class 'True' was comprised of LEs during double-support prior to foot-off (Task1=swing phase) and the class 'False' consisted of the LE during terminal stance (Task2=stance phase). Half of the data were allocated to training and used to find a characteristic pattern of activation for each class by ensemble averaging the LE. The characteristic pattern found for class 'True' was cross-correlated with the LE patterns for classes 'True' and 'False' from the other half of the data (test data).

A normal distribution function was estimated for each class (correlation coefficient) using minimum variance unbiased estimator (Matlab^{TM} 'normfit' function). A Receiver Operating Characteristics (ROC) curve was computed using the estimated normal distribution functions for the classes, 'True' and 'False'. A statistic, the Discriminability Index (DI), was computed from the area under the ROC curve (maximum=1). The bilateral Erector Spinae (ES) exhibited the highest DI (=0.875) and were selected as the command sources for the inference system.

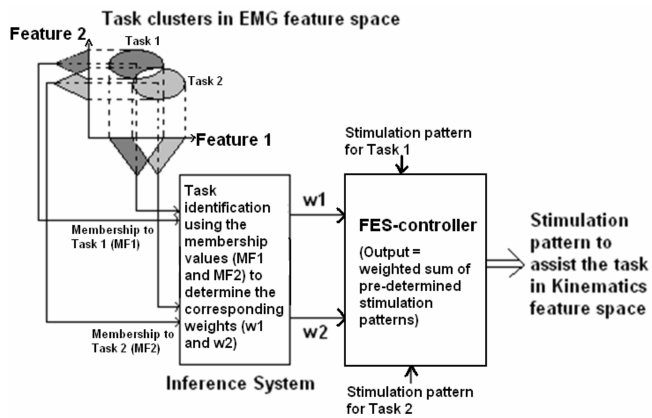

Fig. 2. The task is identified from the task-specific clusters in the EMG feature space and then assisted by task-specific stimulation patterns to provide the coordination in kinematics feature space.

The LE from each ES for both the classes ('True' and 'False') were pooled together and divided into orthogonal basis functions (principal components computed from the covariance matrix) by Singular Value Decomposition. Four principal components (PC) accounted for more than 90% of the variance in the data. The orthogonal factor rotation (MATLAB^{TM} 'varimax') was applied to minimize the number of factors and increase the loading on fewer factors for the classes 'True' and 'False'. The loadings on the factors after the 'varimax' rotation created separate clusters of points for the classes in the feature space (Figure 3). The loadings found for each LE were normalized by the square-root of the sum of the squared loadings to define a unit vector from the origin to each point. The mean of all the points in the cluster was computed for each class and called the centroid of that class.

The inference system estimated the factor loadings from the correlation between the factors and the mean-adjusted LE pattern. The Euclidean distance between the normalized loading and the centroid of the case 'True' was the membership value (MF1) for that task (Task1=swing phase). A threshold was defined for this membership value from the test data based on the ROC plot such that the false positive rate was below a reasonable value of 2%. The rule used to determine the weights, 'w1' (for stimulation pattern of the swing phase) and 'w2' (for stimulation pattern of the stance phase) was; “*If (MF1 <= threshold), then (w1=1 & w2=0) else (w1=0 & w2=1)*”.

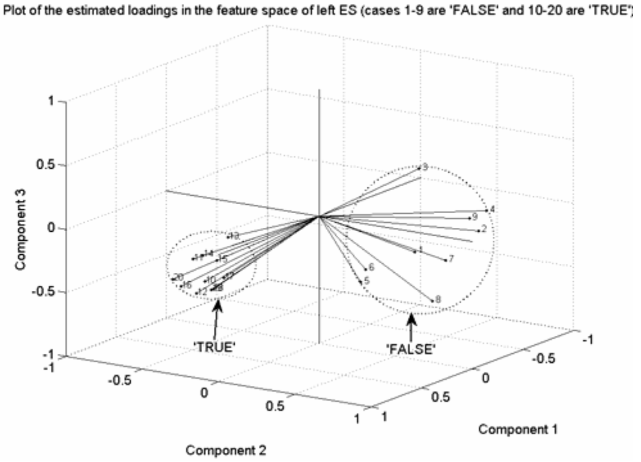

Fig. 3. Example of task-specific clusters for the classes ‘True’ (Task1=swing phase) and ‘False’ (Task2=stance phase) in the feature space

D. *Integrating Inference System with the FES-controller*

The inference system was implemented in Simulink™ (The MathWorks, Inc., USA) and incorporated into the FES-control system. The ECU was controlled in real-time by a dedicated personal computer running xPC-target™ (The MathWorks, Inc., USA) that executed the control algorithm to deliver pre-programmed stimulation patterns to the muscles via the IRS-8. The inference system started scanning the LE from ES after the end of the stimulation pattern (swing-phase) of the contralateral limb to detect the intent (foot-off) to initiate swing of the ipsilateral limb. The change of the weight, ‘w1’, from 0 to 1 triggered the stimulation pattern for the swing phase. The stance phase was completely volitional (no FES assistance). The off-line timing analysis indicated that the trigger from the inference system preceded actual foot-off thus predicting both left and right swing phase.

For safety, the inference system was allowed to trigger the FES-assisted swing-phase of the ipsilateral limb 1.5 sec after the end of the swing-phase of the contralateral limb. The value 1.5 sec, as the minimum duration of the double-support phase was selected by the subject to maintain balance.

E. *Testing EMG-triggered FES-controller for walking*

During testing, the subject was first asked to walk a few steps using the manual switch-triggered FES system to find the maximum muscle activity in ES, which was used to normalize the sEMG for the remaining trials of the session. The subject was then asked to walk across a straight walkway with the sEMG-triggered FES-controller. The subject triggered the first step manually from ‘stand’ state using a start switch after which the subsequent steps were triggered by the sEMG inference system. The FES controller returned to the ‘stand’ state when the inference system failed to trigger the next step within 3 seconds of entering the double support phase. The subject used a switch to manually stop the FES controller if it triggered a step when none was intended (false positive). If the FES controller stopped and failed to trigger an intended step (false negative) then the subject was also able to over ride the controller to manually trigger the step, after which the sEMG inference system would resume operation. At the completion of each trial, the subject stopped at the end of the 8m straight walkway and waited for 3 sec. to return the FES controller to the ‘stand’ state. The performance of the inference system was evaluated over 60 steps (each side) on level ground in a single session (1 day) of data collection.

To evaluate the repeatability of system performance, randomized trials of sEMG-triggered and switch-triggered FES-systems were performed over 2 more sessions distributed over 2 additional days of testing. The sEMG-triggered and switch-triggered FES-assisted gait were compared based on gait-speed, step-length (left and right side), step period (left and right side) and duration of gait phases (left and right double support, left and right swing phase). Side-to-side similarity index (SI) was calculated for a stride, which is modified from Dingwell *et al.* [7]:

$$SI = 1 - \frac{\left| X_{Left} - X_{Right} \right|}{\left(X_{Left} + X_{Right} \right)}$$

where X is the value of the gait parameter of a side.

Estimated Floquet multipliers can show the orbital stability of the gait dynamics by looking at the stride-to-stride error propagation at a certain gait-event with respect to an estimated steady-state periodic gait cycle. The Floquet multipliers were estimated from the kinematics data and the α-measure of dynamic stability (lower value means more stable) was computed at foot off and foot strike, according to Hurmuzlu *et al.* [8].

III. Results

A. *Performance of the Inference System*

The inference system successfully minimized the possibility of triggering a step unexpectedly. During the first session of testing, the false positive rate was limited to 1.66% while the true positive rate was close to 80%. The one false positive observed for the right leg over 60 steps was during terminal stance. Repeatability of system performance was assessed with data collected during the two additional sessions in which switch- and sEMG-triggered controllers were presented randomly to the subject. No false positives were reported by the subject during more than 50 steps taken at these 2 sessions. When combined with data from the initial testing day, the system exhibited overall false positive rate of less than 1% and true positive rate of 82% for left foot-off and 83% for right foot-off over total of 110 steps taken (total 3 sessions over 3 days).

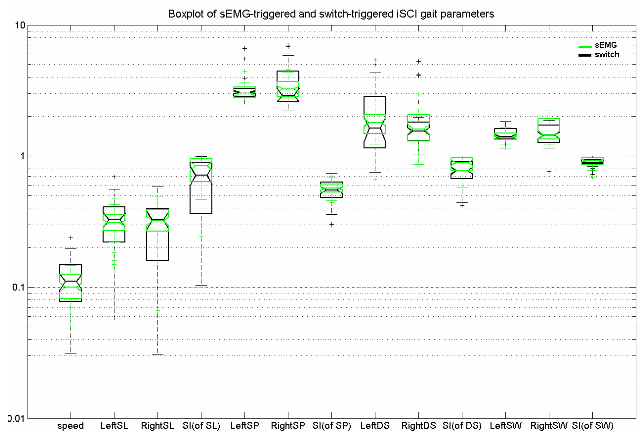

Fig. 4. Boxplot of gait parameters for switch- and sEMG-triggered iSCI gait (SL: step length (m), SI: similarity index, SP: step period (sec), DS: duration of double support phase (sec), SW: duration of swing phase

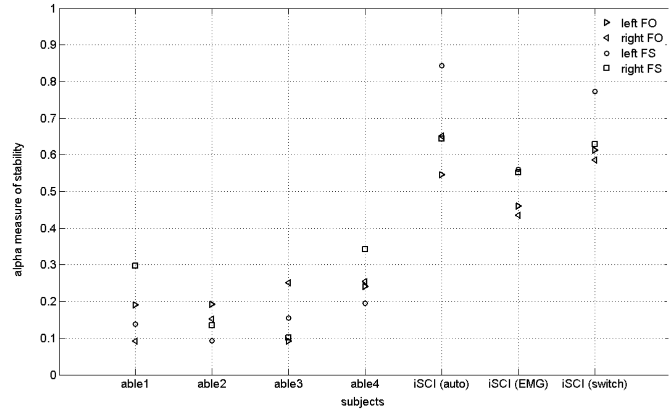

Fig. 5. α-measure of stability for able-bodied and FES-assisted iSCI gait (FO: foot off, FS: foot strike)

B. *Performance of the sEMG-triggered FES-system*

Basic spatio-temporal gait parameters such as speed, step length (SL), step period (SP), and duration of double-support (DS) and swing phase (SW) with the similarity index (SI) for switch and sEMG-triggered walking with the neuroprosthesis are shown in Figure 4. The gait speed was close to 0.1 m/sec for both sEMG-triggered and switch-triggered iSCI gait. The duration of the double-support and swing phase was similar irrespective of the command source. Similarity indices in all the cases were closer to 1 (more symmetric) for sEMG-triggered gait as compared to switch-triggered gait. Although differences were not significant, sEMG control appeared to be more consistent with parameters exhibiting less variability than during switch-triggered walking. The sEMG-triggered iSCI gait was found to be more stable than switch-triggered gait based on the α-measure of stability (Floquet multipliers) as shown in Figure 5.

IV. Discussion

The sEMG-triggered FES-assisted ambulation was successfully implemented and evaluated in the laboratory. A sEMG-triggered implanted FES system reduces the need to manipulate manual switches which actuate each step with current FES systems, or make special adaptations to them for the specific motor deficits of individuals with cervical level injuries. Proximal muscles were found to be more suitable as command sources than the distal muscles during muscle selection. Plans are currently being formulated to test the novel inference system on multiple subjects to determine its generalizability and the potential effects of inter-subject variability on system performance.

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