FES Gait Re-education: The Swing Phase Estimation

Imre Cikajlo Dipl.Ing., Tadej Bajd D.Sc.

Faculty of Electrical Engineering, University of Ljubljana

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Corresponding author:

Imre Cikajlo
Faculty of Electrical Engineering
University of Ljubljana
Tržaška 25
1000 Ljubljana
Slovenija

Tel.: +386(1) 4768 485
Fax: +386(1) 4768 239
E-mail: imre.cikajlo@robo.fe.uni-lj.si

ABSTRACT

This paper presents the use of multiple sensors for walking assessment and provision of cognitive feedback during early re-education of incomplete spinal cord injured (SCI) humans. The paper is focused on the swing phase estimation as an important part of the Functional Electrical Stimulation (FES) gait re-education system for incomplete spinal cord injured persons.

The proposed sensory system comprised four accelerometers, one gyro placed at the shank of the paretic leg, and two goniometers placed at the knee and ankle joints respectively. The data from sensors were input in the mathematical algorithm applied for swing quality estimation. The output from the algorithm was a numerical value. The calculated output was divided to three levels each defining the swing quality in terms of good, sufficient, and poor. This information was provided to the patient as an auditory signal. The patient was taught to maximize his efforts to improve the quality of walking, that is, move the more affected leg in a way that would generate the auditory output corresponding to the level »good«.

The preliminary measurements were performed in healthy subjects walking on even terrain and in an incomplete spinal cord injured person with C6 lesion during walking on the treadmill. FES was in the latter case triggered manually by a physiotherapist.

The results showed that the timing of FES triggering plays important role in sensory supported FES assisted walking, that is, the auditory feedback was also a cue to the therapist controlling the FES. The swing quality estimation enabled patients to voluntarily improve their walking; thus, consequently the intensity of FES assistance was decreased. This suggests that the use of FES, multisensor system for cognitive feedback are efficient rehabilitative method in early stage of rehabilitation of walking.

INTRODUCTION

In the recent years our research studies have been focused on incomplete SCI patients. In our earlier studies we realized the necessity of FES gait training in the early period after spinal cord injury [1]. The candidates were patients with upper motor neuron lesion, in more clinical terms the patients with thoracic or cervical lesion to the spinal cord. Since they were incomplete SCI patients, their movements were disturbed or they were not able to dorsiflex ankle. Among other motor deficits were also insufficient knee and hip flexion. Only a few of incomplete SCI patients were found candidates for permanent FES, most of them used FES only for a period of about one year after being released from the rehabilitation center [2]. In these patients peroneal nerve stimulation was found useful to provoke flexion response resulting in the swing phase of walking.

Several rehabilitative systems employing peroneal nerve stimulation used sensory information in order to trigger FES during walking. The sensory information was usually provided by use of simple artificial sensors. Data collected by a pair of miniature accelerometers were used to distinguish between the stance and swing phase [3]. Automatic detection algorithms were used to identify the appropriate phase of walking and to control FES. On the basis of the results obtained, development of a small implantable sensor-stimulator rehabilitation device was proposed.

The aim of a FES rehabilitative system for re-education of walking is not only to deliver electrical stimulation to the paralyzed muscles, but also to assess the sensory information from the paralyzed limb. The sensory information is provided to the patient and not only to the stimulator control unit. The FES rehabilitation systems for re-education of walking are intended to be used soon after the accident or onset of disease [2]. These systems are to be used within the rehabilitation centers and applied by therapists. Surface electrical stimulation is therefore appropriate. We are developing two separate systems intended for swing or stance

phase re-education. The adequate approach should be selected according to the patient's gait deficits. In this paper we are proposing swing phase quality estimation. In incomplete SCI persons, where usually only one side is affected, we aim to achieve symmetry of right and left leg swinging. Therefore, the gait re-education must be focused on making the swing phase of the affected leg similar to the non-affected swing phase. We are proposing a FES gait re-education system based on multisensor approach, where simple feedback signal is delivered to the patient. The feedback signal can be delivered to the patient through vibrotactile or electrical stimulation or by use of a small earphone. The cognitive feedback (CF) signal represents the degree of success of performing the swing phase. The patient can have also the possibility to voluntarily control the amplitude of FES by a control lever [5] in the handle of the crutch in order to improve swinging of the paralyzed lower extremity. The proposed CF could help the patient and the physiotherapist to accompany the estimated swing quality. This allows the patient to make a better voluntary move within his capabilities and the physiotherapist to hit the right instant of the FES triggering. Besides that the physiotherapist could focus on the duration of stimuli and as will be presented had a great impact on the swing quality coefficient. In previous work [2] we emphasized the FES triggering to be done by the patient himself, in this study we present the task of the physiotherapist.

METHODS

The method that we propose uses the swing phase of the less affected leg as a reference for the more affected leg during the re-education of walking. The goal is to achieve symmetrical swing phases in SCI humans with incomplete lesion. The reference measurement starts from the less affected side, and the results are considered as a reference swing phase. During the training or preliminary measurement the recorded reference was compared to the actual swing

movement. The assessment was done by a multisensor device [6][7] consisting of two pairs of single axial accelerometers (ACCESS, Switzerland), a single axial gyroscope (Murata ENC 03JA) placed at the shank and two goniometers (Penny & Gilles) placed at the knee and ankle joints respectively, using 100Hz sampling frequency. All four accelerometers were used to determine the radial and tangential ankle joint acceleration [6][7]. The ankle joint acceleration was determined by using two pairs of mounted accelerometers. A tangentially mounted pair of accelerometers was used to determine the tangential component of the ankle joint acceleration and the radial component was determined the same way. The absolute value of the ankle joint acceleration $|a_0|$ was calculated from tangential and radial component and applied in all calculations. Most of the software was implemented on personal computer (PC) using MathWorks Matlab, Simulink software.

The training of walking was considered to take place on a treadmill. The first stage of every measurement or training was to determine the reference swing phase. It was possible to select a swing phase pattern typical for able-bodied person, but then the goal seemed to become unreachable. We suggest that at least in the beginning the pattern of the less affected leg should be used as a reference for symmetry. The multisensor device was attached to the less affected leg and assessed data during several walking cycles on the treadmill. The time-course of the assessed ankle joint acceleration in one swing phase was then used as a reference for training or only comparison with the more affected leg during walking on the treadmill.

In order to distinguish between the swing and the stance phase during walking detection of the swing phase onset detection was needed. The algorithm is based on data assessed by a gyroscope. When a lower extremity comes into the swing phase, then a significant change of the angular velocity of the shank would occur. This change was detected, and assigned as a beginning of the swing phase. The end of the swing phase was detected by using the same

presumption. During the swing phase the ankle joint acceleration (a_0) and knee joint angle time-courses were recorded and stored into a buffer.

The swing phase estimation algorithm is based on two inputs, a flag and two input references. When the swing phase takes place, a flag is set to a positive value enabling the two inputs, which store the data from accelerometers and knee joint goniometer into a buffer. At the end of the swing phase the flag is set to zero and data assessment is thus finished. One of the input references is a user set demand for minimum knee flexion during swing phase. Therefore the maximum value of the buffered knee joint angle time-course is searched. The maximum value presents the knee joint flexion in a pre-swing phase. When the user demand is fulfilled the swing phase estimation algorithm proceeds, otherwise the session is finished, and the swing is marked as poor with a corresponding zero output. The second reference is the stored prerecorded acceleration time-course. Since the goal is to provide qualitative swing estimation, a correlation with the actual assessed ankle joint acceleration is calculated at the end of the swing phase:

$$\varphi_{meas,ref}(kT,\tau) = \frac{1}{n}\sum_{k=1}^{n} a_{0meas}(kT)a_{0ref}(kT+\tau) = \mathrm{E}\left[a_{0meas}(kT)a_{0ref}(kT+\tau)\right] \tag{1}$$

The next phase is calculation of the correlation coefficient:

$$\rho_{measref} = \frac{\mathrm{E}\left[\left(a_{0meas}(kT)-m_{meas}(kT)\right)\cdot\left(a_{0ref}(kT+\tau)-m_{ref}(kT+\tau)\right)\right]}{\sqrt{\mathrm{E}\left[\left(a_{0meas}(kT)-m_{meas}(kT)\right)^2\cdot\left(a_{0ref}(kT+\tau)-m_{ref}(kT+\tau)\right)^2\right]}} \tag{2}$$

where m represents the mean of the signal, n number of samples, a_{0ref} the stored reference acceleration and a_{0meas} the assessed ankle joint acceleration. This algorithm is performed for every swing phase end, with the exception when a user demands for knee flexion are not fulfilled. The calculated correlation coefficient presents the signal similarity. At zero output or low correlation coefficient we have no matching between the actual swing and the desired swing. The higher is the coefficient, the better is the similarity and the actual swing quality is closer to the desired.

On the basis of the correlation coefficient we defined the cognitive feedback. The problem arose how to present the feedback information to the walking SCI subject. According to the previous work [8] the cognitive feedback should be simple and easy to perceive. The patient should focus on the walking, not on the feedback. Therefore, we defined only three levels for audio cognitive feedback. In a case of insufficient knee flexion or low correlation coefficient the swing was deemed as poor. When the coefficient was higher than 0.2, then a swing phase was considered sufficient, finally then the coefficient was above the value 0.6 the walking was rated as good. These criteria can be set manually for each individual patient. The coefficient limits are set according to the patient's deficits and therapeutic requirements. In this study we selected the auditory cognitive feedback. Personal computer provided a sound at three different frequencies that were delivered to the patient through loudspeakers.

In a case of incomplete SCI the assistance of FES was needed to perform the expected quality of the lower extremity movement. Here a single channel surface peroneal stimulation was used, triggered by a physiotherapist pressing the pushbutton. Since the physiotherapist could follow the CF and was able to visually accompany the patient walking, there were no specific mapping regarding the FES triggering. We relied upon physiotherapists' experience and CF, which supplemented the visual feedback and returned a valuable estimation.

RESULTS

In the preliminary measurements three healthy subjects and an incomplete SCI patient were involved. In the healthy subjects the developed algorithms for swing phase detection and swing phase estimation were first tested. The swing and stance phase were easily recognized during healthy persons walking (Fig.1 and Fig.2). Figure 1 presents the signals assessed from the gyroscope, the absolute ankle joint acceleration $|a_0|$ and the goniograms of both joints, knee and ankle, while figure 2 shows all the signals assessed together with the output of the

swing phase detection algorithm. The chart below presents the output of the swing phase estimation algorithm. It represents the correlation coefficient ρ that was calculated by Eq 1,2 where a_{0ref} was assessed in preliminary measurement from the right leg and a_{0meas} from the left affected leg. As expected the swing quality was high, since walking of the healthy subject was almost symmetrical.

The second part of measurements was carried out at the rehabilitation center. Patient with C6 lesion to the spinal cord was walking on a treadmill. The goal was to test the suitability of walking re-education based on gait symmetry [4] and at the same time to examine the adequacy of the auditory feedback. The treadmill speed was first set to 0.7 m/s and later decreased to 0.5 m/s. During the assessment of the reference acceleration signal from the less affected leg we did not use any FES. In further walking trials the peroneal surface stimulation was triggered by a physiotherapist using the hand pushbutton. Here, we present only the measurement results from the walking with FES, since those are relevant for swing quality estimation. Figure 3 presents the signals assessed from gyroscope, absolute ankle joint acceleration $|a_0|$ and both goniograms, together with FES sequence, and the correlation coefficient value representing the swing phase quality.

There was a significant difference between healthy person's and incomplete SCI patient's walking (Fig 4). The heel contact can be obviously noticed as peaks in the acceleration signal in figure 3. In general this had no influence on the results, since we instead of using the prerecorded ankle joint acceleration from a healthy extremity used the data from a less affected extremity of the incomplete SCI patient as a reference signal. The lower chart in figure 3 presents the correlation coefficients. Some double signals can be noticed in the record. They were consequence of misdetection of the end of the swing phase, as the patient's walking was very irregular. The algorithm [6] was able to diminish the misdetection, so the CF was not impeded and was provided only at the end of the swing.

DISCUSSION

The shortening of rehabilitation of humans after SCI is extremely important. Many recent studies suggest that this can be achieved by combining voluntary and externally augmented movement exercise, that is, by involving the patient into the process of re-education. The FES gait re-education is an excellent candidate for speeding up the rehabilitation. It allows a patient to relearn the phases of walking by using information gained by a multisensor system in form of a qualitative information about walking. This was allowed by adding the cognitive feedback. The patient was aware of his good or poorly performed swing. When the swing was poor the patient should voluntarily try to perform the better swing. At the same time the cognitive feedback helped the physiotherapist to find the appropriate instant of triggering and the duration of FES. After the physiotherapist had been aware of poor swing, she extended or shortened the duration and appropriately triggered the FES earlier or later, depending on swing quality results, subjective visual impression and experience. Without qualitative estimation we were not able to assure repeatability during FES training. The multisensor system and cognitive feedback resolve the problem of repeatability and applicability in clinical environment.

Results strongly suggest that the timing of FES triggering play the important role in swing phase. All the patients' efforts to improve the swing could be devastated by improper moment of triggering. Thereby the auditory feedback is affective also to a therapist who is controlling the FES system. In addition, in order to improve the repeatability the physiotherapists' task could be replaced by computer. We propose the modification of the present system by adding the automatic FES triggering and control of the stimulation intensity. In parallel, the patient's swing improving could consequently decrease the stimulation intensity if and when required.

ACKNOWLEDGEMENTS

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FIGURE CAPTIONS

Figure 1 Time-course of gyroscope signal, absolute value of ankle joint acceleration and ankle and knee joint goniograms during walking of a healthy person on even terrain.

Figure 2 The output of swing detection algorithm is presented together with signals assessed from gyro and goniometers to emphasize the duration of the swing phase. The swing phase denotes the moment of signals assessment. The appertaining value of the correlation coefficient during walking of a healthy person on even terrain is presented in the lower figure.

Figure 3 Time-course of gyroscope, absolute value of ankle joint acceleration and ankle and knee joint angles during FES assisted walking of C6 patient walking on treadmill. The output of swing detection algorithm and the appertaining value of the correlation coefficient while using FES is also presented. The FES was triggered by the physiotherapist pushing the pushbutton.

Figure 4 The incomplete C6 SCI patient during walking on treadmill. The more severely affected left lower extremity is in the initial swing phase.