Swing phase estimation in paralyzed persons walking

Imre Cikajlo* and Tadej Bajd

Abstract. In the paper we present the sensory system aimed for FES gait reeducation of incomplete spinal cord injured persons. The proposed system consists of four accelerometers, gyro and two goniometers placed along the shank of the paralyzed leg. The data assessed are input into mathematical algorithms estimating the swing phase quality. The estimation is based on swing phase detection and signal correlation and is used to determine the cognitive feedback signal. The feedback signal is delivered to the patient as an auditory signal.

The preliminary measurements were performed in three healthy and two incomplete spinal cord injured persons with C6 and C6-7 lesion during treadmill walking. FES was manually triggered by a physiotherapist. The results have shown that the proposed multisensory system was successful in gait quality estimation. Therefore, the use of FES multisensory system and cognitive feedback could be an efficient rehabilitative approach in gait reduction.

Keywords: Multisensory system, sensory integration, swing phase, walking, spinal cord injury, gait reeducation, functional electrical stimulation

1. Introduction

In recent years our research studies have been focused on incomplete spinal cord injured (SCI) patients. In our earlier studies we realized the necessity of functional electrical stimulation (FES) gait training in the early period after spinal cord injury [1]. The candidates were all patients with upper motor neuron lesion, in more clinical terms the patients with thoracic or cervical lesion to the spinal cord. Only a few of incomplete SCI patients were candidates for permanent use of FES, most of them used FES only during their stay in the rehabilitation center or soon after being released. In these patients peroneal nerve stimulation was found useful to provoke flexion response resulting in the swing phase of walking. Several existing systems employing peroneal nerve stimulation used sensory information to trigger FES during walking. The sensory information was usually provided by use of simple artificial sensors [2]. Data collected by a pair of miniature accelerometers were used to distinguish between the stance and swing phase. 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 device was proposed. Dai et al. [3] proposed an application of tilt sensors in FES and used previously commercially available functional electrical stimulator. Williamson and Andrews [4] presented a gait event detection using three uniaxial accelerometers mounted below the knee of the patient. The gait

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event detection was based on rule based detectors and adaptive logic networks to distinguish between stance and swing phases during the gait cycle and detect the transitions between five gait phases during walking. Kostov et al. [5] used the adaptive logic networks to control the functional electrical stimulator for foot drop correction. The sensory signals were assessed from ENG and a heel switch. All these sophisticated techniques for gait detection were used to trigger the electrical stimulation, but in practice most of the systems in the rehabilitation environment still use the foot switch [6].

The aim of a FES rehabilitative system for re-education of walking [7] 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 fed back to the patient and not to the stimulator control unit. The FES rehabilitation systems for re-education of walking are intended to be used with incomplete SCI persons soon after the accident or onset of disease. 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 for training of proper movements during swing and stance phase. The adequate approach is selected according to the patient's gait deficits. In this paper we are describing the swing phase quality estimation. The gait re-education system for swing phase detection and swing quality estimation is based on multisensor use, simple feedback signal, which is fed back to the patient, and FES. The feedback signal can be delivered to the patient through vibrotactile or electrical stimulation or by the help of an auditory signal. It represents the successfulness or unsuccessfulness of performing the swing phase. In the envisioned gait reeducation system the patient has the possibility to control voluntarily the intensity of FES by a manual control lever [8] in order to improve swinging of the paralyzed lower extremity.

The preliminary measurements, described in the presented paper, are aimed to determine what is an appropriate swinging of the lower extremity. In incomplete SCI, when usually one side is affected considerably more than the other, we want to achieve symmetry of the right and left leg swing. Therefore, the gait reeducation is focused on making the swinging movement of the affected leg similar to the less-affected leg.

2. Methods

2.1. Hardware description

We used the sensory system [9] employing two goniometers (Penny and Gilles), single-axial gyroscope (Murata ENC 03JA) and two pairs of single-axial accelerometers (ACCESS). Accelerometers were mounted on a small aluminum plate and mounted perpendicularly in pairs. The gyroscope was mounted on a board with analog bandpass filter and amplifier and placed in the middle of the plate (Fig. 1). The plate (dimensions 192×42 mm) was attached to the shank of the patient by the help of velcro straps. All sensory signals were low-pass filtered, using the third order Butterworth filter and 10 Hz cut-off frequency, besides gyroscope signal which was first high-pass filtered (0.05 Hz) in order to avoid the temperature drift. The personal computer (PC) with Pentium$^\circledR$ III 500Mhz and Burr-Brown acquisition board were used to assess the data. The data were sampled at 100Hz with the resolution of 12 bits. A computer controlled four channel electrical stimulator developed in our laboratory was used. Single channel surface peroneal nerve stimulation was delivered to the patient. The stimulation frequency was set to 20 Hz while pulse duration and stimulation amplitude were adjustable. The stimulator was controlled by PC via RS232 and was triggered by manual switch.
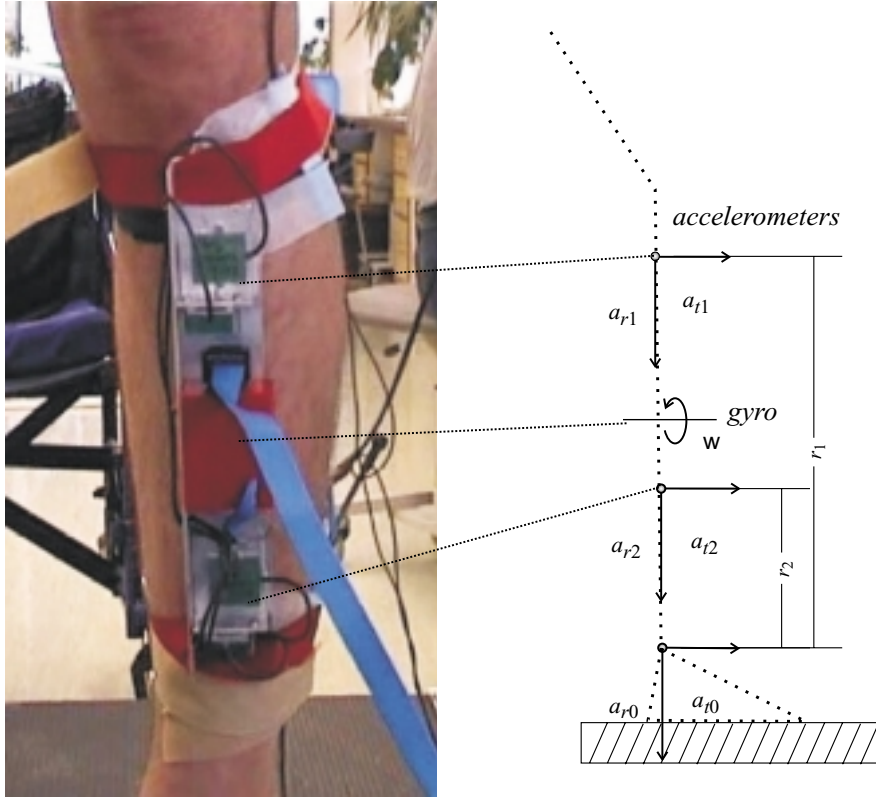
Fig. 1. Multisensor device attached to the shank. It consists of two pairs of single axial accelerometers and a gyroscope. Two goniometers are attached to the lower extremity to assess the ankle and knee joint angles. On the left the photograph of the frontal plane is shown, while the schematic presentation on the right shows the parameters measured in the sagital plane.

2.2. Swing phase estimation algorithm

Multisensor system was assigned several functional tasks. The gyroscope signal was used for swing phase detection, while the other sensors took part in swing quality estimation. The multisensor system was first placed on the shank of the less-affected extremity to assess the ankle joint acceleration during treadmill walking. The time-course of the acceleration assessed during the swing phase was used as a reference a_{ref} for the more affected extremity during the gait reeducation. The acceleration signals were measured at two points on the shank. At both points radial (a_{r1}, a_{r2}) and tangential (a_{t1}, a_{t2}) acceleration were measured. By use of a simple Eq. (1) we get the acceleration in the ankle joint [2]:

$$\begin{bmatrix} a_{t0} \\ a_{r0} \end{bmatrix} = \frac{1}{r_1 - r_2} \cdot \left(r_1 \cdot \begin{bmatrix} a_{t2} \\ a_{r2} \end{bmatrix} - r_2 \cdot \begin{bmatrix} a_{t1} \\ a_{r1} \end{bmatrix} \right) \tag{1}$$

In the Eq. (1) r_1 and r_2 present the distances from ankle joint to the place where particular pair of accelerometers is attached (Fig. 1).

Two algorithms are related to the signals of the multisensory system, swing quality estimation algorithm and swing phase detection algorithm. Inputs to the swing quality estimation algorithm are a pair of radial (a_{r0}) and tangential (a_{t0}) acceleration, ankle and knee joint angles (θ_1) and an output of the swing phase detection algorithm. The aim of the swing quality estimation algorithm is to provide reliable swing quality coefficient which determines the cognitive feedback signal.
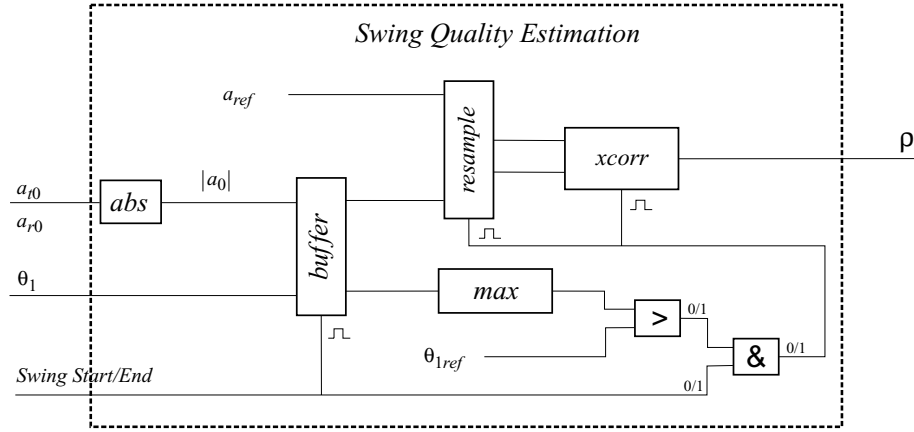
Fig. 2. Swing phase estimation diagram.

Swing phase detection algorithm represents the important part of the system, since it is our aim to estimate the time-course of the signals only during the swing phase. When a lower extremity enters into the swing phase, there occurs a significant change in the angular velocity of the shank. The angular velocity changes the sign from negative to positive crossing the zero what is in our system detected as the start of the swing phase. The end of the swing phase is detected with the same presumption. In a case of unstable knee [10] a false swing phase can be detected during the stance phase. Such a gait event is recorded, but not considered in further processing. The swing phase detection algorithm is the basic part of the swing reeducation algorithm and is input into every other algorithm.

In swing quality estimation (Fig. 2) the swing phase detection algorithm provides the start and the end of signal processing. The swing phase detection signal is set to 1 during the detected swing phase and thus enables the swing quality estimation algorithm. During the swing phase the absolute value of the ankle joint acceleration ($|a_0|$) and knee joint angle (θ_1) time-courses are recorded and stored in a buffer. At the moment, when the end of the swing phase is detected, the data assessment is completed. The algorithm examines the measured knee joint angle. If the maximum value of the knee joint angle exceeds θ_{1ref}, then a swing phase estimation algorithm is calculated, otherwise the swing phase estimation is rejected and the swing is marked as poor. The θ_{1ref} is defined on the basis of previous measurements, patients deficits and therapist requirements. Usually it was expressed in percent of the less-affected leg (80–90%) maximal knee flexion. In most cases the swing phase duration varies from the duration of the desired swing phase. Consequently, the number of samples in each time-course is different. Before making a comparison between the acceleration samples of both legs, a resampling is needed. The absolute value of the recorded ankle joint acceleration [9] is afterwards correlated with the reference acceleration time-course. The reference a_{ref} has been previously recorded on the less-affected extremity and represents the desired swing phase. The assessment cycle is repeated during every swing phase. The correlation is calculated on a basis of the following Eq. (2):

$$\varphi_{m,r}(t,\tau) = \frac{1}{n} \sum_{k=1}^{n} a_m(t) \cdot a_r(t+\tau) = E[a_m(t) \cdot a_r(t+\tau)] \tag{2}$$

where τ means a time delay and n is a number of samples included into computation. In the Eq. (2) a_m is the measured acceleration, while a_r is the reference acceleration time-course. Afterwards the correlation

coefficient is calculated:

$$\rho_{m,r} = \frac{E[(a_m(t) - m_m(t)) \cdot (a_r(t+\tau) - m_r(t+\tau))]}{\sqrt{E[(a_m(t) - m_m(t))^2 \cdot (a_r(t+\tau) - m_r(t+\tau))^2]}} \tag{3}$$

where m represents the mean of the signal. When the calculated coefficient $\rho_{m,r}$ is close to 0, there is no correlation between the signals and closer to 1 it gets, more signal resemblance exists. The correlation coefficient presents the criterion of the swing phase appropriateness (0 – poor, 1 – perfect).

2.3. Cognitive feedback

The cognitive feedback is defined on the basis of the correlation coefficient $\rho_{m,r}$. The feedback signal is divided into three discrete levels. In case of a low correlation coefficient the swing was deemed as poor. When the coefficient reached 0.2, swinging of the leg was considered sufficient and above 0.6 good. These criteria should be adapted to individual patient. The coefficient limits are set according to the patient's deficits and the requirements of the therapist. The feedback signal in this preliminary investigation was delivered as auditory cognitive feedback and was provided by PC as a sound of three different frequencies for sufficient, poor, sufficient and good.

3. Results

In the preliminary investigation three healthy subjects and two incomplete SCI patients were involved. All three healthy subjects and the patients with C6-7 and C6 lesion to the spinal cord, were walking on level terrain in the laboratory. The patient LG with C6 lesion (9 years after accident) was not an FES user. He was using crutches during walking. The other patient BM had a C6-7 lesion (18 years after accident) and was trained to use FES. FES assisted training of walking was performed on a treadmill. The treadmill speed was first set to 0.7 m/s and later decreased to 0.5 m/s. During walking we used the described multisensor measurement system [9] together with the sensory feedback algorithm running on PC, using Matlab$^{\circledR}$ software. The proposed hardware and software were first tested in healthy subjects. Figure 3 presents the signals assessed from gyro and goniometers together with the calculated absolute value of the ankle joint acceleration. The gyro waves correspond to the swing phase of walking. It is obvious that the acceleration during the stance phase is equal to the gravitational acceleration and that the maximum knee angle is reached during the the swing phase. The ankle goniogram shows dorsiflexion throughout the swing phase.

In Fig. 3 there was no error in detection and all swing phases were satisfactory. This was expected since walking of a healthy person is quite regular. In figure the swing phase detection is presented by rectangular pulses. During swing phase the signal is set to the value 100 otherwise it is 0. The swing phase signal corresponds to zero crossing of the gyro signal. The lower figure presents the correlation coefficient which was almost all the time above 0.6, representing thus a perfect swing.

The patient LG was walking on level terrain supported by crutches (Fig. 4). His walking pattern was irregular during the stance phase. Often patients with spinal cord injury avoid knee flexion in loading response of the stance phase because of quadriceps weakness. This can occur when most of the body weight is carried through the arm support provided by the crutches. Consequently, the gyro signal detected this event as swing phase. Furthermore, the algorithm marked it as a false detection and provided a negative feedback signal. It is shown in the lower chart (Fig. 4) as a negative value -1. As
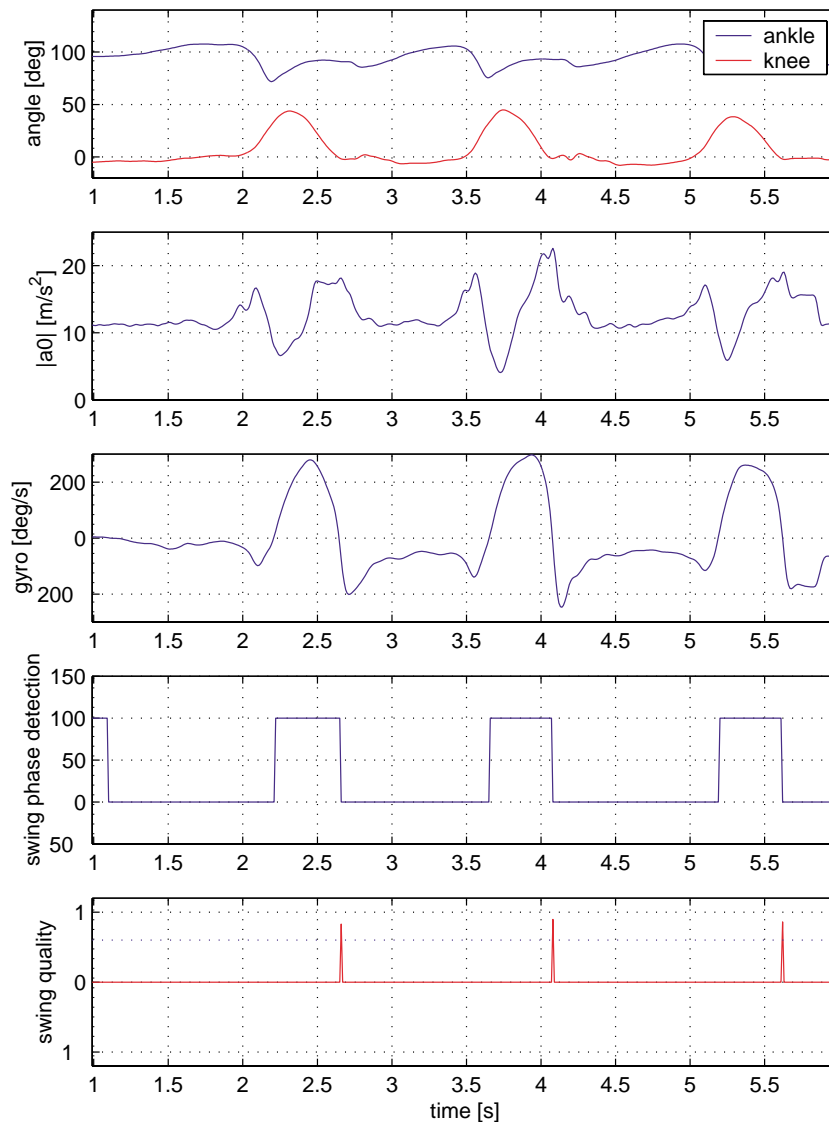
Fig. 3. Time-course of gyroscope, absolute value of ankle joint acceleration and ankle and knee joint angles during level walking of a healthy person. The two charts below present the swing phase detection and the swing quality coefficient with a threshold for good swing.

the patient was a trained crutch walker, the quality of his swing phase was quite good except the last swing was extremely poor. The unstable foot [10] contact caused a swing phase extension which poorly correlated with the reference acceleration.

Patient BM was not a regular FES user. In the rehabilitation center he was using a single channel MicroFES peroneal brace together with crutches, (Institut Jozef Stefan, Ljubljana). He was training FES walking for two months. We measured his walking on a treadmill with and without FES. The patient was asked to walk continuously for 60 s. After every trial he took a rest of 5 minutes. During FES assisted walking the physiotherapist was manually triggering the surface peroneal nerve stimulator according to the provided cognitive feedback. Here we present only the measurement with FES, since
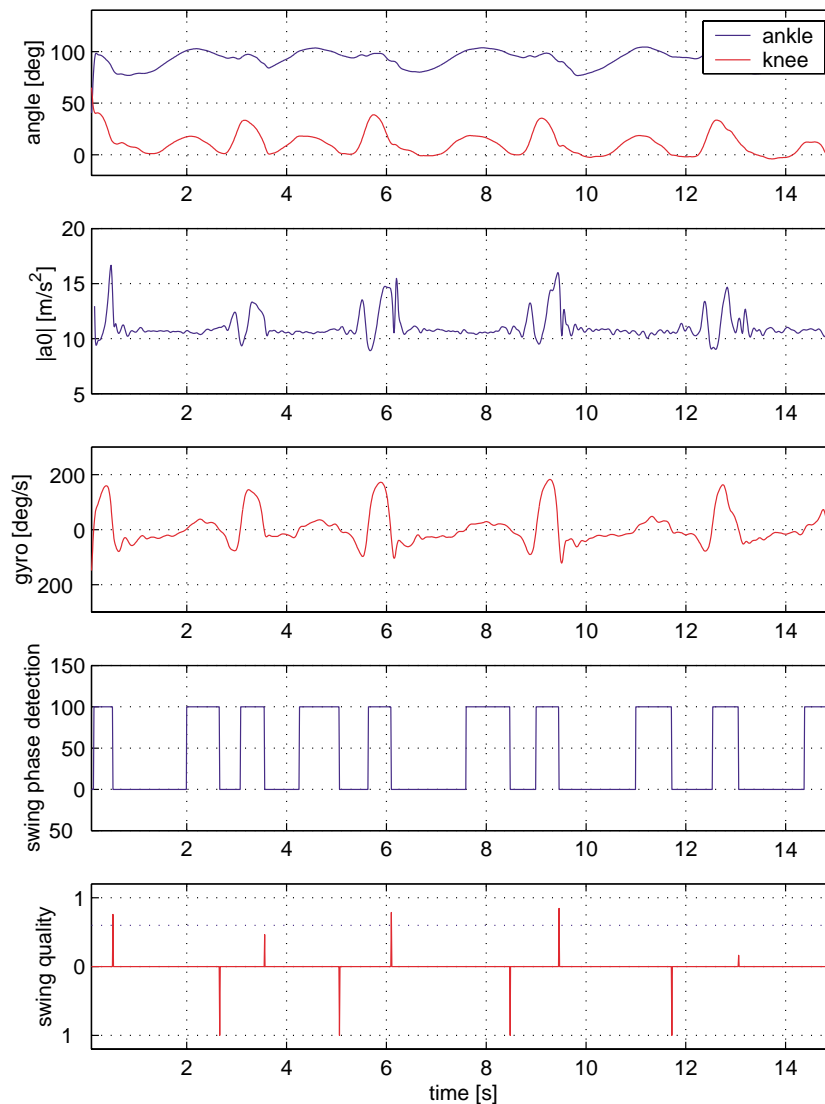
Fig. 4. Time-courses of ankle and knee joint angles, absolute value of ankle joint acceleration and gyroscope during level walking of patient LG using crutches are presented. Swing phase detection and swing quality are presented in two lower charts. False detections occurred in stance phase and were successfully eliminated. These errors are presented in the lowest chart by the negative value -1 of the swing quality indicator.

swing quality can be improved by selecting adequate start, duration, and intensity of train of electrical stimuli delivered to the peroneal nerve. The first trace of Fig. 5 presents the gyro signal which was used as the swing phase detector. The absolute value of the ankle joint acceleration $|a0|$ during the swing phase was selected as the main criterion for the swing quality estimation. During the stance phase it was equal to the gravitational acceleration, while during the swing phase it comprised noticeable waves and spikes which were used for determination of correlation coefficient. The joint angles in ankle and knee are also presented in Fig. 5. The knee goniogram shows obvious difference with regard to walking of a healthy person. The presence of FES is also shown in Fig. 5. In the last trace the correlation coefficient is presented. The swing quality is quite low, since patient BM was using FES for a short time only.
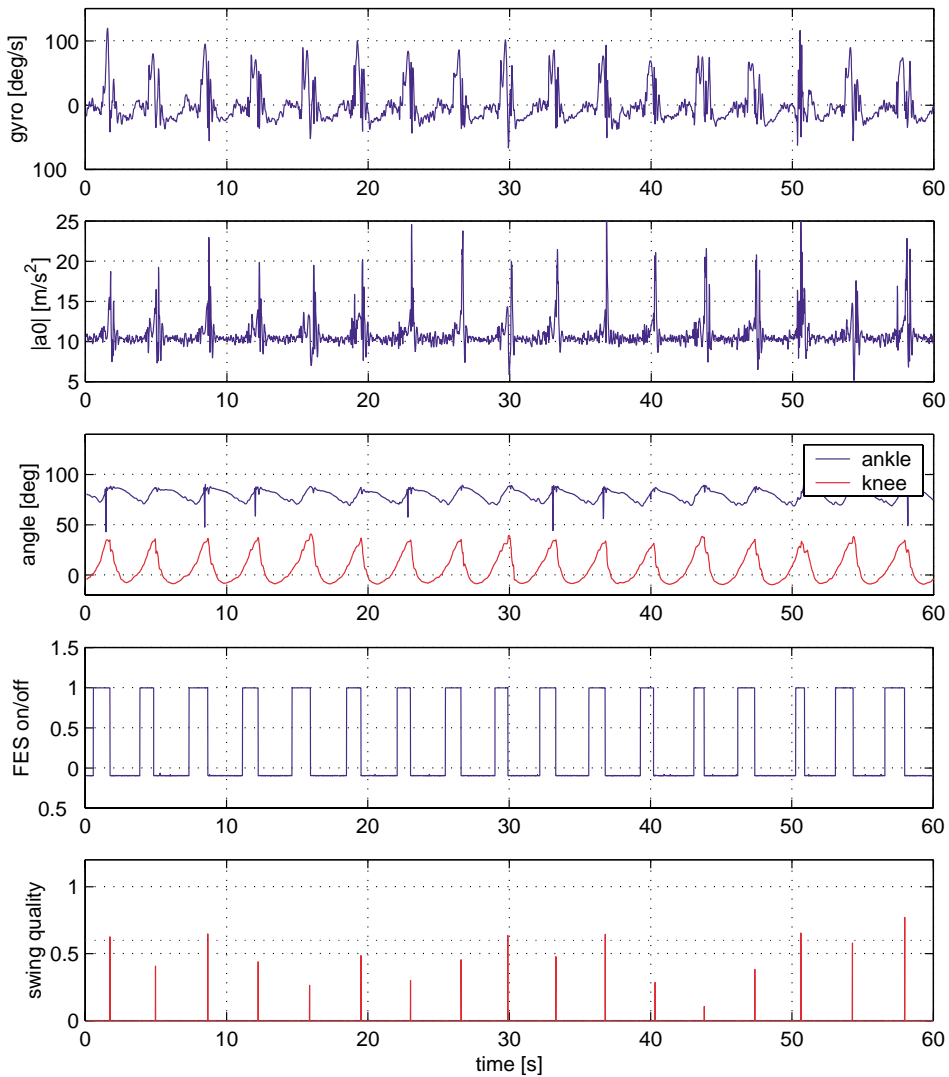
Fig. 5. Sensory signals and correlation coefficient during treadmill walking of patient BM. The cognitive audio feedback was provided according to the swing quality. FES was manually triggered by the physiotherapist.

The swing quality recorded in SCI person differs significantly from that assessed in healthy person. The less-affected extremity swing of the incomplete SCI patient was used as a reference. Cognitive audio feedback was based on correlation coefficient and served as a warning to the physiotherapist who was manually triggering FES by a pushbutton.

4. Conclusions

On the basis of the results obtained we can describe the swing quality estimation algorithm as a successful attempt of multisensor use in the rehabilitation environment. There were several difficulties encountered in the detection of the swing phase of a SCI person. All these false detections were filtered

out so that there was no influence on the feedback signal at the end of the swing phase. The presented preliminary measurements have shown the feasibility of the multisensor use in the gait reeducation. The proposed algorithm can help physiotherapists in the early stage of rehabilitation of SCI patients. In these preliminary experiments the physiotherapist triggered FES according to the auditory feedback. Further development will make possible the FES triggering by the patient. In future measurements we shall have a pushbutton mounted on the treadmill supporting frame to involve the patient directly into the rehabilitation process. Further development will make the patient possible to control the stimulation amplitude by replacing the pushbutton by a special trigger [8]. In this way the patient will be even more actively involved in the gait reeducation what is of utmost importance for the successfulness of the rehabilitative approach.

Acknowledgments

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