光素子を用いた頭部運動による障害者用制御命令入力装置に関する基礎的検討 －第２報：光通信部におけるパルス変調方式の検討－

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Fundamental study on control command input system by using head movement with photo devices for handicapped - Part 2: Pulse modulation method in the optical link -

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1　はじめに

運動機能障害者が機能的電気刺激 (FES) を使用して麻痺した運動機能を再建するには、使用者の意思をシステムに的確に伝達する装置が必要となる。四肢麻痺など重度の運動機能障害者が使用する場合、入力の候補として頭部運動が考えられる。我々は、作業空間内に受光素子を配置し、頭部運動によって所望の命令に対応する素子にレーザ光を照射することで命令を選択するシステムの研究を進めている (1)。

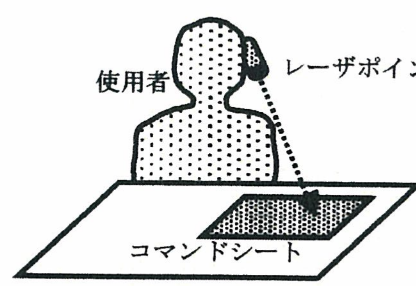
過去に提案した方式では、背景光検出素子を独立に設け、その出力電圧と、命令選択用素子の出力電圧との比較によってレーザ光照射の有無を判定していた (1)。その場合、装置が大きくなるにつれて背景光検出素子と命令検出素子とが離れ、背景光量に差が生ずる。そして、背景光検出素子が作業空間の光環境を正しく反映しなくなると、命令の選択ができない可能性がある。

そこで本研究では、背景光電圧との比較ではなく、レーザ光をパルス変調することで信号検出するシステムを試作し、その実現可能性を確認したので報告する。

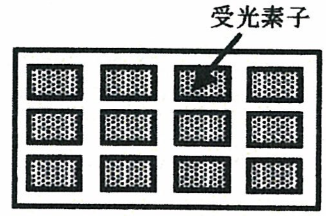
2　システムの概要

2.1　概念設計

システムの概念を図1に示す。本システムは、使用者の頭部（例えば眼鏡フレーム）に取り付



(a) システムの全体像



(b) コマンドシート

図1　システムの概念

けるレーザポインタと、机上に設置する「コマンドシート」から構成される。コマンドシートは、制御命令に対応した受光素子を複数個配置したものである。通常は、受光素子には作業空間の光環境によって背景光が入射し、それに応じた電圧が得られる。使用者が制御命令を選択する時には、命令に対応した受光素子をレーザポインタで指示する。その場合、受光素子からは、背景光電圧にレーザ光電圧が重畳した出力が得られる。システムは、重畳分を検出することで、その制御命令が選択されたことを認識する。

これまでの試作システムでは信号検出方式としては、レーザ光で選択される受光素子の電圧と、背景光検出用の素子の電圧とを比較していた。本研究では、背景光検出素子を用いずに、命令選択用受光素子のみでレーザ光照射を判定する方式を提案する。すなわち、照射するレーザ光にパルス変調を行う。受信側では、バンドパスフィルタによって信号のみを検出する。背景光は周波数成分が変調信号と異なるため、フィルタで除去される。

2.2　試作システム

試作システムは図2のように、送信側としてパルス変調のための発振器、発光素子であるレーザポインタ、受信側として机上に配置する受光素子、他の光源からのノイズを除去するためのフィルタ、評価用のコンピュータから構成されている。

白熱電球や蛍光灯は、通常の交流であれば100～120Hz、インバータを用いている場合にはそれ以上の周波数の光を発生するので、レーザ光の変調周波数はそれより小さいことが望ましい。しかしレーザ光の変調周波数が10Hz程度では、使用者が光の点滅を感じて不快になる。そこで、変調周波数を50Hzとした。なお発振器は岩崎通信機製SG-4111を用いた。

発光素子は、レーザポインタ（プラス社製LP-050、最大出力1mW（クラス2））の発光部を取り出し、図3のように眼鏡フレームに固定した。寸法は長さ35mm×高さ22mm×厚さ（レンズ部直径）12mm、重量は約5gである。電源、発振器、およびスイッチング回路は、コードを介して手元に置いた。

受光素子はEG & G Vastec社製可視光導電素子VT43N2を採用した。暗抵抗は300kΩ以上、10ルクスの入射に対して8～24kΩの抵抗値をとる。この受光素子の抵抗値の変化により光を検出する。受光量が大きいほど受光素子の電圧は小さくなる。

フィルタは、エフエヌ回路設計製P-82を用いてバンドパスフィルタを構成し、通過帯域を40～60Hzとして、信号検出する。

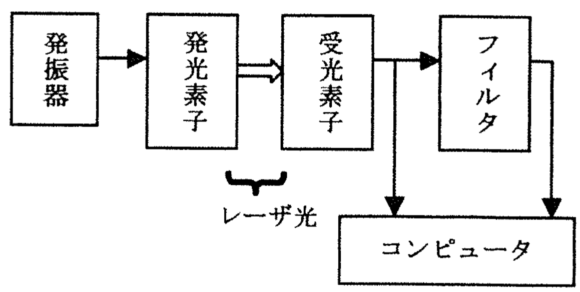


図2　試作システムの基本構成

図3　発光素子

3　信号強度の測定

3.1　実験方法

前章で述べた試作システムを用いて、本研究で提案する方式の実現可能性を確認する。まず、システムを使用する光環境を想定し、検出する信号強度を測定した。本システムを使用する光環境条件として、(a)蛍光灯下の室内、(b)蛍光灯下の室内における電気スタンドの使用、(c)蛍光灯下の室内で使用者の横に立つ人物が受光素子の上に身を乗り出した状態、(d)直射日光の当たらない屋外、(e)直射日光の差す室内、を考える。なお、(b)において電気スタンドは受光素子上30cmに23Wの白熱電球が位置するようにした。

頭部運動による機能的電気刺激（FES）システムの操作を想定して、信号強度測定を実施した。頭部運動による発光素子操作は、健常者(24歳、男性）が行った。

(a)～(e)の条件で机上にシステムを配置し、操作者はその前の椅子に座った。そして発光素子を頭部運動で操作し、それぞれの条件において受光素子にレーザ光を5秒間照射し、受光電圧とフィルタ後の電圧を測定した。なお、レーザ光の非照射中の受光状態を調べるため、レーザ光照射の前後それぞれ5秒間でも電圧測定を行った。

信号強度の評価をコンピュータ上で行うため、電圧信号をサンプリング周波数200HzでAD変換した。照射状態・非照射状態とも5秒間であるが、中間の3秒間について電圧の実効値であるrms値を計算し、それによって信号弁別のしきい値を定めた。

3.2　実験結果

実験風景を図4に示す。なお、いずれの条件でも発光素子から受光素子までは、水平距離60cm、垂直距離40cmであった。

条件(a)～(c)を連続して実施した場合のシステムの出力電圧を図5に示す。グラフ上段には、受光素子出力電圧（以下「受光電圧」)、下段にはフィルタ出力電圧（以下「フィルタ電圧」）を示している。受光電圧波形では、レーザ非照射状態において、それぞれの光環境条件（すなわち背景光）に応じた電圧が発生している。背景光は(c)、(a)、(b)の順に大きくなるので、受光電圧はその順に小さくなる。

レーザ光照射状態においては、パルス変調されたレーザ光に対応する信号が重畳している。一方、フィルタ電圧波形では、レーザ非照射状態では電圧がほとんど発生していない。これは、背景光の主たる周波数成分がフィルタによって阻止されたためである。また(a)から(b)、(b)から(c)の背景光の変化も現れていない。フィルタを用いることで、レーザ光のみを抽出することができる。

条件(d)、(e)の出力電圧を図6、7に示す。波形の傾向としては、条件(a)～(c)と同様であった。しかし、太陽光の影響を受ける(d)、(e)は、背景光が大きい場合ために受光電圧が小さくなり、レーザ光成分の識別が困難であった。

取得した電圧波形から計算したrms値を表1に示す。条件(a)～(c)では、フィルタ電圧の信号強度の違いによって、レーザ光入射が判定できる。非照射状態である0.005～0.007Vと、照射状態である0.043～0.083Vを識別するしきい値としては、それらの中間の0.02V程度が妥当であると思われる。

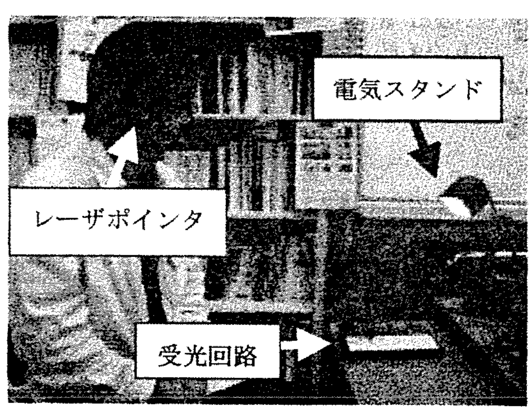


図4　実験風景

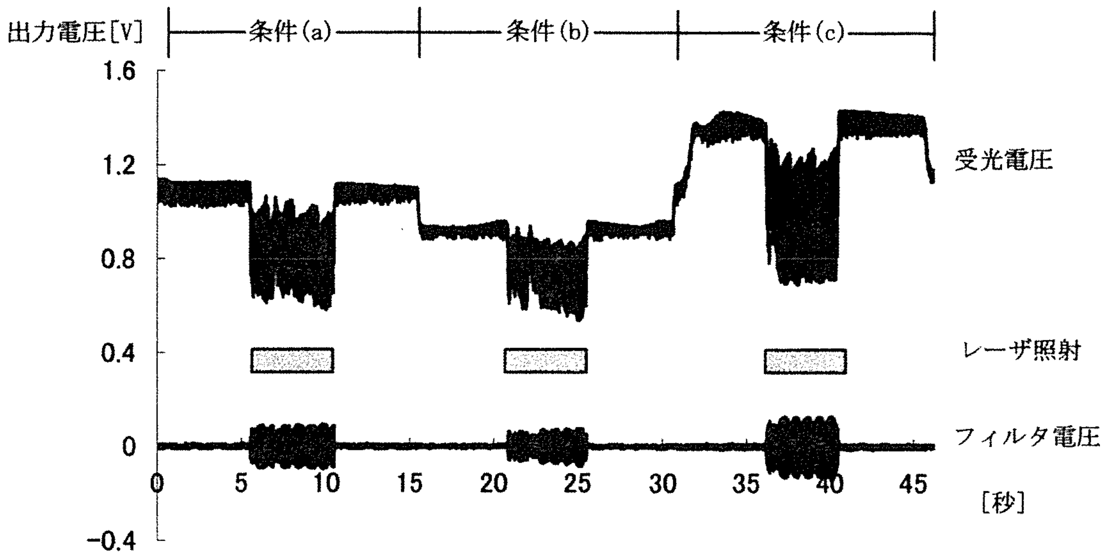


図5　各部の出力電圧波形 (条件(a)～(c)を連続して実施した場合)

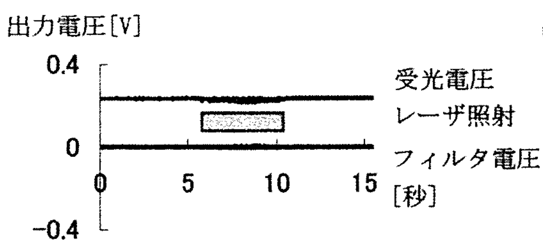


図6　各部の出力電圧波形 (条件(d))

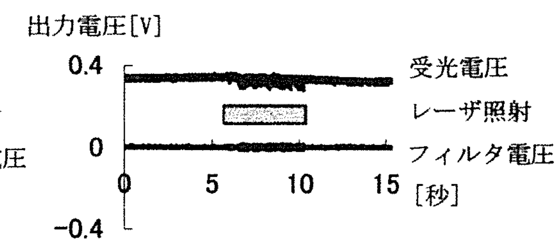


図7　各部の出力電圧波形 (条件(e))

表１　検出した信号強度 (rms 値[V])

	受光電圧		フィルタ電圧	
	レーザ		レーザ	
条件	非照射	照射	非照射	照射
(ａ)	1.083	0.845	0.006	0.061
(ｂ)	0.923	0.771	0.005	0.043
(ｃ)	1.348	1.025	0.007	0.083
(ｄ)	0.239	0.233	0.002	0.004
(ｅ)	0.335	0.334	0.004	0.009

しかし条件(d)、(e)でのレーザ光照射中のフィルタ電圧の信号強度は、(a)～(c)のレーザ光非照射中の強度と同程度であった。よって人工照明環境でのしきい値でそのまま識別することはできない。これについては変調パルスで同期検波を行うことで、信号検出が向上するものと思われる。また、背景光量を学習することでしきい値の決定を自動化し、人工照明と太陽光とでしきい値を変える方法も考えられる。

4　シミュレーション

前章の結果から、人工照明の条件下ではrms値0.02Vによってレーザ光を判定できることを示した。ここではそれに基き、フィルタ電圧波形からレーザ光照射を判定するシミュレーションを行った。フィルタ電圧を整流・平滑化するために、0.5秒の窓関数を1サンプルずつずらしてrms値を計算した。

条件(a)～(c)でのフィルタ電圧波形に対して計算したrms波形を図8に示す。レーザ光の有無で、波形が急峻に変化していることが分かる。フィルタ電圧がしきい値0.02Vを上回ってから、rms電圧が上回るまでの時間遅れは最大でも0.1秒程度であり、操作する上で特に支障はないものと思われる。よってこのような信号を受光電圧0.02Vの電圧比較器に通すことで、スイッチのON/OFF信号が得られるものと思われる。

5　結論

本研究では、四肢麻痺者をはじめとする運動機能障害者が操作できる制御命令入力装置として、頭部運動によって受光素子にレーザ光を照射し命令を選択するシステムに改良を加え、パルス変調方式を採用した。そしてその原理的な実現可能性を確認した。今後は、太陽光環境に対応した検出方法の改良を行うと共に、受光素子数を増やして本システムの操作性の評価を行う予定である。なお、実験に際して御協力いただいた新潟工科大学の篠田優作氏に感謝する。

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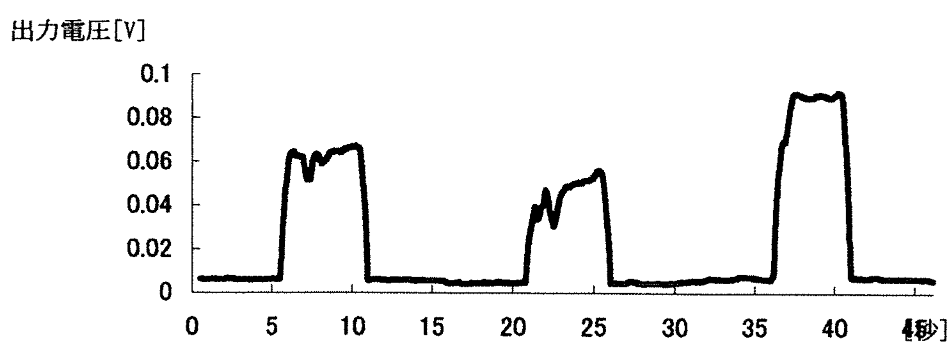

図8　フィルタ後の rms 波形

a more marked improvement in fatigability.

PD-108
Effect of hybrid exercise of the quadriceps femoris

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Objective
To evaluate the hybrid exercise on the quadriceps femoris that uses electrically stimulated antagonists to resist agonist contractions during knee flexion and extension.
Methods
Subjects; Sedentary young 14 men (7 hybrid, 7 control). Hybrid group; Electrical stimulation on the quadriceps was 27.20 %MVC and that of hamstring was 18.58%MVC respectively. Control group; Weight-based volitional quadriceps exercise was performed using 40%MVC. Subjects trained 3 times a week for 6 weeks. Each session consisted of 10 sets of 10 knee flexion and extension exercise. Sets were separated by 1-minute rest intervals and an exercise session required 15 minutes and 40 seconds to complete. Concentric and eccentric knee extension torques were measured.
Results
Concentric extension torque increased 21.18% ($p <= 0.001$) and that of eccentric was 20.31% ($p <= 0.01$) over the hybrid exercise period. They were 23.53% ($p <= 0.01$) and 18.95% ($p <= 0.01$) in the control group. At the 4-week after completion the exercise, they were 19.12% ($p <= 0.001$) and 13.72% ($p <= 0.01$) in the hybrid group, however they were 13.21% (n.s.) and 9.01% (n.s.) in the control group.
Conclusion
Hybrid exercise is effective as well as conventional weight-based exercise in increasing quadriceps strength, although its effect remained longer than that of control.

PD-109
FES control based treadmill rehabilitation after incomplete spinal cord injury

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Objective
The aim of the paper is to present the new approach to the control of the functional electrical stimulation (FES) in incomplete SCI person's treadmill walking (TW).
Methods
The functional movement of the lower extremities during TW of the SCI person was assisted with FES. The appropriate instant of FES triggering was determined by a shank angle (SA) estimated by a sensory device. In the obtained SA the peak value that denotes the pre-swing phase was defined as valuable information for the FES triggering. The intensity of the FES was supervised by a personal computer. The swing phase during TW was also evaluated. If several consecutive swings were "good", the stimulation intensity was decreased and in opposite case when "poor", increased. In the research two incomplete SCI injured patients participated.
Results
The intensity control allowed the patient to put his own effort into the treadmill training. The patient was provided the audio cognitive feedback and was aware of the swing quality. One of the subjects has quickly adapted to the intensity control and during TW he was successfully performing good evaluated swings, consequently the FES assistance was decreased.
Conclusion
We realised that the supervised control of the walking assistance is necessary to involve the patient into rehabilitation process of treadmill training. Strictly speaking the control of FES replaces the strenuous assistance physiotherapists and in combination with the swing phase estimation introduces a new approach in treadmill rehabilitation.

Control of the Functional Electrical Stimulation in Treadmill Rehabilitation

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Summary

The aim of the paper is to present the new approach to the control of the functional electrical stimulation during treadmill training. The emphasis is on sensory supported swing phase estimation and stimulation intensity control depending on the aformentioned estimated swing phase quality. Patients involved in the proposed treadmill rehabilitation are incomplete spinal cord injured with slight ability for walking and therefore may voluntarily influence on their lower extremity movements and therefore on swing phase quality. A patient (C4-5, ASIA C) showed significant decrease of stimulation intensity while maintaining the same swing phase quality.
Keywords: rehabilitation engineering, FES, treadmill, sensors

Introduction

In recent years treadmill training has become a regular daily activity in rehabilitation of the incomplete spinal cord injured (SCI) persons. Treadmill provides ability for cyclical reiteration of the movement to the user and may significantly improve the walking ability [1] and have an indirect effect on reorganization of central nervous system [2]. The aformentioned rehabilitation technique has applied functional electrical stimulation (FES) as an effective motor augmentation aid to redeem the physiotherapist from strenuous work and the longterm activity has shown significant improvement in patient's walking [3, 4].

We have confronted with the fact that in FES treadmill training an improvement is needed. The rehabilitation process repeatabilty was assured by sensory device estimating the swing quality [5] instead of the clasical physiotherapist's visual evaluation. The next step that is presented in this paper was the control of the FES and supporting cognitive audio feedback that awared the patient and notified the physiotherapist of the performed swing quality.

Methodology

The functional movement of the lower extremities during treadmill walking of the SCI person was assisted with FES. The surface single channel peroneal nerve stimulation (laboratory designed experimental stimulator) was used to provoke flexion reflex. The triggering of the FES took place at the fitting moment to achieve an acceptable swing. The appropriate instant of triggering determined the shank angle estimated by a sensory device. The device consists of a two-axis accelerometer and a gyroscope. Kalman filter algorithm was implemented to fuse assessed data and estimate the shank angle. In the obtained shank angle the peak value that denotes the pre-swing phase was defined and in combination with swing phase detection presented the valuable information for the FES triggering.

The intensity of the FES was supervised by special algorithm that runs on a personal computer and is based on sensory information. The swing phase during treadmill walking was estimated and evaluated. If several consecutive swings were evaluated as 'good', the stimulation intensity was decreased and in opposite when evaluated 'poor' increased. The complete scheme of the FES gait reeducation is presented on Fig. 1.

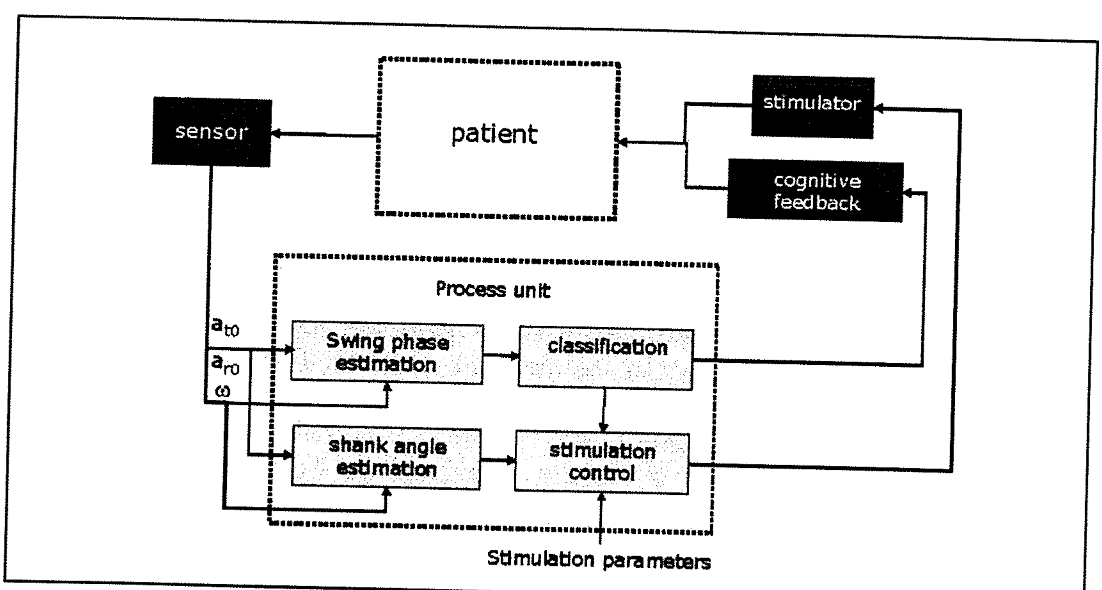

Figure 1. The diagram of FES gait reeducation system. A patient is actively involved and may voluntarily influence on swing quality and consequently on stimulation intensity.

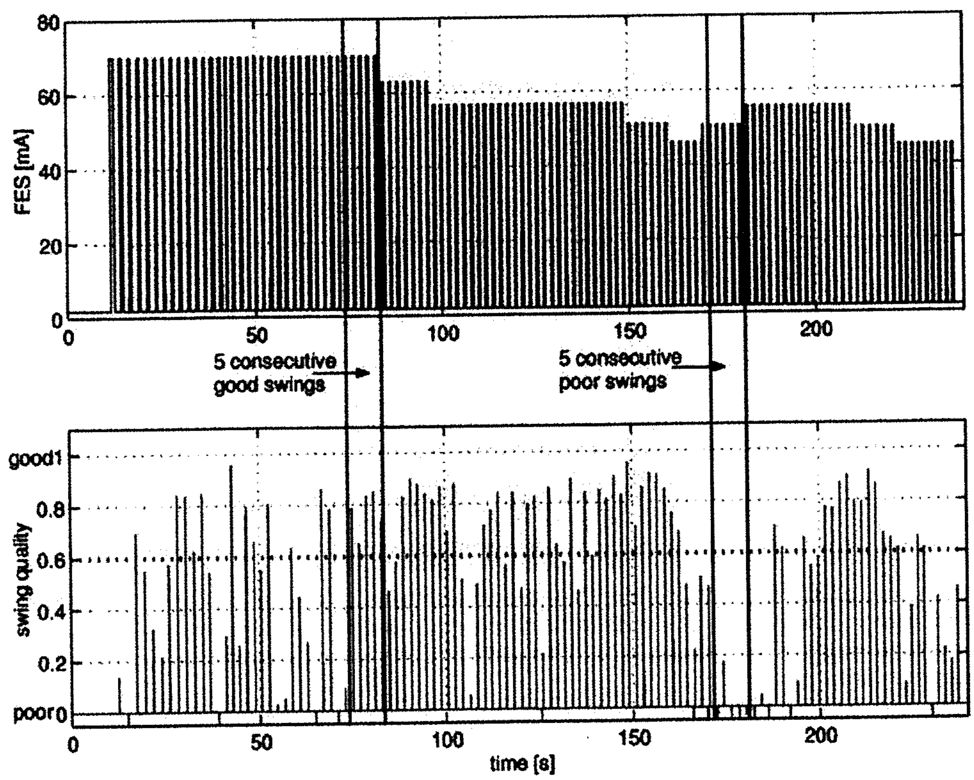

Figure 2. Patient's effort of improving the swing phase has resulted in stimulation intensity decrease.

In the research an incomplete spinal cord injured subject BK (C4-5, ASIA C) and patient ID (stenosis canalis spinalis C2-C3 and protrusion disci C5-C6) collaborated.

Results

The patient BK was trained FES user and quickly adapted to the intensity control. During the 4. min treadmill walking he was successfully performing several consecutive 'good' swings; consequently the FES assistance was decreased (Fig. 2). The control algorithm was also supplemented with cognitive audio feedback (three different sounds, 'poor', 'satisfactory' and 'good').

Discussion

We realised that the supervised control of walking assistance is necessary to involve the patient into treadmill training rehabilitation. Strictly speaking the control of FES replaces the strenuous physiotherapist's assistance

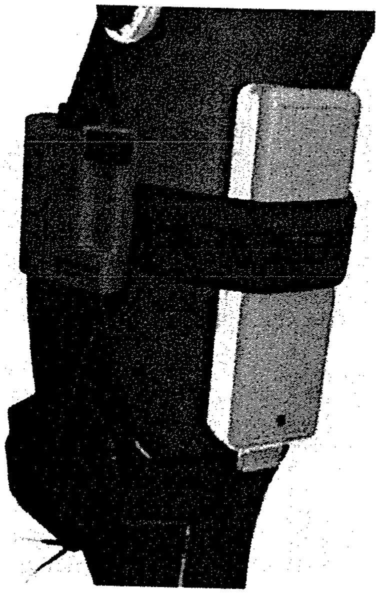

Figure 3. The clinically applicable commercial stimulator will replace the experimental laboratory stimulator. An additional control unit was developed to control the intensity and trigger the stimulator. The unit is controlled by microcontroller built in the sensory device.

and in combination with the swing phase estimation introduces a new approach in treadmill rehabilitation [6]. Since the testing period has shown promising results we have developed the clinically applicable system, where the experimental stimulator has been replaced by clinically evaluated stimulator (MicroFES, Jozef Stefan Institute) (Fig. 3).

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The use of Kalman filtering in assistive device for data assessment and control in gait re-education

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***Abstract*—In the paper an application of Kalman filter technique in rehabilitation engineering is presented. The functional electrical stimulation (FES) is frequently applied motor augmentation aid in incomplete spinal cord injury that is in a need for modification. Therefore the gait re-education system was proposed that successfully applies the presented algorithm.The proposed sensory system is equipped with dual-axial accelerometer and a gyro and is placed at the shank of the paretic leg. The data assessed present input into mathematical algorithms applied for shank angle estimation. The algorithm is based on Kalman filter estimating the angle error and correcting the actual measurement.**

The proposed method was tested during healthy subjects walking on even terrain and treadmill. Afterwards the comparison with optical measurement system Vicon was performed and the triggering of the FES was also tested.

Outcome of the measurements clearly shows how sensory integration algorithm may serve the purpose and its precedence over the single sensor applications.

I. Introduction

In the past most of our research has been focused on restoration of functional movement after spinal cord injury (SCI). We put a lot of effort to improve the role of the functional electrical stimulation as an effective tool in gait restoration of incomplete SCI patients. 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 incomplete SCI patients were candidates for permanent FES. Most of them used FES only as a therapeutic device in the clinical environment or after being released from the rehabilitation center. In these patients surface peroneal nerve stimulation was found useful to provoke flexion response resulting in the swing phase of walking. The most commonly used method for FES triggering is by applying the heel switch. The strain gauge based switch is placed into the shoe and starts the stimulation after the user's heel was off the ground. The precise instant of triggering is set individually for every patient using a time-delay. The heel-switch application is not the most appropriate method, because this forces the patient to be booted while using the stimulator, but is very well accepted by our patients. In therapeutic FES gait training as well as in hemiplegic patients the hand push-button triggering was applied successfully [2]. Several existing systems employing peroneal nerve stimulation used sensory information to trigger FES automatically during walking. The sensory information was usually provided by use of simple artificial sensors [3]. 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 the FES. On the basis of the results obtained, development of a small implantable sensor-stimulator device was proposed. Kostov et al. [4] used the adaptive logic networks to control the functional electrical stimulator for foot drop correction. The sensory signals were assessed from electroneurogram (ENG) and a heel switch. Dai et al.[5] applied various tilt sensors and consequently inferred from the tilt of the body to the patient's intention of walking progress. Therefore the threshold-saturation algorithm was used.

In recent years we put more effort on therapeutical FES combined with treadmill rehabilitation. It has been shown that repeatable action significantly improves any functional movement if the possibility of central nervous system (CNS) reorganization exists [6]. Therefore we suggested an FES rehabilitative system for re-education of walking [7] aimed not only to deliver electrical stimulation to the paralyzed muscles, but also to assess the sensory information from the paralyzed limb and provide the sensory information to the patient and a part of the processed information to the stimulator control unit [8]. The FES rehabilitation systems for re-education of walking are intended to be used for 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. So far we have tried the manual FES triggering using a push-button [9] but soon we realized that the physiotherapist was unable to keep up with the patient's walking speed during treadmill training and consequently triggered the FES earlier or later than required. Therefore we were not able to assure the required repeatability of treadmill walking. To avoid the problem we applied a goniometer placed in knee joint, providing the knee angle. The information was used to automatically trigger the FES before the swing phase took place. To avoid malfunctioning during the stance phase we integrated the knee angle information with the swing phase detection [10]. However we have faced the problem of goniometer slithering and patient's complains of inconvenience. For this reason we considered using already existing sensors in the proposed sensory device that was also used for swing phase detection and estimation. Hereby the efficient algorithm that provide applicable information for FES triggering helped to overcome the existing problems and the patient's inconvenience caused by setting additional sensors. The algorithm for shank angle estimation was developed from widespread idea of orientation estimation with gyroscopes or inclinometers [11],[5]. The main idea however was to develop a successful and applicable algorithm that will be able to take advantage of the applied sensory device in order to replace the unreliable goniometer in FES triggering. In this paper we present the application of such approach during treadmill walking.

Preliminary measurements were aimed at finding the distinctive characteristic signal in patient's walking pattern. For that purpose we performed data assessment with the proposed sensory system together with optical motion analysis system Vicon (© Vicon Motion Systems Ltd.) that has a feasibility of kinematic data analysis. Later on we tested algorithm in the existing gait re-education system [8]. The outcomes were promising and gave us inspiration for further work. We are looking forward to develop a clinically applicable approach.

II. Methodology

A. Hardware

The device consists of two dual-axial accelerometers (Analog Devices ADXL210JE) and a single axial piezo based gyroscope (Murata ENC 03JA). Accelerometers provide digital PWM output that is easily measured by microcontroller (Atmel AT90S4434) timer. Gyroscope provides analog signal that is low-pass filtered with cut-off frequency of 126Hz and afterwards sampled and converted into digital value by microcontroller internal A/D converter (10 bits, up to 15 kHz). Furthermore data are processed into sensible form and transmitted via RS232 communication (38400bps) to a personal computer (PC) or any other process unit, like DSP board we have been also using. The device (Fig. 1) is to be attached by Velcro straps at the shank of the patient.

The existing sensory device, consisting of two major types of sensors for orientation estimation, would be a good solution or substitution for goniometer or heel-switch. The dynamic characteristics of walking, especially heel strike, discouraged us from using accelerometer based inclinometer as the only source of information. Therefore we considered the use of integrated gyroscope signal as the principal source of information to estimate the shank inclination. Gyroscope is liable to bias due to temperature change, random walk and noise [12] thus an efficient mathematical approach like Kalman filter [13] was considered a suitable solution. The Kalman filter implements measured data to correct the estimated model output. In this paper

Fig. 1. The sensory system is attached to the shank. The tangential a_{t0} and radial a_{r0} acceleration assesed by two-axis accelerometer and the shank angular velocity ω presents the measured information during movement in sagital plane.

the Kalman filter is used to correct the estimated shank angle using a gyroscope state-space model.

B. Kalman filter in shank angle estimation

The Kalman filter [13] is a recursive filter (estimator) and presented as a loop algorithm. From initial conditions $\hat{x}(0)^-$ and $\mathbf{P_0}^-$ we are able to calculate the Kalman gain $\mathbf{K_{Kal}}$:

$$\mathbf{K_{Kal}} = \mathbf{P_k}^- \mathbf{H_k}^T (\mathbf{H_k} \mathbf{P_k}^- \mathbf{H_k}^T + \mathsf{R_{vk}})^{-1} \quad (1)$$

where $\mathbf{H_k}$ is the implemented system output matrix, $\mathsf{R_{vk}}$ the measurement noise matrix and $\mathbf{P_k}$ the covariance matrix of Kalman filter. In further step the estimated system state $\hat{\mathbf{x}}_k^-$ is corrected by the measured signal $\mathbf{z}_k$:

$$\hat{\mathbf{x}}_k = \hat{\mathbf{x}}_k^- + \mathbf{K_{Kal}}(\mathbf{z}_k - \mathbf{H_k} \cdot \hat{\mathbf{x}}_k^-) \quad (2)$$

The corrected state of the system $\hat{\mathbf{x}}_k$ is now considered as an output of the filter. The recursive procedure is continued with a calculation of the error covariance matrix:

$$\mathbf{P_k} = (\mathbf{I} - \mathbf{K_{Kal}} \cdot \mathbf{H_k}) \mathbf{P_k^-} \quad (3)$$

After we had determined the present states $\hat{\mathbf{x}}_k$ and appropriate error covariance matrix $\mathbf{P_k}$, we can estimate the one step ahead state of the system:

$$\begin{aligned} \hat{\mathbf{x}}_{k+1}^- &= \mathbf{F}_k \hat{\mathbf{x}}_k \\ \mathbf{P}_{k+1}^- &= \mathbf{F}_k \mathbf{P}_k \mathbf{F}_k^T + \mathsf{Q}_{\mathsf{w}k} \end{aligned} \quad (4)$$

The recursive algorithm of the Kalman filter repeats itself in a loop from the Eq. 4.

There are many ways to estimate the joint angles during human movements. Most of the approaches uses commercial tilt sensors [5]. The simplest and probably effective is the method using two single or one two-axial accelerometer. If the accelerometer is attached to the moving object, in our case along the axis of the shank (Fig. 1), then the inclination of the object axis to the vector of gravity can be calculated with the Eq. 5:

$$\theta_m = [\frac{\pi}{2} - \arctan \frac{a_{t0}}{a_{r0}}] \tag{5}$$

The Eq. 5 determines the shank inclination and is used as an additional measurement for estimated shank angle error correction with Kalman filter/estimator.

In dynamical environments gyroscopes are used to estimate the orientation. Gyroscope provides angular velocity and integration of the signal is needed to define the orientation. In the gyroscope's signal, ie. shank angular velocity, appears the undesired bias b that causes a cumulative error during signal integration. The gyroscope model may be presented in a state-space form in Eq. 6:

$$\begin{bmatrix} \dot{\theta} \\ \dot{b} \end{bmatrix} = \begin{bmatrix} 0 & 1 \\ 0 & 0 \end{bmatrix} \begin{bmatrix} \theta \\ b \end{bmatrix} + \begin{bmatrix} \omega \\ 0 \end{bmatrix} + \begin{bmatrix} n_r \\ n_\omega \end{bmatrix} \tag{6}$$

where the angular velocity is represented by ω, bias by b and uncorrelated Gauss noises n_r, n_ω.

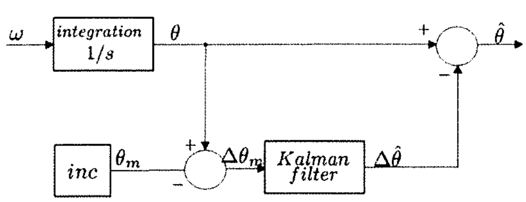

Fig. 2. System for shank angle estimation. Inc is an accelerometer based inclinometer, while ω is the gyroscope input.

Due to the Fig. 2 we need to build an error gyroscope model to be implemented into the Kalman filter. The state-space form of the error model:

$$\begin{bmatrix} \Delta\dot{\theta} \\ \Delta\dot{b} \end{bmatrix} = \begin{bmatrix} 0 & 1 \\ 0 & 0 \end{bmatrix} \begin{bmatrix} \Delta\theta \\ \Delta b \end{bmatrix} + \begin{bmatrix} n_r \\ n_\omega \end{bmatrix} \tag{7}$$

The desired output of the error model, the angle error $\Delta\theta$, is selected with output matrix $\mathbf{H} = \begin{bmatrix} 1 & 0 \end{bmatrix}$:

$$\Delta y = \begin{bmatrix} 1 & 0 \end{bmatrix} \begin{bmatrix} \Delta\theta \\ \Delta b \end{bmatrix} + n_\theta \tag{8}$$

the n_ω, n_θ, n_r are integration noise caused by $1/s$ filter, measurement noise from accelerometer based inclinometer (Eq.5) and gyroscope noise, respectively. Therefore the gyroscope error model may be transformed into state-space

$$\begin{aligned} \Delta\dot{\hat{\mathbf{x}}} &= \mathbf{F}\Delta\hat{\mathbf{x}} + \mathbf{n} \\ \Delta y &= \mathbf{H}\Delta\hat{\mathbf{x}} + n_\theta \end{aligned} \tag{9}$$

to be used in Kalman filter/estimator model (Eq.4). Measurement noise $\mathsf{R}_{\mathsf{v}k}$ and system noise $\mathsf{Q}_{\mathsf{w}k}$ matrix needed for Kalman filter are expressed in the following form:

$$\begin{aligned} \mathsf{Q}_{\mathsf{w}k} &= \begin{bmatrix} N_r & 0 \\ 0 & N_\omega \end{bmatrix} \\ \mathsf{R}_{\mathsf{v}k} &= N_\theta \end{aligned} \tag{10}$$

From the equation for continuous Kalman filter [14]

$$\dot{\mathbf{P}} = \mathbf{FP} + \mathbf{PF}^T + \mathsf{Q}_{\mathsf{w}k} - \mathbf{P} - \mathbf{PH}^T \mathsf{R}_{\mathsf{v}k}^{-1} \mathbf{HP} \tag{11}$$

we may determine the analytical covariance matrix (the analytical solution is possible due to the simple model). Further on we may use the result in equation for Kalman gain Eq. 1. Due to the time-invariant state-space matrix $\mathbf{F}$ and $\mathbf{H}$ the Kalman gain $\mathbf{K_{Kal}}$ remains constant:

$$\mathbf{K_{Kal}} = \begin{bmatrix} k_1 \\ k_2 \end{bmatrix} \tag{12}$$

After the Kalman gain was determined, we may apply the gyroscope error model to the Eq.2 and form the Kalman filter in state-space:

$$\frac{d}{dt} \begin{bmatrix} \Delta\hat{\theta} \\ \Delta\hat{b} \end{bmatrix} = \begin{bmatrix} 0 & 1 \\ 0 & 0 \end{bmatrix} \begin{bmatrix} \Delta\hat{\theta} \\ \Delta\hat{b} \end{bmatrix} + \begin{bmatrix} k_1 \\ k_2 \end{bmatrix} (\Delta\theta_m - \Delta\hat{\theta}) \tag{13}$$

where $\Delta\theta_m = \theta - \theta_m$ is the additional measurement to correct the model estimation.

Considering the variable relationship in the Fig.2 and gyroscope model:

$$\begin{aligned} \hat{\theta} &= \theta - \Delta\hat{\theta} \\ \hat{b} &= b - \Delta\hat{b} \\ \omega &= \dot{\theta} - b \end{aligned} \tag{14}$$

we may describe the system (Fig.2) with 2 inputs ω and θ_m and one output $\hat{\theta}$ transforming the Eq.13:

$$\frac{d}{dt} \begin{bmatrix} \hat{\theta} \\ \hat{b} \end{bmatrix} = \begin{bmatrix} 0 & 1 \\ 0 & 0 \end{bmatrix} \begin{bmatrix} \hat{\theta} \\ \hat{b} \end{bmatrix} + \begin{bmatrix} 1 \\ 0 \end{bmatrix} \omega + \begin{bmatrix} k_1 \\ k_2 \end{bmatrix} (\theta_m - \hat{\theta}) \tag{15}$$

where θ_m presents the measured value, determined from accelerometer based inclinometer in the Eq. 5. The Eq. 15 represents the state-space description of Fig. 2 and is suitable for numerical operations.

III. Results

The preliminary measurements were carried out to determine whether the shank angle estimation algorithm would satisfy the demand for accurate measurements needed to control the FES and to provide reliable information on the quality of each accomplished swing of the extremity. We were looking for significant peaks in the assessed signals that we might be able to use for the purpose of triggering of the electrical stimulation. Therefore we used the optical motion analysis system Vicon (©Vicon Motion Systems) to confirm the approach that could replace the use of goniometers in the gait re-education system [8]. Two healthy subjects AO (male, 172 cm, 65 kg, 24 years) and IC (male, 173 cm, 74 kg, 29 years) participated in the preliminary test. Their task was to walk with normal and slow speed (0.7-1.2 km/h).

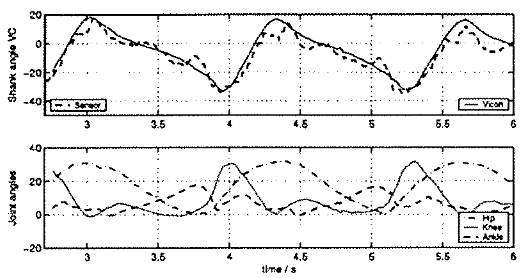

Fig. 3. Kinematic analysis of the subject's IC even terrain walking. The comparison with Vicon Motion Analysis System is shown.

Upper graph of the Fig. 3 presents the time-course of the shank during walking. These measurements were supported by Vicon Motion System and a comparison between the assessed data and the data proposed by the measuring device was carried out simultaneously. The shank angle was determined from pelvis, hip and knee joint angles Eq. 16:

$$\theta_{shank} = \vartheta_{hip} - \vartheta_{pelvis} - \vartheta_{knee} \tag{16}$$

The Fig. 4 clearly demonstrates the advantage of sensory integration. The implementation of Kalman filter could in this case overcome the inconveniences caused by single sensor use. The shank ankle, determined by two-axial accelerometer, contains high frequencies, caused by heel strike, while the gyroscope is well known for its bias as a consequence of temperature change. The integration of the gyroscope bias has resulted in a ramp response. Using the Kalman filter estimating the angle error model both inconveniences were overcome.

Afterwards we have evaluated the effectiveness of the Kalman filter error estimation. Using the cross-correlation coefficient as the curve fitting criterion proved very competent. Therefore the equation (Eq.17) applies to remove the mean from each column before calculating the correlation coefficient (Eq.18) between the data assessed by Vicon Motion System and the Kalman filter based system output for N = 75 gait cycles [15]:

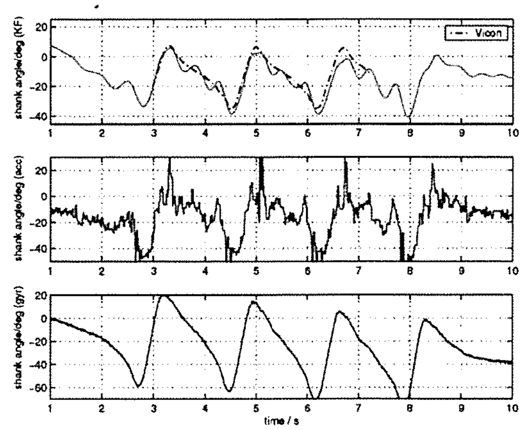

Fig. 4. Kalman filter (KF) is a good solution to overcome gyroscope (gyr) integration error and accelerometers (acc) dynamic restriction.

$$\begin{aligned}\varphi_{\theta_{shank},\hat{\theta}}(t,\tau) &= \frac{1}{N}\sum_{k=1}^{N}\theta_{shank}(t)\cdot\hat{\theta}(t+\tau) \\ &= E[\theta_{shank}(t)\cdot\hat{\theta}(t+\tau)]\end{aligned} \tag{17}$$

$$\begin{aligned}&\rho_{\theta_{shank},\hat{\theta}}(t,\tau) \\ &= \frac{E[(\theta_{shank}(t)-m_m(t))\cdot(\hat{\theta}(t+\tau)-m_r(t+\tau))]}{\sqrt{E[(\theta_{shank}(t)-m_m(t))^2\cdot(\hat{\theta}(t+\tau)-m_r(t+\tau))^2]}}\end{aligned} \tag{18}$$

where N is a number of samples included into computation. In the Eq. 17 θ_{shank} is the shank angle computed from Vicon data, while $\hat{\theta}$ presents the data output of the Kalman filter. m represents the mean of the signal and τ means a time delay. When the calculated coefficient $\rho_{\theta_{shank},\hat{\theta}}$ is close to 0, there is no correlation between the signals and closer to 1 it gets, more signal resemblance is expected.

The outcomes were very satisfying. The correlation coefficient varied from 0.9490 to 0.9615 in several gait cycles. Unfortunately we have not been able to assess all the gait cycles from single walking time-course due to the Vicon motion assessment constraints, therefore we used the data from several measurements of the same person and performed correlation between data assessed by sensor and Vicon in single gait cycle. Afterwards we calculated the mean and the standard deviation(σ) to express the compliance of both signals:

$$\begin{aligned}\bar{\rho}_{\theta_{shank},\hat{\theta}} &= \frac{1}{N}\sum\rho_i \pm 2\sigma \\ &= 0.9553 \pm 0.0088\end{aligned} \tag{19}$$

We realized that the compliance was almost perfect as shown by ρ =0.9553 and the deviation between gait cycles was very low (σ =0.0088). Focusing on some particular regions in shank angle time-course we realized that a good compliance is shown, there is no drifting as in the integrated gyroscope signal, while during initial foot contact and in stance phase we may notice deviations. Those deviations appear as a contribution of the accelerometer signals. On the other hand it is obvious that the Vicon shank angle measurements are repeatable and reliable (4). We did not pay attention to dynamical disturbances during stance phase, as the emphasis in our application is on swing phase. Parenthetically we had to take into consideration that also the placement of the device on the right shank may not have been perfect and might have slipped on occasion. In this case we also presumed that the Vicon outcomes were absolutely accurate although there might have been a slight marker displacement.

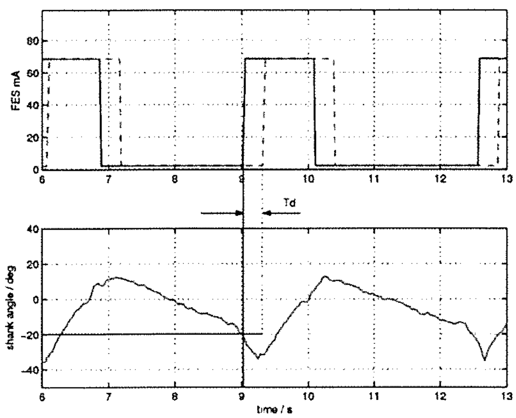

Fig. 5. Shank angle has been used to trigger the functional electrical stimulation. When the subject proceeded to the midswing, the shank angle reached 20° flexion and the stimulation had started with regard to the pre-set time delay T_d.

The purpose of the algorithm is to trigger the functional electrical stimulation. Following our experience and general knowledge on gait analysis [15] we have determined the start of the swing phase. The patients that use peroneal nerve stimulation to invoke the flexion reflex [1] need motor augmentation in midswing phase when the lower extremity proceeds into the forward swing. Thus the stimulation could be triggered when the shank angle reaches 20° flexion. As presented in Fig. 3 the related event occurs before the peak in shank angle time-course. The stimulation trigger was determined using a threshold algorithm in conjunction with the shank angular velocity, assessed by the gyroscope. The gyroscope signal consists of an easily detectable wave that changes the sign. The product of those two signals could be uniformly defined and was applied to trigger the electrical stimulation. Due to the clinical demands and patient's needs as well as physiotherapist's wishes the instant of triggering should have an option to be delayed for the pre-set time constant. Comparison of the both graphs in Fig. 5 gives us the complete picture of the shank angle estimation use in gait re-education system.

In so far as the proposed algorithm was intended for the gait re-education system we have also performed a few tests during subject AO's walking on a treadmill in order to test the repeatability and reliability of FES triggering. In this case, there was no actual stimulation present. Stimulation was only simulated. The Fig. 6 clearly shows that no mistake had occurred during cyclical walking.

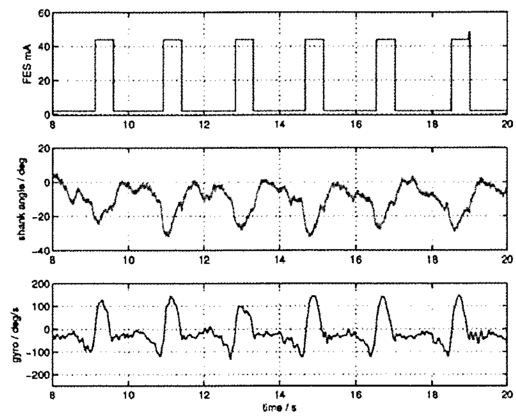

Fig. 6. Shank estimation and simulated FES triggering during healthy person AO's walking on the treadmill. The shank angular velocity is present. The positive angular velocity wave determines the swing phase of walking.

IV. Discussion

The proposed method could outrun the existing single sensor methods for tilt estimation for the gait analysis purposes. Several tilt sensors are based on two-axis accelerometer and the appertaining low-pass filter [5]. The cut-off frequency of such filter needs to be below 2 Hz. Realization of such filter may cause difficulties in a case when a steep slope is required, while the filter order may enormously increase, consequently reflecting as an inadmissible time delay. The steep slope may be necessary due to the higher frequency response in the heel contact phase. Therefore the use of the accelerometer based single sensor tilt has proved unsuitable for shank ankle estimation where the movement dynamics of the extremities requires wider frequency range. One of the possible solutions has been presented in this paper. The implementation of a gyroscope as the main sensory information into the angle estimation algorithm using Kalman filter could be a good solution.

An alternative to the proposed method could have been realized as an algorithm with two separately designed filters, a low-pass for accelerometer based tilt sensor filter and high-pass filter for the integrated shank angular velocity. The problem that arises here is related to the selection of the appropriate cut-off frequency. The latter could have been determined by optimization methods that are based on minimization of the signal error when reiteratively changing the parameters of the cut-off frequency of the low and high pass filters [16]. Hereby we emphasize the systematical approach how to estimate the certain value and make an on-line correction of the error turning up during the session.

The measurements were based on healthy subjects walking in order the evaluate the measurement system in an evaluation process that is ethically incontestable. Hereby we should mention that the sensory system and the algorithm have already been implemented into the FES gait re-education system and tested and applied to the incomplete spinal cord injured patient during treadmill rehabilitation process [8].

V. Conclusion

The method has proved reliable and therefore efficient for FES triggering. As presented in results there have not been any difficulties in FES control. The instant of triggering was also in compliance with the starting moment of the lower extremity forward swing. The outcome of our results rewarded our effort to replace the goniometer and refrain the system from using more additional sensors. In this respect we should mentioned that we would like to refrain our patients from wearing heavy sensors and we keep on minimizing the weight during treadmill walking. The alternative to this method is an optical measurement system we were using. But there are several institution that have no optical measurement system in their possession. Usually the optical systems require several calibrated cameras to be positioned. Therefore they are located in the kinesiology laboratories and unfortunately away from the clinical environment where the gait re-education is applied to the patients.

Acknowledgment

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SENSORY SUPPORTED FES CONTROL IN GAIT TRAINING OF INCOMPLETE SCI PERSONS

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Summary

Sensory supported electrical stimulation of the peroneal nerve during treadmill walking is proposed as a gait training modality in incomplete spinal cord injured (SCI) patients. Multisensor device provides the information on the tilt of the shank during the swing phase. The provided information significantly improves the triggering instant of the electrical stimulation. In the same time the swing phase estimation serves as a reference to determine the needed motor augmentation support. Both approaches, triggering as well as the intensity control of the functional electrical stimulation, were applied on healthy person and on C4-5 SCI patient.

State of the Art

In the last decades more incomplete than complete SCI patients are arriving to the spinal units. One of the primary goals of the rehabilitative program for the incompletely paralyzed subjects is to restore their walking patterns. The gait pattern can be restored in paralyzed persons by surface or implanted multi or single channel functional electrical stimulation (FES), by delivering electrical stimuli to the efferent and/or afferent nerves. The swing phase obtained by eliciting a synergistic flexion response through electrical stimulation of the common peroneal nerve was extensively used by our group [2]. The gait training modalities eliciting reflex responses result in complex and natural like movements which are in consequence provoking afferent signals in joints, tendons, and muscles, important for re-education of walking. Treadmill walking is producing hip extension at the end of the stance phase which is inducing reflex hip flexion and thus initiating the swing phase of walking [1]. The treadmill training can be combined with electrical stimulation of the partially paralyzed extremities and may be used in combination with body weight support (BWS) [4]. The BWS helps the patient to concentrate on walking without having problems with maintaining stability. In the early gait training either physiotherapist or patient is manually triggering the electrical stimulation [2]. When the triggering is performed by the physiotherapist, then the patient is able to focus on the gait performance, while the physiotherapist's task remains to be the estimation of walking quality. Therefore, the instant of triggering is based on physiotherapist's experience and may vary from step to step. Consequently, the patient's walking performance depends on physiotherapist's skills. In order to overcome this undesired dependency, the FES triggering should be automatic, i.e. linked with a selected gait event or gait phase. There were described several attempts of using sensory information for direct control of FES to achieve the desired joint motion [6]. Hereby, we suggest the use of the tilt information of the shank in combination with an algorithm for gait evaluation [3]. The algorithm estimates the quality of the performed swing phase on the basis of the detected gait cycle and assessed acceleration time-course. Afterwards, the swing phase quality is classified into three levels and provided to the patient during treadmill walking as an audio cognitive feedback. The proposed method allows the patient to fully cooperate in the rehabilitation process and to take voluntary actions to improve his/her walking pattern.

In the paper we are proposing the use of the sensory information to trigger the surface peroneal nerve stimulation combined with treadmill walking as a modality for gait training in incomplete SCI persons [5].

Methods

The swing phase of walking in incomplete SCI patients is usually achieved by electrical stimulation of the peroneal nerve. Manual pushbutton or footswitch triggering was replaced by the proposed method based on the assessment of the tilt of the shank. Despite of the fact that the shank angle could be determined by two-axial accelerometer with low-pass filter, we had difficulties designing the efficient low-pass filter due to the time-delay of the high order filter. Therefore, we applied the gyroscope that was built in the multisensor device for the purposes of the swing phase estimation [3]. Using both sensor types and applying a recursive Kalman filter, we

were able to determine the shank angle irrespective of the sensor misplacement or strong heel strike, which were frequent sources of error.
The analysis of the gait cycle shows that the shank angle reaches its peak, in clinical terms the maximal knee flexion, in the pre-swing phase. This is the moment before the toe-off when the lower extremity goes into the swing phase. During the swing phase the knee joint moves toward extension. Therefore, the peak in the shank angle time-course was used to trigger the FES. We have also introduced an adjustable time delay to assure the appropriate instant of triggering for the patients with various motor disabilities.

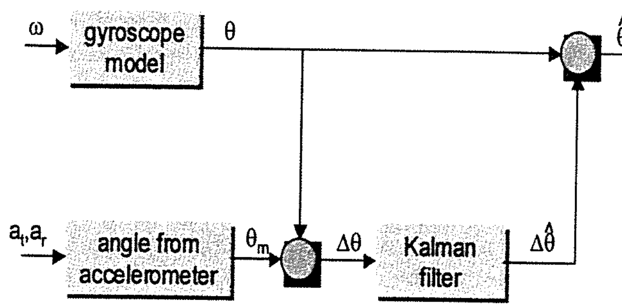

Fig. 1 A Kalman filter is implemented for accurate shank angle (θ) determination.

In the present investigation a healthy subject and a patient with C4-5 spinal cord lesion have participated. The role of the measurements in the healthy subject was to verify the assessed reference shank angle trajectory, while the patient was involved into the FES treadmill walking. The surface electrodes were placed over the peroneal nerve with the aim to evoke the flexion reflex. A stimulation frequency of 20 Hz, a pulse duration of 0.2 ms and intensity of 35mA were used during the swing phase performance.

Results

The electrical stimulation was triggered by the help of the estimated shank angle as presented in fig.2. When the subject entered the pre-swing phase, the shank angle reached the peak value. Considering the predefined time-delay, the train of electrical stimulation pulses was delivered to the peroneal nerve. Before the gait training session also the duration of the train of stimuli, depending on patient's deficits and demands, was set up.
Our patient had difficulties performing a swing phase and was unable to make a progress of his lower extremity into the swing phase without stimulation. The applied peroneal nerve stimulation significantly increased hip and knee flexion and ankle dorsiflexion during the swing phase of walking. Consequently, the leg progressed into the swing phase efficiently as shown in fig.3. Even more, the patient was able to maintain the stability and walk on treadmill without arm support.

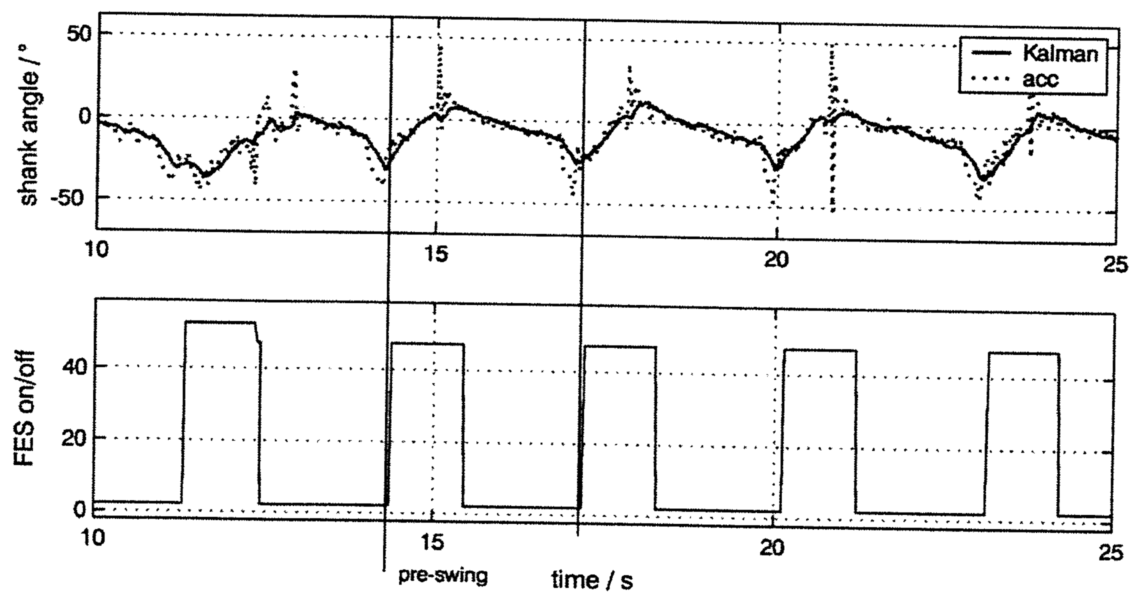

Fig. 2 The instant of FES triggering is determined with shank angle. A time-delay can be set manually.

Fig. 3. Record of an incomplete SCI patient's walking with the peroneal electrical stimulation.

Discussion

FES for lower extremities has been a research issue for several decades. Most of the conclusions were stating that permanent use of FES cannot be very efficient due to the muscular fatigue and patients' rejection of cumbersome devices. We can claim that therapeutic FES has proven successful, especially in combination with other rehabilitation techniques, such as treadmill walking. We have shown an efficient cooperation of the patient which is demonstrated by the successful combination of patients' voluntary action and the use of FES during treadmill walking [5]. The method proposed in the paper is also trying to point out the use of small, portable multifunction sensory system to control the FES for therapeutical purposes after incomplete spinal cord injury.

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Sensory Supported FES Control in Gait Training of Incomplete SCI Persons

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Sensory supported electrical stimulation of the peroneal nerve during treadmill walking is proposed as a gait training modality in patients with incomplete spinal cord injury. A multisensor device provides information about the tilt of the shank during the swing phase of walking. The information, which is simultaneously delivered to the patient and the electrical stimulator, significantly improved the instant of triggering of the electrical stimulation. At the same time the swing phase estimation served as a reference to determine the required motor augmentation support. Both approaches, triggering as well as the intensity control of the functional electrical stimulation were tested in a healthy person and in a C4-5 incomplete spinal cord injured patient.

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Spastic Bladder in Spinal Cord Injury—17 years of Experience with Sacral Deafferentation (SDAF) and Implantation of an Anterior Root Stimulator (SARS)

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Purpose: Spinal cord injured patients with a suprasacral lesion usually develop a spastic bladder. The hyperreflexia of the detrusor and the external sphincter cause incontinence and threatens those patients with recurrent urinary tract infections (UTI), renal failure, and autonomic dysreflexia. All of these severe disturbances may be well managed by sacral deafferentation (SDAF) and implantation of an anterior root stimulator. Materials and Methods: 464 paraplegic patients (220 female, 244 male) received a SDAF-SARS between September 1986 and December 2002. Almost exclusively the SDAF was done intradurally, which means with one operation field two steps (SDAF and SARS) can be done. Results: 440 patients have been followed over an average of 6.6 years (range >6 months–17 years). The complete deafferentation was successful in 94.1%. 420 paraplegics can use SARS for voiding (frequency 4.7 per day) and 401 use it for defecation (frequency 4.9 per week). Continence was achieved in 364 patients (83%). UTI declined from 6.3 cases per year preoperatively to 1.2 per year postoperatively. Kidney function was stable. Early complications were 6 CSF leaks and 5 implant infections. Late complications with the receiver or cable failures led to surgical repairs in 34 paraplegics. A step-by-step program for trouble-shooting differentiates implant failure, myogenic, or neurogenic failure. Conclusion: SDAF is able to restore the reservoir function of the urinary bladder and to achieve continence. Severe autonomic dysreflexia disappeared in most of the cases. SARS certainly allows poststimulus voiding, but by means of an accurate adjustment of stimulation parameters it is possible to accomplish low resistance micturition. The microsurgical technique requires an intensive education. One has to be able to manage late implant complications.

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Calculation of Current Density Distribution in Biological Matter with the Finite Element Method

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The current density distribution during electric stunning of slaughter pigs was calculated and visualized with the Finite Element Method (FEM) using the computer program Ansys (Ansys Inc.). The different anatomic structures like muscle, bone, fat, blood