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Use of telekinesthetic feedback in walking assisted by functional electrical stimulation

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A telekinesthetic feedback implemented into functional electrical stimulation (FES) orthosis is described. Single channel FES is used to provoke ankle dorsiflexion during walking. FES is controlled manually by a special lever, built into the handle of the crutch. The angular position of the lever defines the intensity of stimulation and thus the magnitude of the ankle dorsiflexion. The measured joint angle provides the feedback information about the ankle joint position, which is presented to the user as a force feedback applied to the control lever. As the first step in the development of a complex micromechatronic device, a simulated testing environment was prepared. A computer model, comprising dynamic foot characteristics, as well as agonistic and antagonistic muscle groups, substitutes the ankle joint. The model also includes fatiguing of the electrically stimulated muscles. For experimental purposes an actuated control lever was built. The efficacy of the telekinesthetic feedback was evaluated in a group of six healthy persons.

Introduction

Walking is one of the most important human activities. Most spinal cord injured (SCI) persons, however, are deprived of this advantage. In some cases their locomotion is severely disturbed or they are not able to walk at all. With the assistance of crutches and functional electrical stimulation (FES) standing and simple gait were restored [1].

Because of the absence of sensory feedback from the paralysed extremity, human vision remains the only useful feedback for a SCI walker. This of course interferes with the primary function of the vision which is observing the surroundings. The simplest command sensor, independent of human vision, is the heel switch used for FES assisted walking in hemiplegic patients [2]. The switch is located in the sole of the shoe on the affected side. When a patient voluntarily raises the heel of the affected leg, the heel switch triggers the stimulator, causing foot dorsiflexion during the swing phase of the gait.

The great majority of the incomplete SCI subjects, who are candidates for a FES orthosis, are crutch users [3]. Because of frequent malfunctioning of the heel switch, a pushbutton built into the handle of a crutch was introduced [4]. In this application, as long as the subject was depressing the pushbutton, the stimulation was applied to the electrodes placed over the ankle dorsiflexors. In this way, the FES assisted swing phase of walking was produced in incomplete SCI persons.

A major issue associated with the FES is the reduction of muscle force in time, as a result of fatigue. In electrically induced contractions, the muscle fatigue is essentially peripheral and presents a severe problem, particularly in those cases where detection of fatigue is not possible due to the absence of sensory feedback [5]. A possible way to compensate for this difficulty is to replace the pushbutton by a proportional control built into the handle of a crutch in a form of a lever. In this way the subject can increase the amplitude of the electrical stimulation and thus overcome the effect of muscle fatigue.

The purpose of the paper is to present an improved FES orthotic system with manual control lever and telekinesthetic feedback built into the handle of the crutch. The aim of the telekinesthetic feedback is to provide information about the position of the paralysed joint to the patient. The ankle joint angle is measured by an electrogoniometer and transferred to a torque motor actuating the axis of the control lever. This way, the resistance of the control lever to manual voluntary control is proportional to the magnitude of the joint angle. By adjusting the pressure on the control lever, the walking subject can overcome the problems of fatiguing of electrically stimulated muscles and also can deal with the obstacles on the ground.

Methods

FES orthosis

The basic concept of the described FES orthosis including the telekinesthetic feedback consists of the following components (figure 1):

(1) control unit with electrical stimulator;

(2) stimulation electrodes;

(3) manual control lever with electrical motor;

(4) goniometer.

The manual control lever is built into the handle of the crutch and is connected to the shaft of an electrical motor directly or through a gearbox. The user determines the amplitude of electrical stimulation by changing the lever rotation angle. The measured angle

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represents the input information to the control unit where the corresponding FES amplitude is computed. The stimulator, integrated into the control unit, uses the computed data and stimulates the ankle joint dorsiflexors through surface stimulation electrodes. The resulting dorsiflexion is measured by a goniometer, placed at the ankle. The magnitude of the joint angle is translated into the counteracting torque in the manual control lever, produced by a torque motor installed in the handle in the crutch.

In a preliminary study the FES orthotic system was replaced by a simulated environment. For the purpose of variable amplitude FES triggering and telekinesthetic feedback an experimental hardware unit was built (figure 2). The goniometer, the electrical motor and the control lever are replaced by a much bigger motor with a steering handle and an optical encoder. The biomechanical computer model of the human ankle replaces the real human lower extremity and is software generated. The control unit with the electrical stimulator is software generated as well.

The functional electrical stimulator is presented by a pulse generator having a variable amplitude output and serves as the input into the musculoskeletal model of the human ankle. The amplitude of electrical stimulation is proportional to the control lever angle. Figure 3 shows two control loops. The first loop is used for returning the manual control lever into the starting position corresponding to the neutral position of the ankle joint. This loop has no special effect on the torque feedback; in this way the user develops a feel for the floating zero. The starting point is defined by the input parameter ϑ_{ref} and is set by the user. The second loop controls the counteracting torque. The output of the musculoskeletal model, the ankle angle, is turned into a control signal delivered to the electrical motor by the control unit.

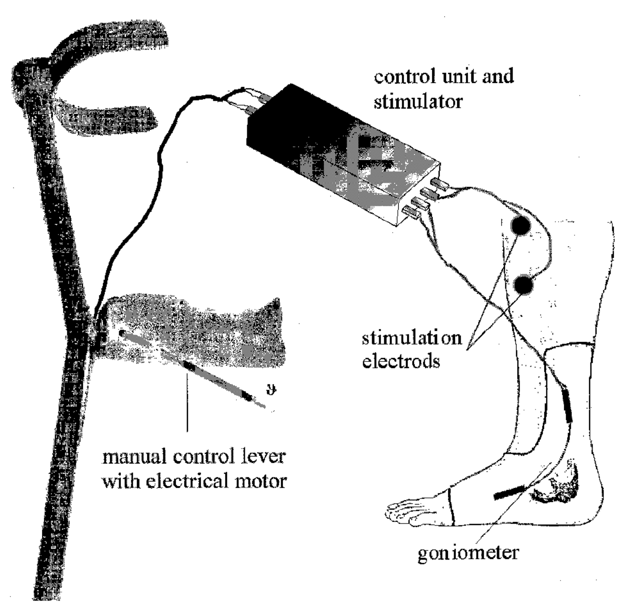


Figure 1. FES orthosis based on telekinesthetic feedback.

Musculoskeletal model

The ankle joint model includes several independent subsystems: the ankle muscle groups, the joint characteristics and the foot being part of a whole musculoskeletal system. The ankle joint muscles are divided into two muscle groups, performing the opposite tasks: the agonistic muscle group performs the dorsal foot flexion and the antagonistic muscle group performs the plantar foot flexion.

The muscle group characteristic parameters are presented as viscous damping $B_1(B_2)$, inner moment of inertia $J_1(J_2)$ and elasticity $k_1(k_2)$. The index 1 belongs to agonist while 2 to the antagonist muscle group. The foot moment of inertia J_3, the joint elasticity k_3 and the joint viscous damping B_3 supplement the model in conditions of free movement. The moment of inertia of the foot was calculated by replacing the foot with a prism [6]. The following equations describe the behaviour of the ankle joint in unconstrained conditions (1):

$$\begin{aligned} J_1\ddot{\varphi}_1 &= -B_1\dot{\varphi}_1 + T_{iso} - k_1(\varphi_1 - \phi), \\ J_3\ddot{\phi} &= -k_1(\phi - \varphi_1) - k_3\phi - B_3\dot{\phi} - k_2(\phi - \varphi_2), \\ J_2\ddot{\varphi}_2 &= -k_2(\varphi_2 - \phi) - B_2\dot{\varphi}_2, \end{aligned} \tag{1}$$

where ϕ represents the ankle joint angle, while φ_1 and φ_2 are the displacement of the agonist and antagonist muscle respectively.

The dorsiflexor moment generator is denoted by T_{iso}. It only acts in the direction of muscle contraction. The moment generator T_{iso} is not linearly proportional to

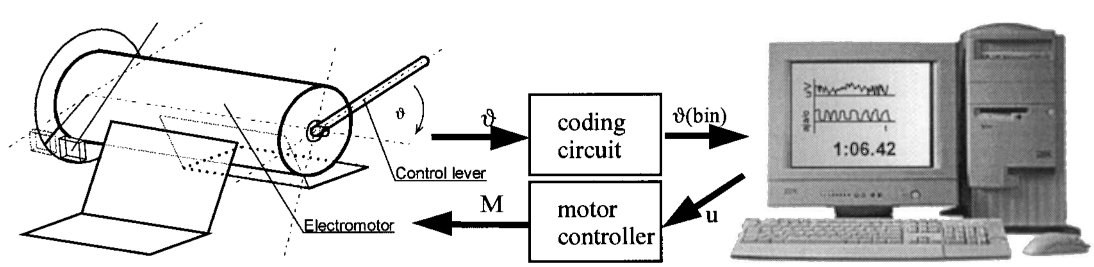


Figure 2. The experimental set-up consisting of hardware and software components. ϑ represents the steering handle angle, ϑ(bin) is the binary value transferred to the computer. u is the control signal to the motor controller and M is the desired counteracting torque, produced by the electrical motor.

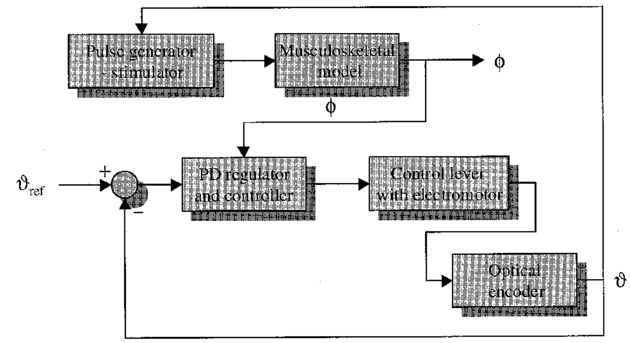


Figure 3. Block diagram of the system. υ is the angle of the manual control lever while ϕ represents the ankle joint angle. The starting point of the steering handle is defined by the input parameter υ_{ref} and is set by the user.

the stimulating voltage U_{st}. Two nonlinearities are incorporated, the threshold stimulation voltage and the saturation voltage [6]. T_{iso} is also a time-dependent function, defined by the muscular fatigue.

The numerical values of these elements cannot be determined by a direct measuring method. For that purpose frequency response based methods and identification approaches were used [6] (k_1=3 mN rad^{-1}, J_1=0.0245 mN s^2 rad^{-1}, B_1=0.17 mN s rad^{-1}, k_2=3 mN rad^{-1}, J_2=0.0245 mN s^2 rad^{-1}, B_2=0.17 mN s rad^{-1}, k_3=13 mN rad^{-1}, J_3 =0.024 mN s^2 rad^{-1}, B_3 = 2.7 mN s rad^{-1}).

Fatigue in muscles activated by FES

In general, the fatigue is the inability of a muscle or muscle group to continuously produce the required force, regardless of the type of work in progress. Muscular fatigue is a time-dependent process that may take place gradually or abruptly [5]. The fatigue can also be considered as a protective mechanism of the body to prevent permanent muscle injuries.

The nature of fatiguing observed in paralysed muscles activated by FES is different from normal fatigue. If the comparison with voluntary activated normal muscle fatigue is made, the following differences can be noticed [5]:

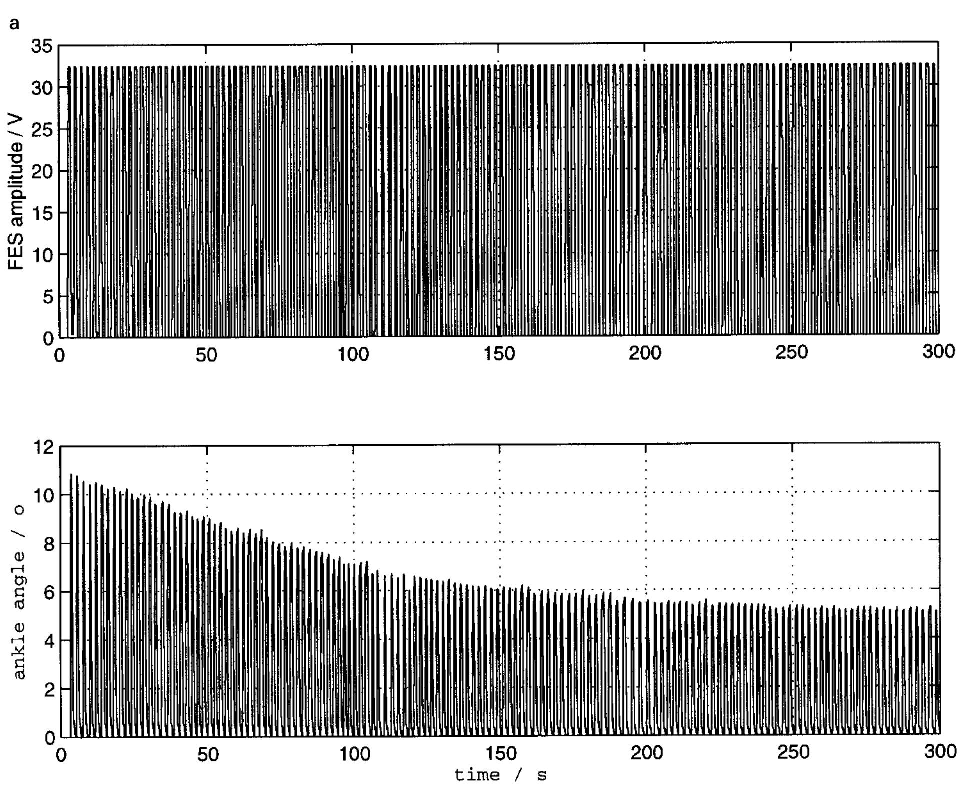
(1) mode of activation: FES synchronously activates all muscle motor units;

(2) reverse recruitment order: the rapidly fatiguing fast motor units are recruited at low stimulus intensities;

(3) peripheral fatigue: the central component of fatigue is missing because of the spinal cord injury;

(4) lack of sensory feedback: the muscle fatigue is not percepted by pain.

Electrically stimulated muscles fatigue more rapidly than when voluntarily activated. With intermittent stimulation,



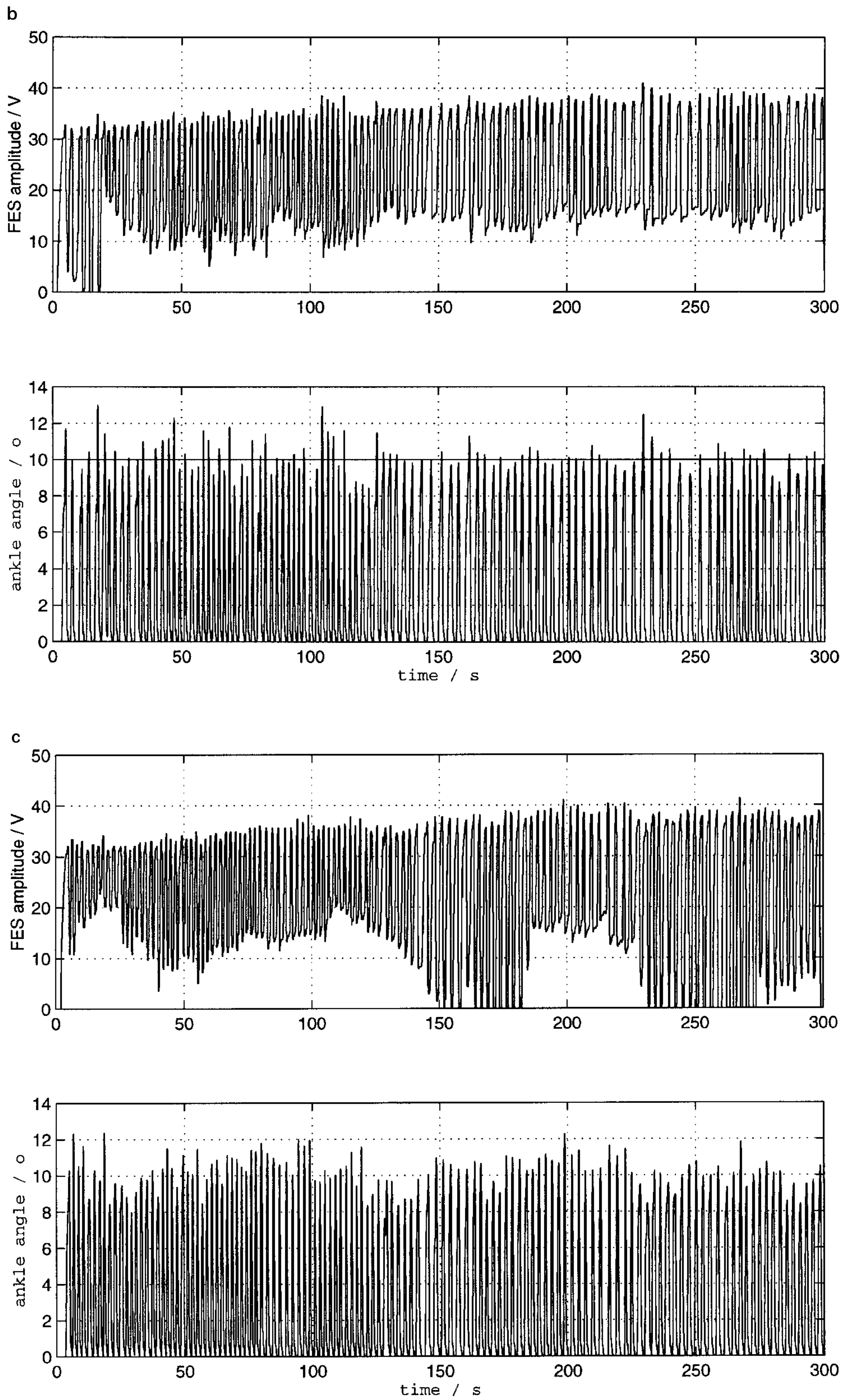


Figure 4. (a) The upper diagram shows the trains of electrical stimulation while the lower diagram presents the effect of the muscular fatigue on the ankle goniogram. (b) In the second test the tested person was trying to reach the desired ankle joint angle (10 °) while using both visual and telekinesthetic feedback. The effect of the muscular fatigue was overcome by increasing the stimulation amplitude (upper diagram). (c) The relationship between the ankle joint angle and the counteracting moment in the control lever was the only feedback information in the third test. The lower diagram presents a successful attempt of using the telekinesthetic feedback.

Table 1

	Test									
Test	1		2		3		4		5	
Subject	$\overline{x}_i$	s_i	$\overline{x}_i$	s_i	$\overline{x}_i$	s_i	$\overline{x}_i$	s_i	$\overline{x}_i$	s_i
1	9.85	1.19	9.94	1.22	10.1	1.67	9.15	1.75	9.99	1.71
2	10.6	1.29	10.5	1.27	11.1	1.41	10.6	1.47	10.3	1.53
3	10.3	1.59	10.9	1.16	9.52	1.66	10.1	1.38	10.9	0.99
4	10.3	1.83	10.7	1.74	8.88	1.68	11.1	1.02	10.3	1.39
5	8.70	0.89	9.08	1.46	8.92	1.05	9.67	1.23	10.7	0.97
6	10.3	1.19	8.62	1.08	9.50	1.25	9.80	1.26	10.6	1.43

where trains of stimuli are followed by a pause, fatiguing of electrically stimulated muscle is considerably less when compared to continuously delivered stimulation.

For the purpose of muscle fatigue modelling the following time course of the muscle moment was selected [5]:

$$T_{\mathrm{iso}} = \{a_0(1 - a_1 \exp^{(-t/\tau_1)} - a_2 \exp^{(-t/\tau_2)}) - a_3 \tanh[(t - t_3)/\tau_3]\}T'_{\mathrm{iso}}, \tag{2}$$

where T'_{iso} is the moment generator with excluded muscular fatigue.

The parameters were identified from the measurements performed with intermittent stimulation where a train of pulses and the pauses lasted one second each [1]. On the basis of the comparison with curve (2) the following numerical values were obtained: $a_0 = 1$, $a_1 = -0.95$, $a_2 = 1$, $a_3 = 0.55$, $\tau_1 = 60$, $\tau_2 = 60$, $t_3 = 8$, $\tau_3 = 120$. Such a stimulation pattern to some extent corresponds to slow walking which is often exhibited by severely paralysed SCI subject.

Instrumentation

During experiments, a personal computer (PC) Pentium$^{\mathrm{TM}}$ with data acquisition board AD512 from Humusoft s.r.o. was used. As a programming tool, Matlab® with the Simulink toolbox from The MathWorks Inc. was selected. The program uses block programming with built-in mathematical functions and makes possible the implementation of the control and communication tasks via C written S-functions. The complete musculoskeletal model including muscular fatigue was realized using block programming.

The control lever of the telekinesthetic feedback system was directly connected to the electromotor (dc electromotor from Minertia® Motor, Yaskawa Electric Japan) providing the counteracting torque. An optical incremental encoder was used for position information of the control lever. The encoder output was in standard quadrature signal output form; A,B impulses, phase shifted by $\pi/2$, and index signal R, defining the home position of the control lever. These output signals were converted into a parallel 12 bit signal by external hardware.

Results

Six unskilled users were selected to test the designed telekinesthetic feedback system. They were asked to repeatedly actuate the control lever in a way similar to walking. The simulated gait was divided in two phases, swing and stance. During the swing phase, the simulated muscle was electrically stimulated for the duration of approximately one second. The same duration was used for muscle relaxation during the stance phase of the virtual walking. The computer displayed the time in seconds (figure 2). The subject triggered the train of pulses by moving the control lever. Discontinuation of FES was achieved by releasing the handle. The duration of each test was five minutes.

In the first test electrical stimuli of constant amplitude were applied to the simulated ankle dorsiflexors. The influence of fatiguing of electrically stimulated muscle is noticeable (figure 4 (*a*)). This pattern corresponds to the stimulation responses provoked by the presently used drop-foot electrical stimulators.

In the second test the visual feedback was included. The visual feedback was presented as the time course of the computer simulated ankle angle displayed on the computer screen while the telekinesthetic feedback was simultaneously providing the counteracting torque. The displayed diagram comprised the desired peak value of the ankle joint angle (the line in figure 4 (*b*)). The subjects were asked to keep the maximal value of the ankle angle as constant as possible (10° of ankle dorsiflexion). The inherent muscular fatigue made this attempt difficult. Because of the muscular fatigue, the ankle angle was decreased and correspondingly also the counteracting moment of the control lever. This second test was also considered a learning phase for the third test. The users were taught about the relation between ankle joint angle and the counteracting torque.

The third test presents the behaviour of the telekinesthetic feedback system alone. A walking subject should be able to control the ankle joint dorsiflexion and use the telekinesthetic feedback information without the visual feedback. A skilled user (after two hours of training) was able to control the ankle joint angle within a range of $\pm 2°$ during the cyclical stimulation pattern which is more than satisfactory regarding the requirements for a FES orthosis (figure 4 (*c*)). Table 1 presents the results gathered in six healthy persons using the telekinesthetic feedback only. The main goal was to maintain the maximal value of the ankle angle (10°) during the five minutes test as constant as possible. For every person an average value and a standard deviation were calculated. The test was performed in five trials to achieve higher statistical reliability.

Conclusions

The telekinesthetic feedback was found efficient for the elimination of the effects of muscular fatigue during simulated gait of spinal cord injured persons. The muscular fatigue decreases the effect of FES without the user's awareness. When using the telekinesthetic feedback the user becomes aware of the fatigue and can increase the FES amplitude to reduce its influence.

In this paper the idea of the use of telekinesthetic feedback was presented on the example of the ankle joint FES control. The telekinesthetic feedback could be applied in several other applications. One of them is FES assisted hand grasping [7]. Here the hand control is achieved by a control lever attached to the shoulder. In this case the telekinesthetic feedback could be applied to the grasping force.

The use of the telekinesthetic feedback is helpful not only when dealing with the fatigue of the electrically stimulated muscles. It can also diminish the variability in responses of stimulated muscles caused by inaccurate daily positioning of surface electrodes. Furthermore, the telekinesthetic feedback can be found useful when walking over uneven terrain.

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