

CONTROL OF FUNCTIONAL ELECTRICAL STIMULATION FOR RESTORATION OF MOTOR FUNCTION

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Abstract. *An injury or disease of the central nervous system (CNS) results in significant limitations in the communication with the environment (e.g., mobility, reaching and grasping). Functional electrical stimulation (FES) externally activates the muscles; thus, can restore several motor functions and reduce other health related problems. This review discusses the major bottleneck in current FES which prevents the wider use and better outcome of the treatment. We present a control method that we continually enhance during more than 30 years in the research and development of assistive systems. The presented control has a multi-level structure where upper levels use finite state control and the lower level implements model based control. We also discuss possible communication channels between the user and the controller of the FES. The artificial controller can be seen as the replica of the biological control. The principle of replication is used to minimize the problems which come from the interplay of biological and artificial control in FES. The biological control relies on an extensive network of neurons sending the output signals to the muscles. The network is being trained through many the trial and error processes in the early childhood, but staying open to changes throughout the life to satisfy the particular needs. The network considers the nonlinear and time variable properties of the motor system and provides adaptation in time and space. The presented artificial control method implements the same strategy but relies on machine classification, heuristics, and simulation of model-based control. The motivation for writing this review comes from the fact that many control algorithms have been presented in the literature by the authors who do not have much experience in rehabilitation engineering and had never tested the operations with patients. Almost all of the FES devices available implement only open-loop, sensory triggered preprogrammed sequences of stimulation. The suggestion is that the improvements in the FES devices need better controllers which consider the overall status of the potential user, various effects that stimulation has on afferent and efferent systems, reflexive responses to the FES and direct responses to the FES by non-stimulated sensory-motor systems, and the greater integration of the biological control.*

Key words: *functional electrical stimulation, neurorehabilitation, optimal control, finite state control*

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1. INTRODUCTION

An injury or disease of the central nervous system (CNS) (Fig. 1) leads to disability expressed with decreased sensory-motor performance (e.g., tetraplegia, paraplegia, hemiplegia, multiple sclerosis, cerebral palsy, etc.).

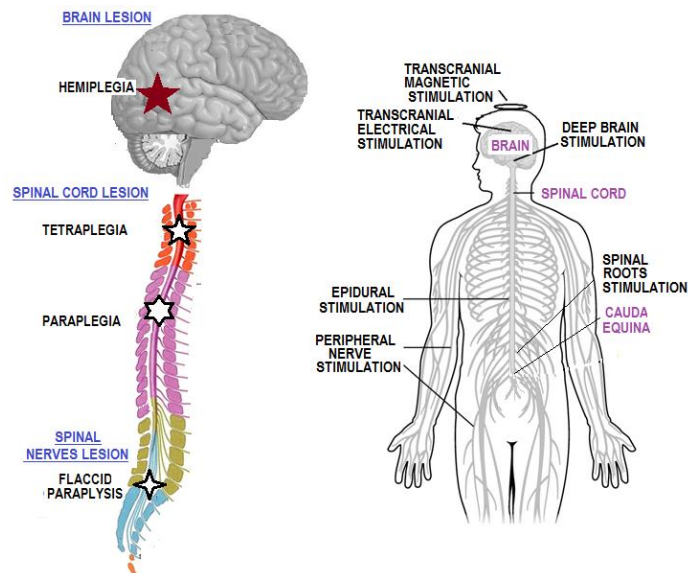


Fig. 1 The sketch of the central nervous system with the annotations of particular subsystems and types of disability (left panel) and the indication on possible sites of electrical stimulation (right panel).

The disability changes the lifestyle and results with other medical problems (e.g., muscle atrophy, contractures, frequent bladder infections, reduced cardiovascular capacity). Electrical stimulation (ES) is used for external activation of sensory-motor systems after an upper motor neuron lesion to decrease the disability (Fig. 1).

Fig. 1 shows the stimulation sites at the head (including the neck), spinal cord, and periphery. Some of the sites are above the lesion and in complete transections do not reach the parts that are paralyzed, and vice versa when the FES is applied at the periphery, and there is a complete spinal cord lesion only parts of the body are directly activated. However, in both cases in persons with incomplete lesion (likely about 90% of the total population), the stimulation can be seen above and below the lesion and has multiple effects as suggested in Fig. 2.

When the ES generates a function, then it is termed functional electrical stimulation (FES). A motor neural prosthesis (MNP) is the system which by employing the FES restores a motor function [1 - 25].

To regulate externally the activation of muscles and consequently the movements of the body parts one must understand the richness of natural mechanisms being involved in the operation. The reasons are the following: 1) muscles, tendons, ligaments, and joints differ from the man-made active kinematical chains; 2) the musculoskeletal system is

highly redundant; 2) the biological control system comprises a highly large number of nerves that serve as sensors, transmission channels to the muscles, and the decision-making circuits operating as a neural network physically defined to an extent by the interconnections that are hard wired [26], and partly the networks operating based on the unsupervised and supervised training. The external control must implement natural like model since they work jointly because only some parts of the body are paralyzed and require artificial external control, while the rest of the body is controlled by the central nervous system.

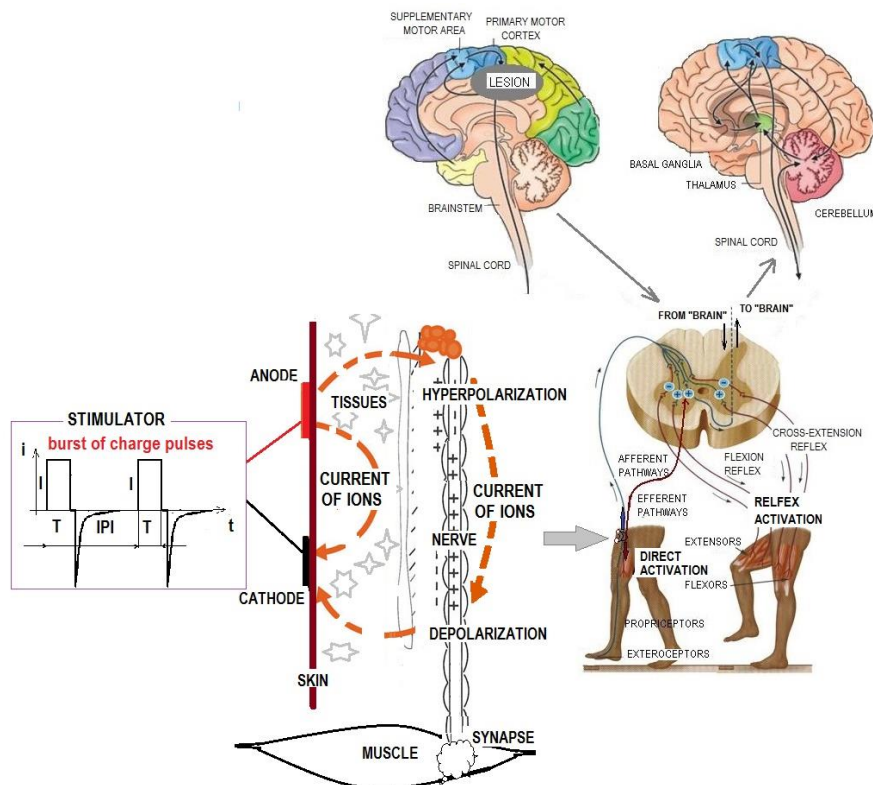


Fig. 2 The schema of the mechanism of FES of peripheral electrical stimulation. The stimulation activates afferent and efferent nerves. The afferent activity results with the reflex response from lower and upper motor neuron, while the efferent stimulation generates muscle contraction.

The hierarchical, self-organized biological control system relies on the extreme redundancy of the sensory-motor systems. Motion performance in a healthy person appears to be flexible and uncomplicated, although the neuronal operation which controls the system is still vaguely understood, even for the movements which comprise a small number of body parts [27]. The question which comes out to mind from the motor control studies is: Are the activities of the system components chosen randomly from the variety of possibilities to accomplish a given task, or there is a consistently reproducible pattern

of behavior that must be used? If the later, would it be possible to understand the constraints imposed to reproduce the same motor behavior whenever the motor task needs to be executed? A human can capture a spectrum of functional movements during the life. Most of the movements are mastered in early childhood; however, the repertoire is increasing and changing throughout the life, if so required. A functional movement relies upon perceptuomotor coordination that involves three main elements: the sensory information, the internal coding that is appropriate, and the generation of motion.

The author of this review assumes the model of control as a "black box" with an internal structure that is partly known [28 – 30]. The black-box approach follows many studies that the author proposed and carried out working with clinicians and motor control scientists. He started his research from the model-based control to design exoskeletons and hybrid assistive systems [31]. The clinical applications of these early developments made clear that the modeling is essential for studying the behavior and design components of the systems. Fig. 3 summarizes the effects of the FES applied to a peripheral nervous system. The modeling approach limits in most cases the control to the sketch of the stimulator and the efferent activation of nerves (Fig. 3, bottom left panel); while all other effects are not considered. The modeling approach carries the following: 1) over simplification of the real system; 2) no considerations of the time variability of the system; 3) not observable and measurable parameters of the model; 3) the target trajectory that is not known; 4) feedback which is not applicable due to the delays of the nerves, synapses and muscles; and 5) the fact that only parts of the system are controlled externally, while the remaining of the body is controlled by biological structures.

The black-box model of the FES must consider that the input to the user comes through a variety of sensory modalities which continuously updates information about the state of the system and the interactions with the environment. The black-box model needs to include that the sensory input triggers perceptual elements which assist selecting the decision what and how to execute the task. The information is processed at several structures of the brain in serial and parallel operations. The intention command (function selection) signal travels to lower centers of the cortex, midbrain, brainstem, and to particular segments of the spinal cord. The signals from the spinal cord in synchrony activate muscles via peripheral nerves and adjust the commands based on the afferent inflow [32, 33]. Muscles receive signals and produce forces which generate a relative movement of bodily parts, and due to the interaction with the environment the movement of the body in the space. The engineering description of this model is the following: 1) the control has a hierarchical structure with many parallel channels; 2) feedback tunes control at the lower and upper levels; 3) time delays characterize the operation and required a combination of the feedback and predictive control; and 4) the control needs to overcome the ambiguity of the redundant system.

In this review, we concentrate on the stimulation of peripheral nervous system (PNS). An FES generated bursts of short pulses of electrical charge triggers a series of action potentials in afferent and efferent neural pathways. The externally triggered efferent pathways directly activate muscles that are innervated by the said neurons; yet, not in the same manner as the volitional motor command would be coming from the upper motor neuron (Fig. 3). In parallel, the activity triggered in afferent pathways carry action potential to the spinal cord where various reflexes are generated (e.g., cross-extension reflex, flexion reflex), and interneurons are activated transmitting signals which eventually reach the cortex [15].

The FES delivers bursts of the electrical charge that are converted into the ionic currents in the tissues *via* electrodes. The time varying magnetic field can induce ionic currents; thereby

the electrodes could be replaced with magnetic coils. The stimulus waveform selected for the excitation process must take into consideration the physiological effect (action potential generation), potential damage to the tissue, and possible degradation of the electrode [34, 35]. The amplitude modulation (AM) or pulse width (duration) modulation (PWM) controls the level of recruitment [36 - 41]. PWM utilizes a slightly lower charge density compared with the AM to evoke a response of equal magnitude. Since the timing circuits (i.e., regulating pulse width) can be easily constructed and controlled with a resolution of 1 μ s or less, many stimulators implement the PWM. The typical duration of excitation pulses is in range to 250 μ s with implanted electrodes, but longer with surface electrodes. The threshold for excitation of the fibers of a peripheral nerve is proportional to the diameter of the fiber. Since the nerve is composed of a mixture of afferent and efferent fibers with a spectrum of fiber diameters, short pulses of constant amplitude will excite large afferent and efferent fibers. Longer pulses may also stimulate smaller fibers, including afferents typically carrying information of noxious stimuli, and therefore may be painful to the subject. For this reason and to minimize the electrical charge injection, short pulses are preferred. Recent research by our group resulted with multipad electrodes and the distributed stimulation which contribute to better selectivity, comfort for the users and substantially postponed muscle fatigue caused by the ES [42 - 44].

The regulation of the strength of a motor response depends on the number of activated nerve fibers and the rate at which they are activated. These two mechanisms are called recruitment and temporal summation, respectively. The same terms are utilized to describe electrically elicited events. When the stimulus is sufficiently large, an action potential will be evoked in the nerve. In a healthy person the slow, fatigue-resistant motor units are activated at a lower effort compared with the larger, fast fatigable units. In the FES the recruitment order depends upon the variables of position and geometry as well as fiber size [36 - 41].

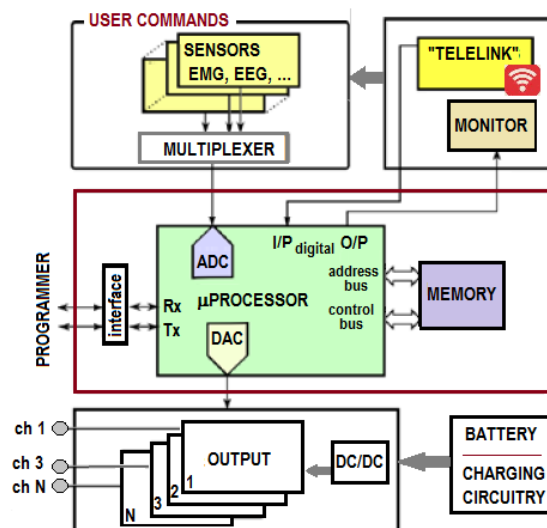


Fig. 3 Block diagram showing components for an FES stimulator. The acronyms used are: ADC – analog to digital converter, DAC – digital to analog converter, DC/DC – converter of the voltage from low to high level, I/P and O/P are digital input and output ports, EMG – electromyography, EEG – electroencephalography.

The second mechanism affecting the force developed by the muscle is the temporal summation. At low frequencies the response is unfused, and the variations of the muscle force are expressed. The frequency at which the mechanical responses produced are sufficiently smooth is known as the fusion frequency. In most human upper extremity muscles the fusion occurs at about 16 pulses per second. The unit Hz is often used instead of the correct term pulses per second. A maximum force (tetanus) can be reached at higher frequencies.

The fast development of electronics, microcomputers, wireless communication, and sensors makes the design of stimulators relatively straightforward. The requirements are the following: the system needs to be safe, small (portable), efficient, battery operated, and adaptable for various applications. The stimulator needs to control the frequency of pulses, pulse duration, pulse amplitude, the shape of the pulse and support several output channels. The stimulator needs to communicate with the user and use sensors signals including electrophysiological signals for the selection of the pattern of bursts delivered as particular channels. Battery supply and the charging circuitry are a must for an FES system. The block diagram of a stimulator is shown in Fig. 3.

The elements that need the attention in implantable systems are the biocompatibility, miniaturization and recharging circuitry.

2. CONTROL FOR FES-BASED MOTOR NEURAL PROSTHESIS¹

The multi-level control appropriate for the FES system which operates as a motor neural prosthesis (MNP) is presented in Fig. 4.

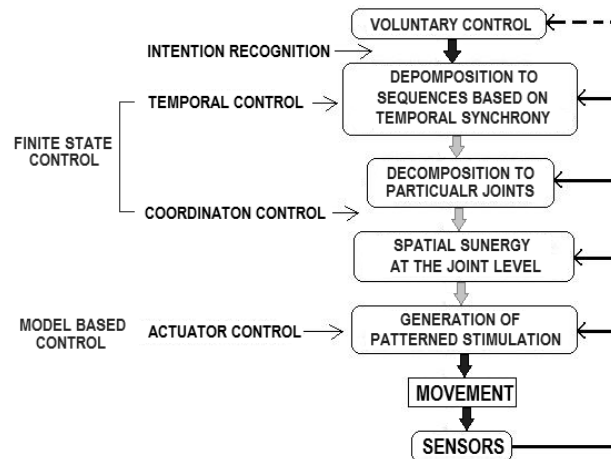


Fig. 4 The model of the hierarchical hybrid controller for FES aiming to restore the movement.

¹ The presentation uses the modified material from the publication by the author [30].

The intentions and decisions of the user are at the top of the hierarchical controller. The user communicates to the external controller decisions what and how he/she is planning [45]. The appealing method for the communication is a brain-computer interface (BCI). The BCI is these days one of the top research topics. Many techniques to extract signals are reported in the literature [46 - 56]. The interfaces for the BCI are not perfected yet (Fig. 5). To support this statement cite the conclusion from one of the most respected research groups in the world [46]: "Despite a growing animal literature demonstrating on-line control of functional hand movements from spike patterns recorded with microelectrodes in the motor cortex, BCI applications in neurological patients are rare and hampered by methodological difficulties. BCIs using EEG measures allow verbal communication in paralyzed patients with ALS; BCI-communication in totally locked-in patients, however, awaits experimental confirmation. Movement restoration in patients with little residual movement capacity using noninvasive BCI is possible, but a generalization of improvement to real life needs further experimentation". Events related recordings from the skull, correlated with a particular cortical activity, can be used as the trigger to start or stop an operation of the FES system [57].

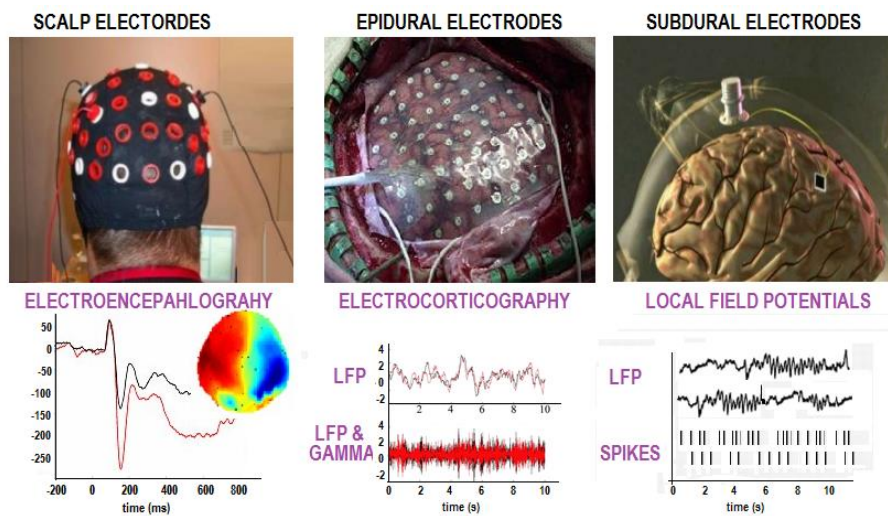


Fig. 5 Three interface types for the BCI: noninvasive (EEG, left panel), and invasive (ECOG and LFP, middle and right panels)

The implantable BCI systems provide much more detailed information compared with the skull recordings. The electrodes for the reproducible cortical recordings need to be improved. The positional instability of the electrodes is a critical issue. The longevity and reaction of the cortical tissues to the biocompatible materials need consideration. Implanting electrodes and electronic circuitry are rather invasive techniques. The recordings from the cortical electrodes will create eventually an extensive communication channel between the user and the microcomputer driving the FES or some other computerized device (Fig. 5, middle and right panels). The complexity of capturing motor commands from the brain can be easily seen from the reduced model of the flow of information within the brain,

cerebellum and brainstem (Fig. 6). Namely, the immediate question is where and what to record to allow the direct BCI?

The electrophysiological signals from the periphery (nerve or muscle) are an option to replace the higher centers CNS interface. The myoelectric control is widely used in hand prosthetics with success. There are various sites in persons with the CNS lesion that can be utilized. Surface electrodes could be used for recordings of muscle activities. Implantable electrodes are required for the interface with nerves. The plurality of electrophysiological signals can be used for multiple tasks MNP controllers. Some of the limitations listed for the BCI are valid for the implantable systems for the peripheral recordings and can be even more expressed.

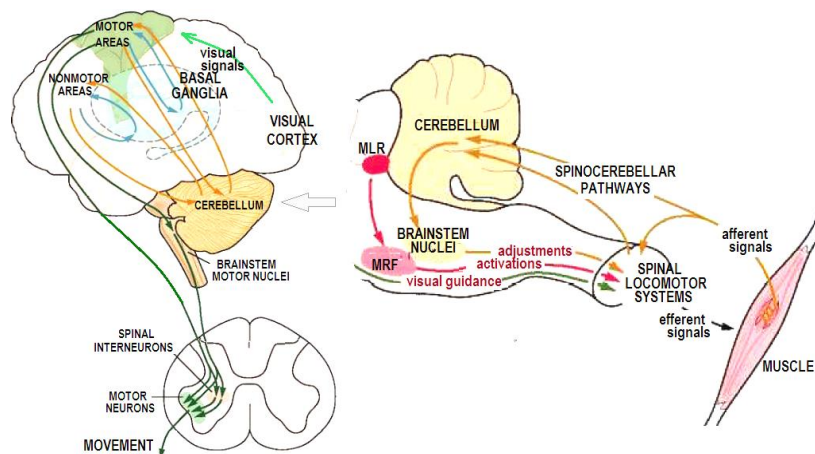


Fig. 6 The sketch of the flow of signals and major brain parts participating in the control of movement.

Switches and other computer inputs which utilize artificial sensors are the most efficient interface. The sensors measuring the acceleration, angular rate, joint angle, interface force/pressures when manipulated by the user provide a reliable and reproducible set of control signals. If the computer inputs are "mounted" on the body in a manner that allows subconscious sending of the appropriate command, then the multichannel control is facilitated. Recently, artificial visual perception [58, 59] was introduced as an input for the FES systems. It has been tested with success in artificial hand prostheses [60]. The artificial visual perception allows the planning that is ultimate interest for control (e.g., type of grasp, the position of the object with respect to the hand, size of the target object when grasping; curb, stairs, slope, obstacle, perturbation when walking/standing).

State control for coordination of movement. The two top levels of the control use the finite state control (FSC). The division to two levels is made to mimic the biological control where the temporal synchrony and spatial synergies are interplaying. The FSC implements non-analytical tools and non-parametric models of movement [61]. The FSC inherently deals with the following problems of movement control: 1) redundancy, nonlinearity, and time variability of the plant; 2) redundancy of plausible trajectories; and 3) the significance of the preference criteria based on the task.

Non-numerical tools are the identification techniques. In many cases, the identification is strictly based on heuristics. The heuristic procedures consist of choosing methods, which seem promising while allowing the changes if the originals do not lead quickly enough to a solution [61]. This procedure allows that the FSC learns from "mistakes" and improves the performance based on the acquired skill.

The FSC use set theory to define the behavior based on the states and rules to set transitions between these states. The states are movement representations in the multidimensional phase space (e.g., joint locked, joint free to move, flexion, extension expressed regarding muscle forces or joint kinematics). The transition from state to state is defined by rules. The rules are logical relations (e.g., IF-THEN, AND, OR) that connect state variables. A production rule is a state - action pair; *i.e.*, whenever a particular state is encountered, given on the left side of the rule, then the action on the right side of the rule needs to be executed. There are no a priori constraints on the forms of the states or the actions. A system based on production rules have three components: 1) the rule base, consisting of the set of production rules; 2) one or more data structures containing the known facts relevant to the domain of interest called facts bases; and 3) the interpreter of the facts and rules, which is the mechanism that decides which rule to apply to initiate the target action.

Each rule is an independent item of knowledge, containing all the conditions required for the application. Due to the modularity RBC systems can be changed by the addition, deletion or modification of a rule. An important feature of the RBC systems is the ability to look first at the established facts and use forward chaining, or to start from the task and implement backward chaining.

The problem of knowledge representation (determination of rules) is fundamental to the operation of the RBC. The rules these days are automatically generated through a procedure known as machine learning and classification (e.g., inductive learning, artificial neural networks, adaptive logic networks, fuzzy-logic networks, wavelet networks, Hebbian learning, stochastic classification techniques like principal component analysis, etc.). The data for the machine classification come are sensory patterns acquired while observing the process and the plant. The sensory patterns are coded (e.g., single threshold, multi-threshold, timing, local *vs.* absolute minimum or maximum), and the rules define the relationship between sensory patterns and required motor activities. A set of sensors providing feedback signals has been so far arbitrarily determined (e.g., ground reaction force or pressure sensor, switch, goniometer, inclinometer, accelerometer, and proximity sensor); the choice is based on availability of sensors, reproducibility of the sensory recordings, and overall practicality of plausible day to day usage. Sensors that are functionally equivalent to those used in biological control systems are preferred. Increasing the number of sensors produces very fast growth of the number of control rules making the definition process time consuming and difficult, yet very functional. Our group demonstrated and tested several systems that use the RBC for the control of the grasping, reaching and walking [62 - 66].

The FSC is not applicable directly to the FES systems because the on-off activation will result in jerky movements. Also, the time delays require a predictive rule base to allow that adaptation of the stimulation to the tasks. The FSC is a result of machine mapping, and even the high level of confidence close to 100% is no guaranty that a correct classification will follow since the hypothesis that the events are stochastic and independent is semi correct.

Model-based control for the FES. The FSC cannot be directly implemented for control because it ultimately creates jerky movements. To eliminate the jerks the physical

characteristics and properties of the system components must be incorporated. The method that is most appropriate is the model based control (MBC). The MBC considers the human body as a system of rigid bodies (skeleton) connected with rotational joints driven by joint actuators (muscles). The MBC considers the elastic properties of tendons connecting the muscles and bones, and the ligaments connecting the neighboring bones. The MBC for FES systems deals with two tasks: standing and walking and goal-directed movement (manipulation and grasping). The differences are the consequences of the tasks: walking is a near cyclic operation where the legs provide support for the trunk and propulsion, while the manipulation requires complex temporal and spatial activation to allow goal-directed movement. The muscle forces for the standing/walking are an order of magnitudes larger compared with those required for the arm movement and grasping. The walking introduces one tough task: balance. At present, there is no solution how to ensure balance while standing on a small surface (sole of a single foot).

The MBC is a highly complex task. The human body comprises more than 200 bones driven by about 600 skeletal muscles. The skeleton is presented with a spatial or a planar model comprising segments and joints with externally driven muscles. The FES drives only a subset of the whole system that is cut out from the biological control due to a CNS lesion. The only method to come to a solution is to use a reduced model. The reduction relates to the number of segments included in the analysis and the type of joints that connect the selected segments.

Fig. 7 shows the 12-segment model and the two-segment planar model of the leg. We present here the model of the planar model shown in panel B (Fig. 7) as an illustration of the complexity of the significantly reduced mechanical model of the body for the analysis of the walking [31].

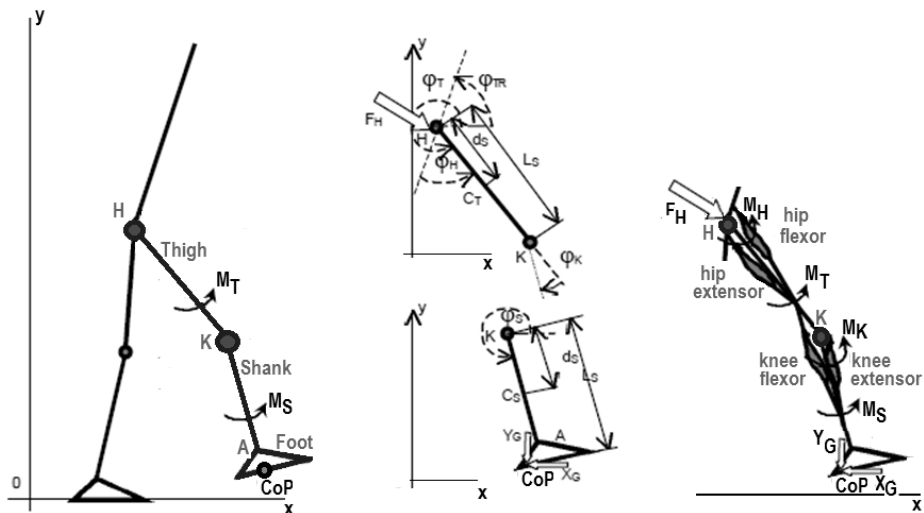


Fig. 7 The two-degree of freedom model of the leg. The middle panel shows the models of the thigh and the shank. The right panel shows the actuators (agonist and antagonist muscles) at the hip and knee joints.

The following notations are used: φ_K and φ_H are the knee and hip joints; φ_S and φ_T are the angles of the thigh and shank with respect the x -axes direction; m_S and m_T the masses of the shank and thigh; J_{CS} and J_{CT} are axial moments of inertia of the thigh and shank for

the axes perpendicular to the sagittal plane through the corresponding centers of masses; L_S and L_T are the lengths of the segments; d_S and d_T the distances between the hip and knee and the corresponding centers of masses. The torques M_S and M_T are the torques acting at the thigh and shank, while M_H and M_K are net joint torques at the hip and knee joints. The gravity direction is opposite to the y -axes.

The double pendulum representing the leg (Fig. 7) allows the knee and hip to flex and extend within the typical physiological range of movement. Two pairs of monoarticular muscles acting around the hip and knee joints (right panel in Fig. 7) drive the leg.

The second order nonlinear differential equations describing the model are:

$$\begin{aligned}
 & A_1 \ddot{\varphi}_S + A_2 \ddot{\varphi}_T \cos(\varphi_T - \varphi_S) + A_3 \dot{\varphi}_T^2 \sin(\varphi_T - \varphi_S) - A_4 \ddot{x}_H \sin \varphi_S \\
 & - A_5 (\ddot{y}_H + g) \cos \varphi_S - X_G L_S \sin \varphi_S + Y_G L_S \cos \varphi_S = M_S \\
 & B_1 \ddot{\varphi}_T + B_2 \ddot{\varphi}_S \cos(\varphi_T - \varphi_S) + B_3 \dot{\varphi}_S^2 \sin(\varphi_T - \varphi_S) - B_4 \ddot{x}_H \sin \varphi_T \\
 & - B_5 (\ddot{y}_H + g) \cos \varphi_T - X_G L_T \sin \varphi_T + Y_G L_T \cos \varphi_T = M_T
 \end{aligned} \tag{1}$$

where the coefficients are

$$\begin{aligned}
 A_4 &= m_S d_S, \quad B_4 = m_S L_T + m_T d_T, \quad A_5 = -A_4, \quad B_5 = -B_4 \\
 A_2 &= m_S d_S L_T, \quad B_2 = A_2, \quad A_3 = -A_2, \quad B_3 = B_2 \\
 A_1 &= J_{C_S} + m_S d_S^2, \quad B_1 = J_{C_T} + m_S L_T^2 + m_T d_T^2
 \end{aligned}$$

We used the notations for the angles to match the anatomical definitions of flexion and extension at the hip and knee. The net torques M_S and M_T generate the motion, and the torques M_K and M_H are a linear combination of the M_S and M_T . The simulation requires the following inputs: joint angle (trajectory), acceleration of the hip (body motion), and ground reaction force. The simulation needs the parameters that are geometrical and inertial parameters of the leg segments. The model presented is deterministic; hence there is a unique solution for the joint torques. The simulation result for one healthy person walking 0.9 m/s is presented (Fig. 8). The parameters for the model are from the literature [67, 68].

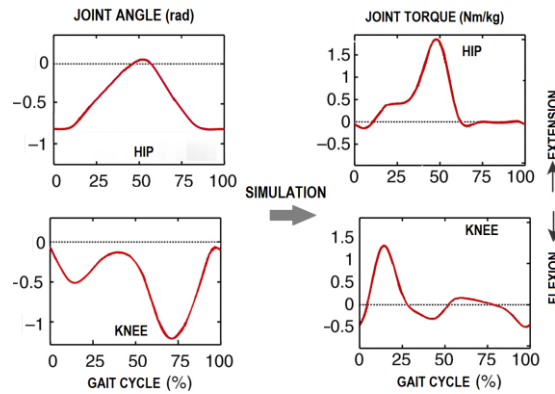


Fig. 8 The inputs for simulation (target trajectory) are the joint angles, ground reaction forces, and hip motion. Geometrical and inertial parameters are for a healthy young person. The speed of the walking was 0.9 m/s. Right panels show the output from simulation: joint torques at the hip and the knee.

However, in the FES system, the joint torque is generated by flexor and extensor muscles at joints as shown in Fig. 7, right panel. Two complex elements introduced are the redundancy, and nonlinearity and time variability due to the characteristics of the actuators and their attachments to the skeleton. The Eq. 1 now includes a more complex term replacing the pure torques acting at the high and shank (Eq. 2):

$$\begin{aligned} M_T &= T_{H,f} - T_{H,e} - T_{H,p} - T_{K,f} + T_{K,e} + T_{K,p} \\ M_S &= T_{K,f} - T_{K,e} - T_{K,p} \end{aligned} \quad (2)$$

The active flexor and extensor torques $T_{H,f}$, $T_{H,e}$, $T_{K,f}$, and $T_{K,e}$ are given by Eq. 3:

$$T_{m,n} = A(u)f(\varphi)g(\dot{\varphi}), \quad m = H \vee K, \quad n = f \vee e \quad (3)$$

The muscle response to electrical stimulation can be approximated by a second order, critically damped, low pass filter with a delay as shown in Eq. 4:

$$\frac{A(j\omega)}{U(j\omega)} = \frac{\omega_p^2}{\omega^2 + 2j\omega_p + \omega_p^2} e^{-j\omega\tau_d} \quad (4)$$

The nonlinear joint torque resulting from the contraction of muscles depends on the length of the muscle and the rate of the change of the length [69, 70]. The simplified model of this behavior is presented with the parabolic dependence of the torque from the joint angle and the hyperbolic fall of the torque with the increased joint angular rate (Eq. 5):

$$f(\varphi) = \begin{cases} a_0 + a_1\varphi + a_2\varphi^2, & F \geq 0 \\ 0, & F < 0 \end{cases}, \quad g(\dot{\varphi}) = \begin{cases} b_1, & \dot{\varphi} \leq 0 \\ b_1(1 - \dot{\varphi}/b_1), & 0 < \dot{\varphi} < b_1 \\ 0, & \dot{\varphi} \geq b_1 \end{cases} \quad (5)$$

The coefficients a_0 , a_1 and a_2 and b_1 are specific for joints and vary between uses for the same joint. The passive torques at the hip and knee is (Eq. 6):

$$T_{i,p} = c_{1,i}e^{c_{2,i}\varphi_i} - c_{3,i}e^{c_{4,i}\varphi_i} + c_{5,i} + c_{6,i}\dot{\varphi}_i + K_i\varphi_i \quad (6)$$

where the c_i and K_i are constants that are user dependent.

The redundancy can be resolved only by applying the optimization. The cost function needs to be formulated based on the tasks of simulation. The cost function can include the time, energy, force, torque, jerk, fatigue, muscle activation, non-physiological loading, the number of muscles used for the task, tracking error, and any combination thereof. Based on heuristics we suggest that the minimization of the tracking error and the muscle's efforts, with the preset level of co-contraction of agonist and antagonists (minimum stiffness) for the stabilization of joints and minimization of jerks.

To demonstrate the value of the model based analysis we show a result from one of our earlier studies [67]. The cost function is given by Eq. 7

$$R(u) = \int_{t_0}^{t_0+\Gamma} \left\{ \lambda_1 \left[\left(\frac{\Delta\varphi_S}{\varphi_{Sm}} \right)^2 + \left(\frac{\Delta\varphi_T}{\varphi_{Tm}} \right)^2 \right] + \lambda_2 [u_1^2(t) + u_2^2(t) + u_3^2(t) + u_4^2(t)] \right\} dt \quad (7)$$

The right panel (Fig. 9) shows substantial tracking errors and the recruitment level that is at the maximum. The direct conclusion is that the selected target trajectory is not achievable for the patient.

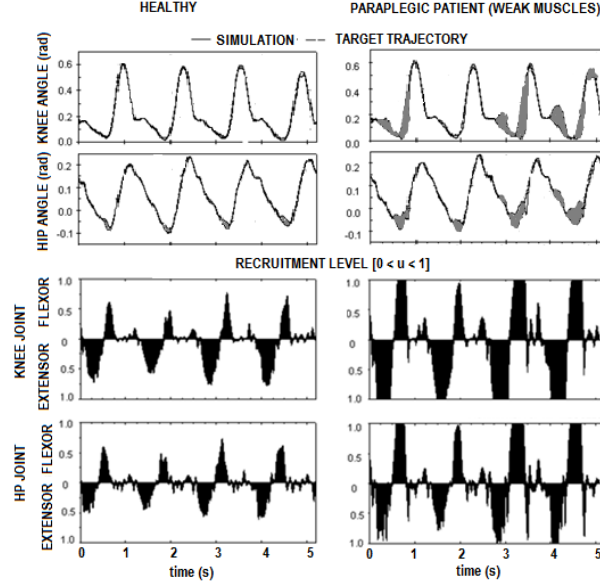


Fig. 9 The recruitment level for four consecutive steps. The recruitment reaches a maximum, and a substantial tracking error occurs for the paraplegic subject. The optimization does not consider the minimum stiffness.

We show the difference in the cost function (Eq. 7 and Eq. 8) affects the total cost.

$$R(u) = \int_{t_0}^{t_0+T} \left\{ \lambda_1 \left[\left(\frac{\Delta \varphi_S}{\varphi_{Sm}} \right)^2 + \left(\frac{\Delta \varphi_T}{\varphi_{Tm}} \right)^2 \right] + \lambda_2 [u_1^2(t) + u_2^2(t) + u_3^2(t) + u_4^2(t)] \right\} dt$$

and

$$u_i(t) > \lambda_3 \quad (8)$$

The graphs suggest the optimal conditions are when the effort is about 50%, and the tracking error is small (only a few degrees), while the lower values are obtained if the cost function does not consider the minimum cocontraction. The values on the axis of the graphs are normalized (Fig. 10), and we intentionally do not discuss this because of the scope of this presentation.

After discussing how a model is defined for the design of a controller we go back to the review of the literature. The simplest MBC operates without feedback (Fig. 11, open-loop). The controller is the inversed model of the plant [e.g., 71 - 74]. There are two major problems with this type of control: 1) the model is reduced in comparison with the real plant and parameters of the model are not reflecting the properties of the system adequately, and 2) the disturbances are not part of the model. The operation of this open-loop controller might be sufficient for simple systems; yet, in most cases, the system is not

robust enough [75 - 77]. An open loop controller uses the trajectory as the input. The term trajectory is used in broad sense (position, angle, velocity, acceleration, etc.).

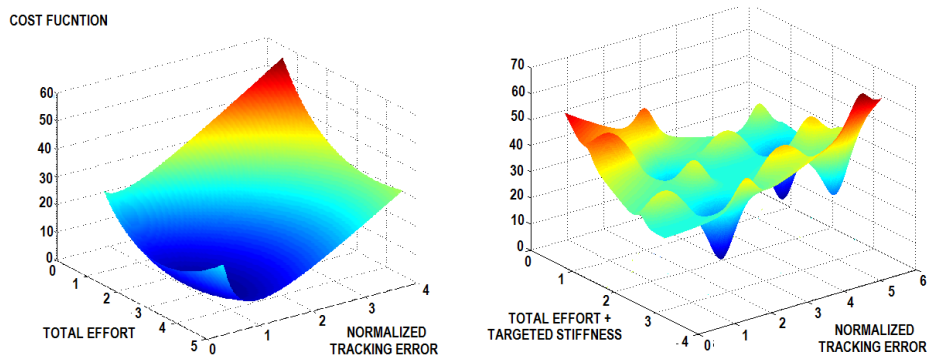


Fig. 10 The optimization surfaces for the two optimization functions (Eq. 3) for the two segmented model during the walking cycle at speed 0.9 m/s.

A closed-loop MBC (Fig. 11, close-loop), often termed "error driven control" uses feedback information from sensors measuring the achieved trajectory and corrects the command signal. The practical problem with the error-driven MBC is the biological delay; therefore, a predictive closed-loop control needs to be implemented. The term delay is related to the response of muscles (actuators) to the stimulation of the neural pathways. The response of the muscle to stimulation can be described with the low pass filter with the delay. The closed-loop control also requires a model that reflects the complexity of the organism being controlled and parameters which characterize the system. Also, the control requires precisely defined trajectory and the permitted errors that would not compromise the use of the NP.

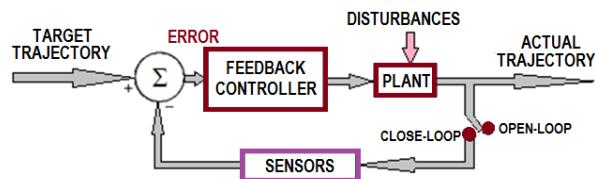


Fig. 11 Model-based control is operating without feedback (open-loop) and with feedback (closed-loop).

The important aspects which need to be considered when using the model-based control are the time variability of the responses of muscles (e.g., muscle fatigue, habituation, etc.) and nonlinearities.

The model based control is not applicable for multichannel electrical stimulation systems because the model is oversimplified, there is no preferred trajectory, the interaction with biological control is not included, the parameters of the systems change with time, the geometry of stimulation changes during the movement, reflexive behaviors are not included (dystonia, hyperreflexia, spasticity), some muscles are atrophied or even denervated, the

overall changes occurred in the central nervous system due to the lesion and modified periphery, stimulation activates afferent system that generates reflex responses (e.g., contralateral extension or flexion), stimulation is not selective enough, muscle fatigue could interfere due to the nonphysiological order of recruitment.

3. CONCLUSION

Control of FES for restoring movements as a highly complex task since it required full integration of the external components into the remaining and modified biological motor control. The plant is nonlinear and time variable, the system is not observable; thereby the stability of the system is questionable.

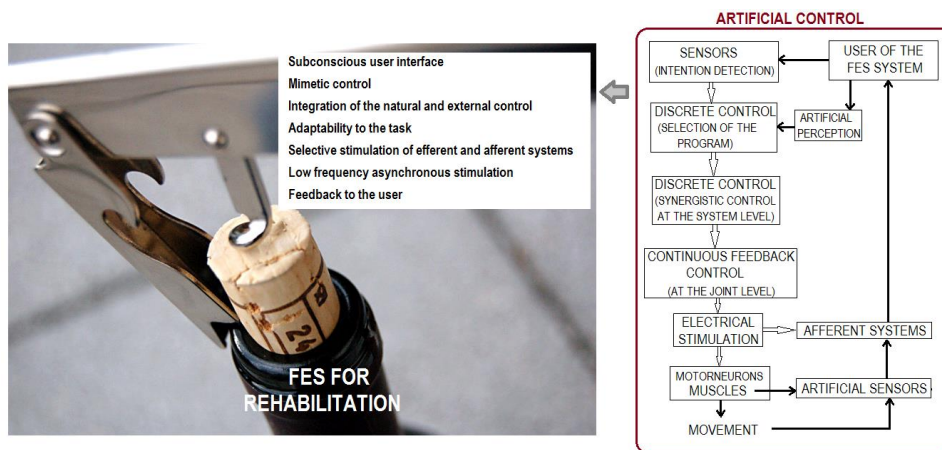


Fig. 12 The mimetic model of the hybrid hierarchical controller for control of movement using FES. The left panel shows the components that need to be integrated into the corkscrew to release the full potentials of the FES for rehabilitation.

We suggest that the concept presented and summarized in Fig. 12 has the best chance to contribute to the translation of the research into the clinical and home use by persons with disability as an orthosis or just a therapeutic device. Other aspects of the implementation of the FES are much more technological changes and cosmetic features. The interface to the user is also touched in this review, and the author is not convinced that invasive techniques which output a small number of commands are appreciated by the users and greatly increase the costs for the logistic support and original installation. The tests of the efficacy of the system with noninvasive interfaces should always be a step before considering the implant that improves the operation.

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