Chapter 11
Technical Rebuilding of Movement Function Using Functional Electrical Stimulation

Margit Gföhler

Abstract To rebuild lost movement functions, neuroprostheses based on functional electrical stimulation (FES) artificially activate skeletal muscles in corresponding sequences, using both residual body functions and artificial signals for control. Besides the functional gain, FES training also brings physiological and psychological benefits for spinal cord-injured subjects. In this chapter, current stimulation technology and the main components of FES-based neuroprostheses including enhanced control systems are presented. Technology and application of FES cycling and rowing, both approaches that enable spinal cord-injured subjects to participate in mainstream activities and improve their health and fitness by exercising like able-bodied subjects, are discussed in detail, and an overview of neuroprostheses that aim at restoring movement functions for daily life as walking or grasping is given.

11.1 Introduction

Injuries or diseases can interrupt the conduction of action potentials in the neural system. Depending on the kind and severeness, this may lead to complete or partial loss of control of the muscles of the lower and/or upper extremities.

The application of electrical stimulation in a rehabilitative setting was initiated in 1961, when W.T. Liberson, a physical rehabilitation specialist and medical researcher, developed a heel switch-triggered personal electronic stimulator device to correct foot drop [1]. Functional electrical stimulation (FES) aims to generate movements or functions which mimic normal voluntary movements and so restore

M. Gföhler (✉)
Research Group for Machine Elements and Rehabilitation Engineering, Institute of Engineering Design and Logistics Engineering, Vienna University of Technology, Karlsplatz 13, 1040 Vienna, Austria
e-mail: margit.gfoehler@tuwien.ac.at

P. Gruber et al. (eds.) *Biomimetics – Materials, Structures and Processes*, Biological and Medical Physics, Biomedical Engineering, DOI 10.1007/978-3-642-11934-7_11,

the functions which these movements serve. Devices that are delivering FES are a type of neuroprosthesis.

Reactivation of skeletal muscles and hence movement functions by FES may have impact on general life, reduce secondary health problems, and increase overall quality of life. Specific training with FES can cause significant improvements of the cardiovascular and pulmonary systems, reduce atrophy of skeletal muscle, increase bone density, and also lead to mental benefits ([2, 3]; Faghri et al. 1992).

11.2 Principle

Naturally, the signal for muscle contraction is generated in the central nervous system (CNS). This signal is propagated to and along the peripheral nerve and via the synapsis transferred to the muscle, where it induces the contraction. If this natural muscle activation process is interrupted by a lesion, the activation signals from the CNS cannot reach the muscles and consequently the muscles are paralyzed. FES is a method to artificially generate an activation potential in the peripheral nerve. A stimulator sends a stimulation pulse to the electrodes, and an activation potential is generated in the peripheral nerve and propagated to the muscle in the same way as in the physiologically intact body.

Figure 11.1 shows the main components required for a neuroprosthesis based on FES. Central element of an FES-based neuroprosthesis is the FES controller, which receives command signals and sensory input from artificial and natural sensors and controls the electronic stimulator. The stimulator generates stimulation pulses and induces muscle actuation via electrodes.

11.3 Actuation

The goal of FES is to stimulate the paralyzed muscles in as natural manner as possible. This requires that the muscles are activated selectively and produce reproducible graded forces. However, many poorly controllable factors related to neuromuscular anatomy and electrode placement make these goals difficult to achieve.

Muscle activation by means of electrical stimulation usually aims at generating an activation potential in the peripheral motor nerves that innervate the muscle, presuming that the peripheral motor nerves are not damaged. Also reflexes can be elicited by electrically stimulating the afferent nerves. Muscle fibers themselves are in principle electrically excitable but require very high stimulation intensities.

The stimulation signal for FES is generated by a programmable stimulator and transferred to the electrodes which transduce electron current into ionic current in the tissue. If the depolarization is strong enough, an action potential is induced in the nerve and propagated along the nerve fiber. This activation potential is then chemically transferred to the muscle fibers via the synapsis and induces muscle contraction and consequently the tendon force. The activating function $f(x,t)$ is

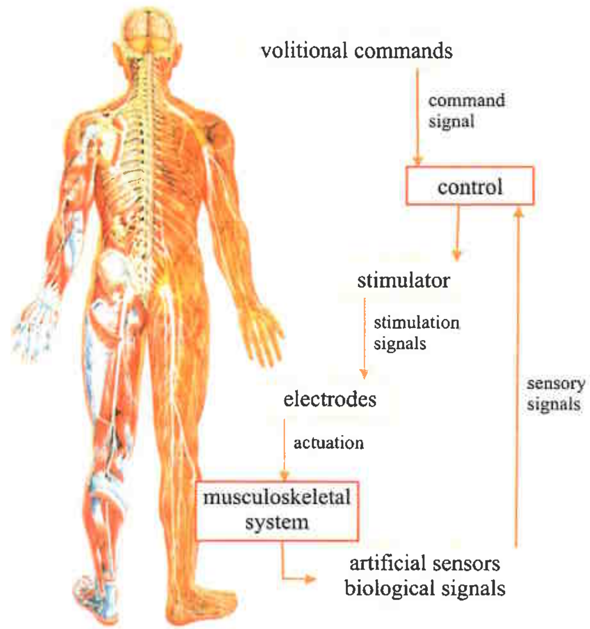

Fig. 11.1 Main components of an FES-based neuroprosthesis (figure of the human ©2010 3B Scientific GmbH, Germany)

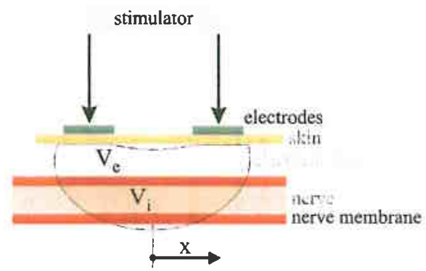

Fig. 11.2 Schematic of electrical field distribution in the tissue under surface electrodes

defined at each instant of time t as the second derivative of the extracellular potential V_e in direction x along the nerve fiber (Fig. 11.2) [4]:

$$f(x,t) = \frac{\partial^2 V_e(x,t)}{\partial x^2}.$$

11.3.1 *Stimulation Signal*

The stimulation signal usually consists of a train of biphasic rectangular current pulses with a frequency of between 0 and 100 Hz. Too low frequencies below the critical fusion frequency may lead to rippled muscle force output, and too high frequencies may increase fatigue [5]. For some applications, voltage-controlled pulses are used instead of current-controlled pulses; these are more easy to control but the current is the critical parameter that has to be above threshold level to depolarize the tissue and generate an action potential. The advantage of current-controlled stimulation is that if the resistance of the skin increases due to electrode drying out or sweating, the constant current stimulator will adjust automatically, whereas in a voltage-controlled stimulator the current flowing through the electrode has to be measured and the voltage adapted accordingly to achieve constant stimulation conditions. On the other hand, if an electrode loosens from the skin, the current density flowing through the remaining small contact area may increase to the level where skin damage can occur in a constant current stimulator. Generally, charge balanced pulse types are used so that no net charge is introduced to the body.

Parameters of the stimulation signal that influence the muscle force output are stimulation intensity and pulse frequency. The stimulation intensity can be varied by pulse amplitude and pulse duration, which is limited by the signal's frequency. Figure 11.3a shows a measured isometric recruitment curve (IRC), the relation between stimulation intensity (the pulse duration is varied at constant pulse amplitude) and isometric muscle force. If the stimulation intensity is higher than a threshold value, the force increases almost linearly until saturation is reached. Figure 11.3b points out that the isometric muscle force increases with stimulation

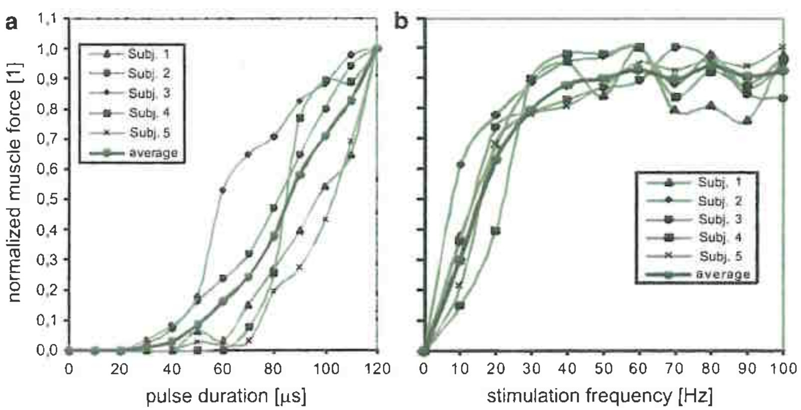

Fig. 11.3 (a) Isometric recruitment curve IRC, (b) relation between stimulation frequency and isometric muscle force, results of measurements on the Quadriceps muscles of five male paraplegic subjects and average [6]

frequency. The maximum force is reached at about 30 and 100 Hz for slow and fast contracting motor units, respectively. For frequencies above 50 Hz, muscle fatigue, which is a severe problem in FES applications, increases rapidly [5].

Muscle composition is changed in paralyzed muscle due to inactivity. The percentage of fast contracting, fast fatiguing muscle fibers increases, which is one reason for quick fatigue in paralyzed muscle. These inactivity-associated muscle changes can at least partially be reversed by FES training (Mohr et al., 1997).

In the case of physiological activation, first the thin, slow contracting motor units of the muscle are activated, and when higher forces are needed, bigger, fast contracting motor units are subsequently added. Similarly, the stimulation frequency is low at the beginning and raised for higher forces. The activated motor units are distributed over the muscle. In the case of artificial stimulation, the recruitment order is reversed to so-called inverse recruitment [7]. This means that big, fast contracting and fast fatiguing motor units are activated first. Additionally, motor units in the region of the muscle where the electrical field is stronger are activated first. Therefore, some parts of the muscle might be active while other parts are totally inactive.

11.3.2 *Electrodes*

Electrodes build the interface between the neural system and the technical device of the neuroprosthesis. A variety of interface concepts have been developed, ranging from simple wires to complex microsystems with integrated electronics. In general, selectivity increases with invasiveness.

11.3.2.1 Surface Electrodes

Surface electrodes are attached to the skin above a nerve or motor end plate. Their advantage is that they are noninvasive and easy to use. Disadvantages are the high influence of the electrical resistance of the skin and other tissues between electrode and nerve with respect to the distribution of the electrical field and the geometrical restrictions. It is impossible to reach deep-lying muscles without also stimulating overlying superficial muscles. High stimulation intensities are necessary, and it is difficult to predict which portions of the muscle are reached by the electrical field. One way to improve selective activation is to dynamically switch the cathode between sets of small transcutaneous electrode elements. Recently, novel embroidered electrodes have been used [8].

11.3.2.2 Subcutaneous Electrodes

Intramuscular electrodes are fine wire electrodes that are either inserted directly through the skin as percutaneous electrodes or tunneled subcutaneously. Percutaneous electrodes are less invasive than fully implanted electrodes, but positioning is difficult and relative movements can occur during movement. In addition, stress points occur where the wires cross the skin and at fascial planes between muscles. Frequent bending at these points can cause the wire to break, and the wire from the electrode comes through the skin providing a path for infection. Therefore, percutaneous electrodes are rarely used for long-term systems.

Epimysial electrodes are surgically placed on the muscle near the motor point. They consist of disk-shaped metals with a polymer shielding the surface away from the muscle.

11.3.2.3 Nerve Electrodes

Nerve electrodes are placed directly at the nerve – either adjacent, encircling, or intraneural.

Extraneural cuff electrodes consist of an insulating tubular sheath that encircles the nerve and contains two or more electrode contacts at their inner surface that are connected to insulated lead wires. The electrodes distributed around the circumference of a peripheral nerve are intended to activate different populations of axons. Cuff electrodes are easy to implant but may lead to nerve damage if their size is not well adjusted to the nerve diameter.

Intraneural electrodes are placed either longitudinally (Longitudinal Intra-Fascicular Electrode, LIFE [9]) or transversally (Utah Slanted Electrode Array, USEA [10]) in the peripheral nerve endoneurium and have higher recording selectivity and signal-to-noise ratio than extraneural electrodes. The LIFE is used for neural recording or stimulation small subsets of axons within a nerve fascicle. Typical records from LIFEs show multiunit activity where it is sometimes possible to resolve single units. LIFEs with 10 μm thickness and 50 mm in length have been realized using thin-film microfabrication techniques on polymer substrates, which also makes them more flexible and mechanically compatible [11]. The USEA is a silicon-based, three-dimensional structure consisting of a 10×10 array of tapered silicon electrodes that project out from a 4 mm $\times$ 4 mm substrate that is transversally inserted into the peripheral nerve for neural recording or stimulation. The lengths of the electrodes are graded from 0.5 to 1.5 mm along the length of the array to ensure that when it is inserted into a peripheral nerve, the electrode tips uniformly populate the nerve.

Sieve electrodes consist of a matrix of holes that is positioned at the end of a nerve. Ideally, the axons of the nerve will grow through the holes and build electronic contacts. Sieve electrodes have not yet been tested in human applications [11].

11.4 Stimulators

External stimulators: several multichannel programmable devices with analog and digital input and output lines are commercially available.

A commercialized *implantable* device is the eight-channel receiver-stimulator IRS-8, which receives power and control via an external close-coupled radio

frequency signal. It is used in the Freehand system® for active grasp and release. Based on the IRS-8, two implantable stimulator–telemeter systems (IST) have been developed which have additional input lines for sensory signals.

Loeb et al. [12] developed a fully *implantable wireless microstimulator*, the BION ("bionic neuron," 2 mm diameter $\times$16 mm long). Multiple BIONs can be injected through the barrel of a hypodermic needle near the nerve or neuromuscular junction of interest. Each BION receives power and digital commands from a telemetry link and delivers current pulses of the requested duration and amplitude via electrodes that are mechanically fixed on either end of its elongated capsule.

11.5 Control

11.5.1 Modeling/Simulation

For simple tasks, muscle stimulation patterns are developed by combining clinical experience with trial and error, but it is difficult to find smooth and energy efficient movements with trial and error because of the dynamic interactions between the segments. For more complex movements, muscle stimulation patterns have to be determined mathematically by establishing a dynamic model of the musculoskeletal system. This model usually consists of rigid body segments that are linked by joints and the musculotendon actuators. Due to the complexity and variety of the biological system parameter, identification is a main problem. Many parameters are difficult to access in vivo, and there are big differences between subjects. Additionally, in the case of physically disabled subjects, changes in muscle structure occur depending on the injury. Recently, MRI techniques have brought advancements in estimating musculoskeletal data of individual subjects.

To determine the impact of electrical stimulation on a movement, the resulting muscle force has to be determined. Muscle models with varying complexity are available [13]. Generally, the muscle force generation is divided into two processes, activation and contraction dynamics, as shown in Fig. 11.4. Both activation and contraction dynamics act as a low pass filter with the output responding slower and more smoothly than the input [14]. Muscle activation corresponds to the

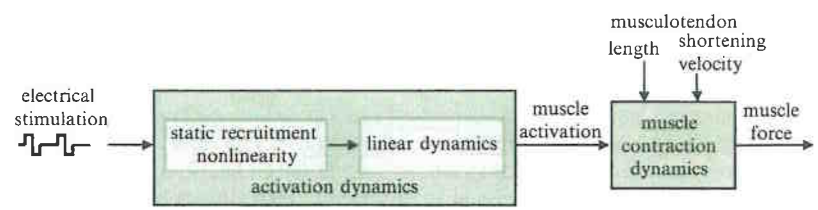

Fig. 11.4 Schematic of muscle activation and contraction dynamics

Ca-concentration and is described by the linear activation dynamics in the case of artificial activation by electrical stimulation. The static recruitment nonlinearity additionally accounts for the impact of stimulation intensity and stimulation frequency on muscle force according to the relations shown in Fig. 11.3. Muscle activation is slower and deactivation is faster in electrically stimulated muscle in comparison to physiological activation. From measurements on paralyzed leg muscles, a rise time of 108 ms was determined for 0–70% of maximal activation, and a fall time of 65 ms for 100–30% [15]. Muscle contraction dynamics describe the generation of force by activated contractile elements and basically shows the same behavior in healthy and artificially activated muscle. A muscle's force at each instant of time is a function of the instantaneous musculotendon length and shortening velocity, the tetanic muscle force, and muscle activation.

The muscle forces act on the body segments. The behavior of the musculoskeletal system is described by the equations of motion which are derived from the Newton–Euler equations. A system with n degrees of freedom (joint angles) has n equations of motion, which can be represented in vector form:

$$[\mathbf{A}].\underline{\ddot{\theta}} + [\mathbf{B}].\underline{\dot{\theta}}^{2} + [\mathbf{C}].g = \underline{M},$$

where $[\mathbf{A}]$ is the $n \times n$ mass matrix, $\underline{\boldsymbol{\theta}}$ is the vector of the system's n degrees of freedom, $[\mathbf{B}]$ is the gyroscopic matrix including centrifugal and coriolis terms, $\underline{\boldsymbol{M}}$ is the vector of joint torques directly due to muscle forces, and the term $[\mathbf{C}].g$ represents the torques due to gravity. As $[\mathbf{A}]$ is generally a full matrix, a muscle acting on one joint can accelerate all other joints of the system, this is called dynamic coupling.

To generate a defined movement by FES, it has to be determined which of the stimulated muscles have to be active in which phase of the movement and at which level. A forward dynamic model (Fig. 11.5) is used to calculate the resulting movement trajectories from muscle stimulation. Muscle forces and musculoskeletal geometry give joint torques, then the equations of motion are used to determine the joint angular accelerations, double integration finally gives the joint angles.

If the desired kinematics is known, inverse dynamics (Fig. 11.6) can be used to determine the optimal timing of the muscle forces. The joint torques are calculated with the system's equations of motion. As each joint is usually spanned by more than one muscle, there is a distribution problem when calculating muscle forces from

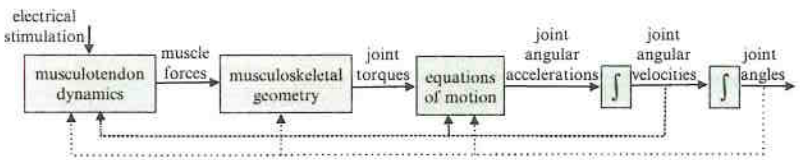

Fig. 11.5 Forward dynamic model

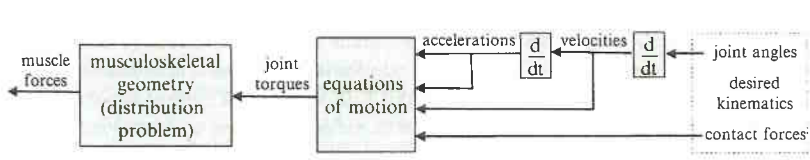

Fig. 11.6 Inverse dynamic model

joint torques. To split up the total torque to the single muscles, static optimization has to be applied. A commonly used performance criterion is to minimize muscle stress, squared, summed across all muscles [16]. As the static optimization does not consider musculotendon dynamics, the resulting muscle force trajectories may be irrealistic because a muscle cannot develop a force instantaneously.

To determine optimal stimulation patterns, muscle activation dynamics have to be considered. This is possible using a forward dynamic model where stimulation patterns are the input and the resulting movement is the output. A performance criterion applicable to the entire task has to be defined and forward dynamics combined with dynamic optimization methods to determine the optimal stimulation patterns. Parameter optimization methods as described in [17] are suitable to solve this nonlinear optimization problem, though these methods cannot differentiate between local and global maxima. Statistical methods as simulated annealing or genetic algorithms are likely to find global maxima but have the disadvantage of high computational expense. An alternative for estimating muscle forces and muscle excitations that requires three orders of magnitude less CPU time than parameter optimization is neuromuscular tracking [18].

11.5.2 Control Systems

Designing a control system that tracks a joint trajectory or torque profile by regulating the timing and levels of the electrical stimulation delivered to the muscles is a challenging problem due to the nonlinear response of the muscles to electrical stimulation and the complexity and redundancy in the musculoskeletal system.

Electrical stimulation may be delivered through either open or closed loop control systems. The FES controller attempts at taking over control tasks from the natural sensorimotor system and interfaces with the natural system at stimulation output and depending on the control approach also at command input.

In *open loop* control systems (Fig. 11.7a), the controller determinates the muscle stimulation according to the desired movement trajectory, and no information about the actual trajectory is fed back to the controller. The performance of open loop systems was found unsatisfactory for the generation of accurate movements, because external disturbances as obstacles or internal disturbances as variations in muscle force generation (e.g., due to fatigue) or inaccuracies in the model of the musculoskeletal system have impact on the actual movement trajectory.

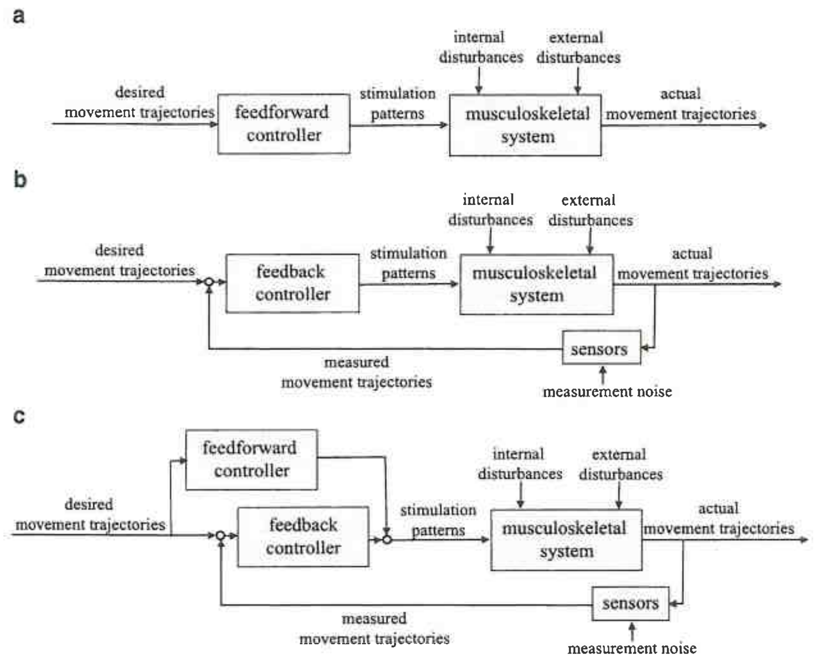

Fig. 11.7 Schematic of (**a**) open loop control, (**b**) closed loop control, and (**c**) hybrid control with both a feedforward and a feedback controller

For more complex or accurate movements, *closed loop* control (Fig. 11.7b) has to be established. During normal voluntary movements, the CNS receives sensory information on muscle-, tendon-, and cutaneous forces for neurophysiological control. In closed loop control systems, electrical stimulation is being initiated by the user's command and then modified based on some feedback measurement such as force or position. With closed loop control, the delivery of electrical stimulation is continuously modulated to control the parameter being measured by the sensors. The benefits of closed loop control are obvious, but closed loop control systems are more complex to design and implement. Furthermore, measurement noise or errors in the feedback signals can lead to unexpected behavior of the system.

For dynamical systems that are subject to both disturbances and measurement noise, hybrid control systems (Fig. 11.7c) combining feedforward and feedback control can improve the total performance [19]. Hybrid control systems have frequently been applied for control tasks of the musculoskeletal system [20–22].

In model-based control, a musculoskeletal model is directly used as the controller. An inverse dynamics model can be used as a forward controller with the desired trajectories as input and optimal stimulation patterns as output. Muscle

activation dynamics have to be linearized. Ferrarin et al. [21] designed a model-based feedforward control of the knee joint angle and combined it with a PID feedback controller. Jezernik et al. (2004) developed a sliding mode closed loop controller based on a musculoskeletal model for controlling the shank movement by FES.

The internal parameters of the musculoskeletal system may change because of internal disturbances such as fatigue-induced changes in muscle force generation. *Adaptive controllers* in which the controller parameters are allowed to adapt to changing plant parameters have been used to cope with such phenomena [23].

Due to their ability to map arbitrarily complex nonlinear input/output relationships from a given data set, *artificial neural networks* have been successfully applied to predict patterns of muscle stimulation needed to produce complex movements with FES-based neuroprostheses [20, 24–26]. EMG recordings from muscles under voluntary control and/or kinematic data have been used as input for the training of the neural controller [27].

11.6 Sensors

11.6.1 Artificial Sensors

Artificial sensors that are suitable for closed loop control of FES are force or pressure sensors. They are mostly placed at the point of contact, like ground contact in walking or grip force. Magnetic goniometers based on the Hall effect are used for measuring joint angles. DC accelerometers can be placed on the limb segments. Ambulatory position and orientation of human body segments can be measured accurately by combining an inertial measurement unit consisting of miniature gyroscopes, accelerometers, and magnetometers [28]. MEMS technology even allows incorporating accelerometers into injectable stimulation devices [29]. Most of the artificial sensors are placed externally on the moved limb, imposing further limitations on size, shape, and weight.

11.6.2 Natural Sensors in the Peripheral Nervous System

Natural sensors in the peripheral nervous system such as those found in the skin, muscles, tendons, and joints present an attractive alternative to artificial sensors for FES systems. Most of the peripheral sensory apparatus is still viable after injuries in the brain or spinal cord, yet not connected to the CNS.

Nerve cuff electrodes similar to the cuff electrodes for nerve stimulation have been used for chronic recording of *ENG signals* from sensory nerves. The ENG has a very small amplitude and a major source of interference is the myoelectric activity of nearby muscles. The EMG amplitude is approximately three orders of magnitude larger than the μV ENG and their spectra overlap. Implantable amplifiers have been designed, which, placed close to the recording electrode, remove EMG overlap. Control of FES thumb force using slip information obtained from the cutaneous electroneurogram has been used to control FES grip force [30].

Multicontact nerve cuff electrodes can work bidirectional by stimulating individual fascicles of nerve trunks and recording multiunit afferent activity from peripheral nerves [29].

The Utah Slanted Electrode Array USEA is inserted into the peripheral nerve for neural recording or stimulation.

11.6.3 Volitional Biological Signals

11.6.3.1 EMG

EMG is used to assess residual volitional motor activities. In so-called EMG-triggered stimulation, movement phases are initiated by volitionally activating the muscle whose EMG is measured. More sophisticated approaches establish closed loop control by modulating the stimulation intensity proportional to the measured EMG signal. EMG signals can be recorded by transcutaneous electrodes giving a noninvasive and relatively robust method for sensory input to the FES control. Any muscle the user can volitionally activate can be used for EMG recording. In the case of incomplete paralysis, it is also possible to record voluntary EMG from the same muscle that is stimulated. Bidirectional electrodes are available that can both record EMG and stimulate the muscle.

11.6.3.2 Brain Computer Interfaces

Brain computer interfaces (BCI) systems extract commands directly from the brain. The user imagines to perform a movement and brain activity signals are gained directly from the neuronal activity patterns in the corresponding motor areas of the brain. An advantage for the control of neuroprostheses is the fact that the imagined movement need not necessarily be the desired movement. Any type of command signal that is convenient for the user to generate can be used by the FES system. For example, foot movement can be imagined to trigger the FES system to open/close the hand. Due to the high inter- and intrasubject variability motor learning strategies have to be applied.

Noninvasive systems record the electroencephalogram (EEG) from the scalp or use functional magnetic resonance imaging (fMRI). The acquisition of high levels of control usually requires extensive user training. EEG-based BCIs are frequently used to trigger preprogrammed movements by FES-like hand grasp. Classifier functions are used to choose between two or more different brain states. These signals are then used like switches between different phases of a movement pattern.

Invasive methods use local activity from multiple neurons recorded within the brain. They show higher selectivity and are more successfully applied for complex control tasks but have the disadvantage of significant clinical risks and limited stability.

Electrocorticographic (ECoG) recording from the cortical surface has been tested as an alternative to current noninvasive and invasive recording methods [31].

First human pilot trials with both invasive and noninvasive systems suggest that BCIs could be a future option for the control of neuroprosthesis in patients with high-level SCI [32–34]. Still, there is little information available on the changes in the neural circuits in the brain after spinal cord injury, and optimal signal processing techniques have to be found to convert the existing brain signals efficiently and accurately into operative control commands.

11.7 Applications for the Lower Limb

Lower limb FES systems are used to restore walking [35], standing [36], sit-to-stand [37], cycling [38], and rowing [39]. Balance and the risk of falling are main problems in all upright body positions. As relatively high muscle forces are necessary for carrying the body weight, muscle fatigue is a serious problem in lower limb systems because it can cause falls and possible injury.

11.7.1 Cycling

Mobile FES cycling outdoors is attractive for paraplegics because they can use a standard bike (tricycle) with only a few modifications and move independently, powered by their own muscle force, over relatively long distances. Problems with balance are avoided by the seated body position, and compared to other types of movement, cycling has the advantage that the force applied to the pedal is converted into motion with very high efficiency.

FES leg cycling ergometry is frequently applied for muscle training in rehabilitation. The first commercialized leg cycling exercising system was ERGYS (Therapeutic Alliances Inc.) in 1984. So far, only external FES systems with surface electrodes have been used for FES cycling. For paraplegics, a number of leisure and sport activities are available, like basketball or hand-cycling, where only the intact upper extremities are activated. But the muscle mass of the upper extremities alone is not big enough to achieve oxygen consumption and heart rates above threshold, where the training is effective for reducing risk factors for cardiovascular and metabolic diseases. In comparison, the physiological benefits of the FES cycling training are relatively high [40].

Research on cycling by means of FES has been a focus of rehabilitation engineering at the Vienna University of Technology for several years. An instrumented

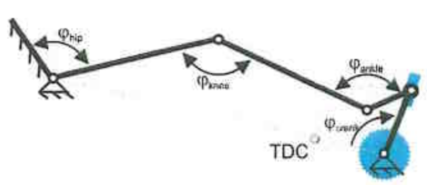

Fig. 11.8 Two-dimensional skeletal model [41]

FES cycling system has been developed [38] that serves as both a stationary cycle ergometer and a mobile tricycle for paraplegics.

11.7.1.1 Simulation

A forward dynamic simulation was established to optimize the stimulation patterns for FES cycling and to determine the influence of parameter changes.

A musculoskeletal model of paraplegic isokinetic cycling on a recumbent cycle was established [41]. The skeletal model is two-dimensional and effectively consists of five rigid segments connected in frictionless hinge joints (Fig. 11.8). These segments represent the crank, foot, lower leg, upper leg, and head-arms-trunk (HAT). The point of contact between foot and pedal is under the metatarsophalangeal (MTP) joint by default. Usually, in FES cycling the ankle joint is fixed by an orthosis that also stabilizes the leg. This means that the skeletal system has only one degree of freedom and consequently the leg kinematics are entirely determined by the imposed crank kinematics, and the muscle stimulation affects the forces but does not affect the kinematics. But as the power output in FES cycling is usually quite low and overcoming the dead center is sometimes problematic, it was investigated what effect releasing the ankle joint and additionally stimulating the muscles spanning the ankle joint has on the power output and overcoming the dead center. Releasing the ankle joint adds a second degree of freedom to the linkage and thus aggravates a control problem. Not only the force applied to the pedal but also the movement has to be controlled by the muscle stimulation. On the other hand, releasing the ankle joint and additionally stimulating the muscles spanning the ankle joint brings additional physiological benefits. The equations of motion of the skeletal system were derived from the Newton–Euler equations.

The skeleton is actuated by muscles/muscle groups of the lower extremity that are stimulated during FES cycling. For fixed ankle, these are Quadriceps (Vastii and Rectus Femoris receiving identical stimulation), Gluteus Maximus, and Hamstrings. For released ankle, in addition Soleus and Gastrocnemius (receiving identical stimulation) and Tibialis Anterior are included (Fig. 11.9). A Hill-type muscle model [42] is used to represent these muscles. It consists of a contractile element, a series elastic element, and a parallel elastic element. The latter element is present in the model but has no effect in the optimal solutions. According to measurements on

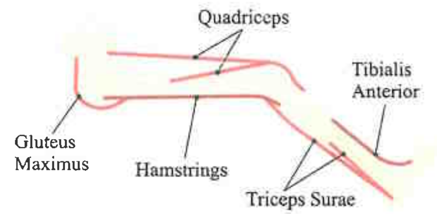

Fig. 11.9 Electrically stimulated leg muscles during FES cycling

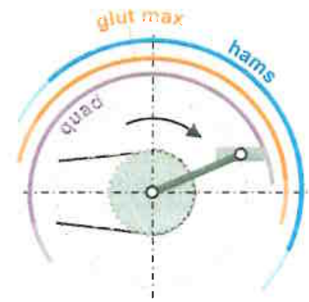

Fig. 11.10 Optimal stimulation pattern for fixed ankle joint for isokinetic FES cycling at 45 rpm. The light regions at the beginning and end of each stimulation interval indicate that the stimulation is switched on and off gradually along a ramp to avoid spasms and after-twitches

paralyzed muscles [15], the muscle activation and deactivation constants were set to 0.108 s and 0.065 s, respectively, based on the 0–70% rise time and 100–30% fall time for muscle force during isometric contraction and maximum isometric forces were set to 17% of the mean values for able-bodied subjects.

A forward dynamic simulation of isokinetic FES cycling at 30/45/60 rpm was performed. As optimization method, a parallel genetic algorithm [43] was applied. Input is muscle stimulation. The optimization criterion was to maximize mean mechanical power output over one full rotation of the crank. Figure 11.10 shows the optimal stimulation patterns for isokinetic cycling at 45 rpm, and the generated drive power is 90 W. For released ankle, the optimization results with the described model show that the ankle plantar flexors are unable to resist the torque of the pedal reaction force. To avoid this problem, it is either necessary to shorten the effective foot length by moving the position of the contact point of the foot sole and the pedal or to increase the maximal isometric force of the ankle musculature. Higher maximal isometric force of the ankle musculature might be realistic in many patients because spastic contractions reduce muscle atrophy. Shortening the effective foot length from 0.165 to 0.055 m resulted in a 10% power increase. Double maximal isometric force in the ankle musculature plus shortening the effective foot length