184 Impaired Motor Function: Functional Electrical Stimulation

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Introduction

Electrical stimulators to restore the rhythmicity of the heart (cardiac pacemakers) are currently being implanted in more than 400,000 people/year [1]. These stimulators provide patterned activation of the heart muscle to replace the natural pacemaker function. Similarly, patterned stimulation of cochlear neurons can provide remarkable restoration of hearing in completely deaf individuals and these cochlear stimulators have been implanted in more than 80,000 individuals worldwide [2]. Finally, thousands of stimulators have been implanted to restore micturition (bladder stimulators) [3,4]and respiration (phrenic nerve stimulators) [5]. These are all examples of "functional electrical stimulation (FES)" in that the stimulus provides a pattern of stimulation that directly generates a functional movement or sensation. This is in contrast, for example, to a deep brain stimulator where a constant frequency of stimulation is applied that modulates the activity in particular brain regions. The stimulation may enable functional movements to occur or block unwanted movements from occurring, but the applied stimulus pattern has no direct relation to the movements.

This chapter reviews the application of FES to restore limb movements in people who have a variety of CNS disorders (e.g., stroke, spinal cord injury (SCI), brain injury, multiple sclerosis). The number of people who could benefit from such an approach is potentially enormous. For example, there are more than 2 million stroke survivors in the U.S. alone and this disease constitutes the largest cause of disability in all developed nations [6]. Yet, the total number of FES stimulators applied worldwide is on the order of 10,000 [7]. In this Chapter we will discuss some of the issues that have limited FES applications to restore limb movements. We will also argue that many of these issues are soluble and that the numbers of FES applications to limbs may increase rapidly in the next few years. The following sections will consider basic issues related to electrodes, stimulating and recording methods. The final sections will discuss application of FES to restore movements of the upper and lower extremities (arms and legs).

Electrodes

Electrodes can be divided into three categories: (1) noninvasive surface electrodes, (2) minimally invasive electrodes and (3) fully implanted electrodes. Noninvasive electrodes are used widely for therapy, for example to strengthen muscles that have been weakened following disease or surgery. These electrodes typically consist of a flat conductor made of metal, carbon or carbonized rubber ranging from 1 to 25 cm^2 in area with an interface of hydrogel or moistened material that delivers electrical current from the stimulator and its attached wires to the skin overlying motor points. The gel or absorbent material interface traps water and ions that are dissolved in the water.

The current has both capacitive and resistive components. In electronics a capacitor is a device that has the ability or capacity to store charge. Electric charge delivered to the interface by the

stimulator attracts equal and opposite charge to the dermis from deeper lying structures. The capacitive component of current corresponds to the charge stored on the interface and dermis as the voltage changes. The other, resistive, component of current comprises charge that actually crosses from the interface through the dermis and epidermis to the underlying tissues. Resistive currents have a greater chance of producing allergic reactions to the substances crossing the boundary or causing skin breakdown, if the currents applied are sufficiently large. Simple water-filled electrodes that pass ions dissolved in tap water were used for over 2,000 patients with stimulators made in the former Yugoslavia [8]. Water-filled electrodes are less likely to cause an allergic reaction, but may harbor bacteria or other organisms. If they dry out, they can develop a higher resistance to passage of current than a hydrogel. If the area through which water flows is limited, this may increase the chance of skin breakdown. Finally, a standard electrode gel, such as that used in EEG or ECG measurements can be applied to lower the resistance. However, these gels often contain a highly concentrated salt solution that will irritate the skin if used for several h/day as is typical in FES applications. All commercial, surface FES systems currently use hydrogels to our knowledge.

A motor point is empirically found as a low threshold point for activating a particular muscle or nerve and is often near the neuromuscular junctions of a muscle. Even though the electrode is over a muscle all FES applications involve stimulating nerves. Muscle cells have a much higher electrical threshold. Though denervated muscles can be stimulated in experimental animals with very large, long-lasting pulses, these stimuli have not produced functional movements that could be used clinically [9].

Surface electrodes. The obvious advantages of surface electrodes compared to implanted electrodes are: (1) no surgery is required, thus reducing the associated costs, risks of infection,

postoperative pain, etc.; (2) a variety of placements and combinations can be easily tried to see which is most effective and (3) if the subject decides he/she doesn't want the stimulation or he/she recovers function through remission of the disease or regeneration, the electrodes are easily removed. However, surface electrodes have many disadvantages: (1) the greater the number of electrodes, the more tedious it is to apply them on a daily basis and the more variable the responses from day to day; (2) deep-lying muscles or nerves are difficult to activate; (3) passage of current through the skin may activate pain fibers in the skin and limit tolerable stimulus levels. Incomplete activation of an already weakened muscle may lead to inadequate functional movement.

Minimally invasive electrodes, such as percutaneous wires, are an attempt to combine some of the good features of surface and fully implanted electrodes. The simplest have a hook or barb ("tine") at the end and are commonly used for recording EMG or stimulating muscles selectively in a single experiment [10]. If implanted for longer periods of time in moving muscles, these electrodes will bend back and forth and eventually break. Forming the wire into a helix or providing other means of strain relief [11,12], as well as using improved materials, have increased longevity and the average time to breakage can be on the orders of months or even years [13]. In addition, since the wires emerge through the skin, a small, but significant risk of infection exists and care of the entry sites is an issue. Nonetheless, percutaneous electrodes offer a flexible method for trying out various potential FES approaches that could not be easily implemented if at all with surface electrodes. Successful approaches can then be implemented in systems that would be fully implanted at a later date.

An interesting example of a minimally invasive approach is the use of BIONs (❯ *Figure 184-1*). A BION is a microstimulator encapsulated in a glass or ceramic case that can be implanted via a

◘ Figure 184-1

Devices under development. (a) Original design of the BION microstimulator designed to be inserted through a hypodermic needle into a muscle or near a nerve of interest. The electrodes are at the two ends of the device and a coil inside the glass package is used to receive power and control signals from an external coil (not shown). American and Canadian pennies are shown to indicate the overall size. (b) Stimulus Router System (SRS) comprising implanted leads that pick up some of the current delivered by an external stimulator through surface electrodes. (c) Schematic of a complete SRS system for eliciting hand opening and closing, triggered by an earpiece sensor that detects small voluntary tooth clicks
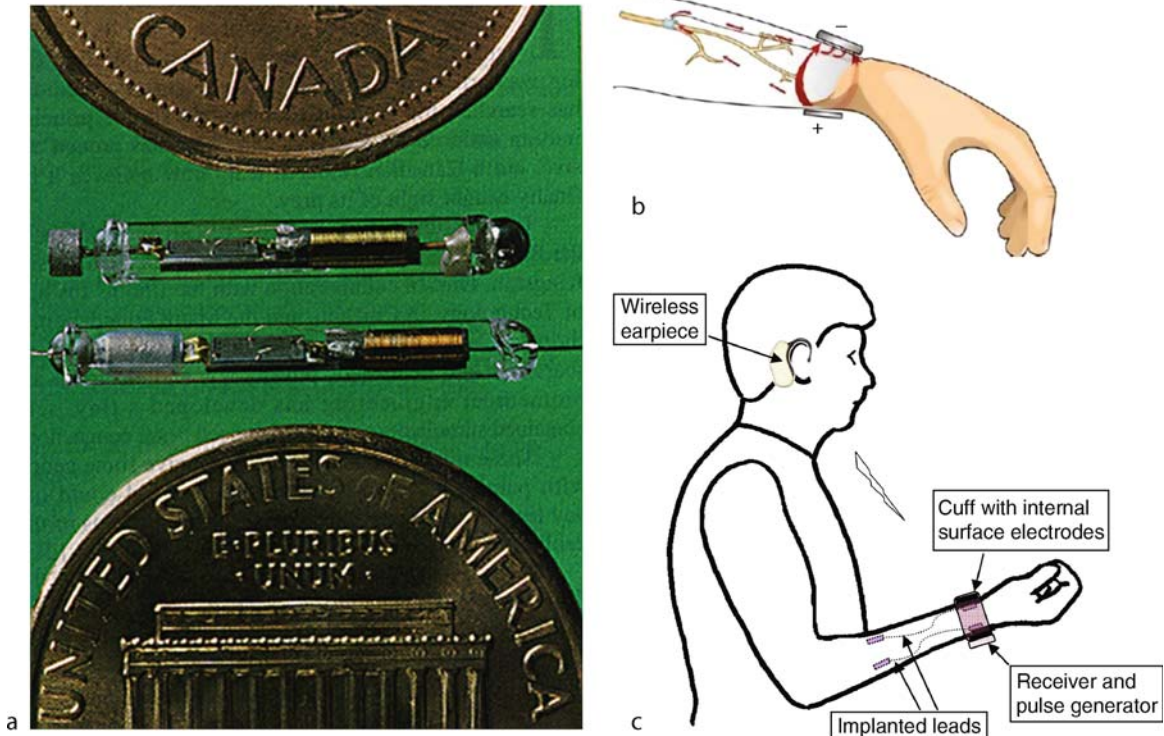
hypodermic needle [14,15]. Once the stimulation site is found the hypodermic needle is removed. The skin will seal around the puncture site so in effect the BION becomes a completely implanted device. However, we have included them in the section on minimally invasive approaches, since they do not require open surgery beyond that required to insert a hypodermic needle.

BIONs have been used in FES and other applications [16,17]. BIONs contain a coil that receives radio frequency signals that are decoded to produce a pattern of stimulation. One transmitter can send signals to a number of BIONs so in principle they can be used for quite complex FES systems. In the original design power as well as control signals were transmitted, and the efficiency of coupling is quite limited. The complexity of possible systems is limited by the number and placement of coils needed to communicate with and power the implanted BIONS. BIONs containing a rechargeable battery are being developed [18]. Without the need for transmitting power continuously, communication over longer distances is possible and a number of BIONs placed in various parts of the leg can be controlled by one central, external transmitter. However, the batteries would still need to be recharged periodically and the technical problems have not been fully overcome.

Fully implanted systems containing from 1 to 24 channels have been tested in human applications [19,20]. The simplest systems have been used for example to prevent foot drop, which can result from several CNS disorders and will be discussed in more detail below [21–23]. The most complex systems have been designed to restore walking and other functions in people with a complete, thoracic SCI. The 24-channel stimulator was a modified cochlear stimulator. However, in contrast to the cochlear application, where all the stimulation leads are localized in one place (the cochlea), wires were led from a pacemaker-like unit implanted in the chest to a variety of muscle and nerve locations in both legs. This involved three long surgeries and the project was discontinued after implantation in three subjects [20]. Also, the large volume and extent of the implant will increase the cost and risk of infection.

Alternative, fully implanted systems have been considered. For example, the area of the spinal cord controlling the leg (the lumbar enlargement) is small (about 5 cm in length in humans), compared to the lengths discussed above which may be a meter or more from a chest-based stimulator to the distal parts of both legs. In addition, motor pools (groups of motor neurons innervating a particular muscle) are arranged quite systematically in the ventral horn of the spinal cord [24]. Stimulating through one electrode may activate a synergy involving muscles spanning the hip, knee and ankle. As few as four electrodes can produce alternating flexion and extension of the lower limbs in a cat with low current intensities [25], but unwanted co-contractions often occur, particularly after some weeks [26,27]. Although promising, this approach, known as intra-spinal micro-stimulation (ISMS), has only been studied at present in animal experiments and numerous obstacles must be overcome before it can be considered as a clinical modality.

A novel approach to stimulating the nerves innervating muscles is the "Stimulus Router System" (SRS). It comprises an implanted lead that picks up some of the current delivered through the skin by a surface stimulator and delivers it to a target nerve via a nerve cuff. There are no implanted electronic components. Animal data have shown that the SRS can activate target nerves and muscles without stimulating local nerves under the surface electrodes [28]. A recent test during human peripheral nerve surgery showed that the SRS works similarly in humans [29]. The SRS has the advantages of an implanted stimulator: selectivity, reproducibility and convenience, at a lower cost, since only passive leads are implanted, the stimulator remaining external.

Stimulating and Recording Methods

Some of the issues associated with surface electrodes were discussed above. Implanted electrodes for FES applications have some of the same problems as those used for the other implanted stimulators discussed in the Introduction, but there are additional problems that will be discussed below.
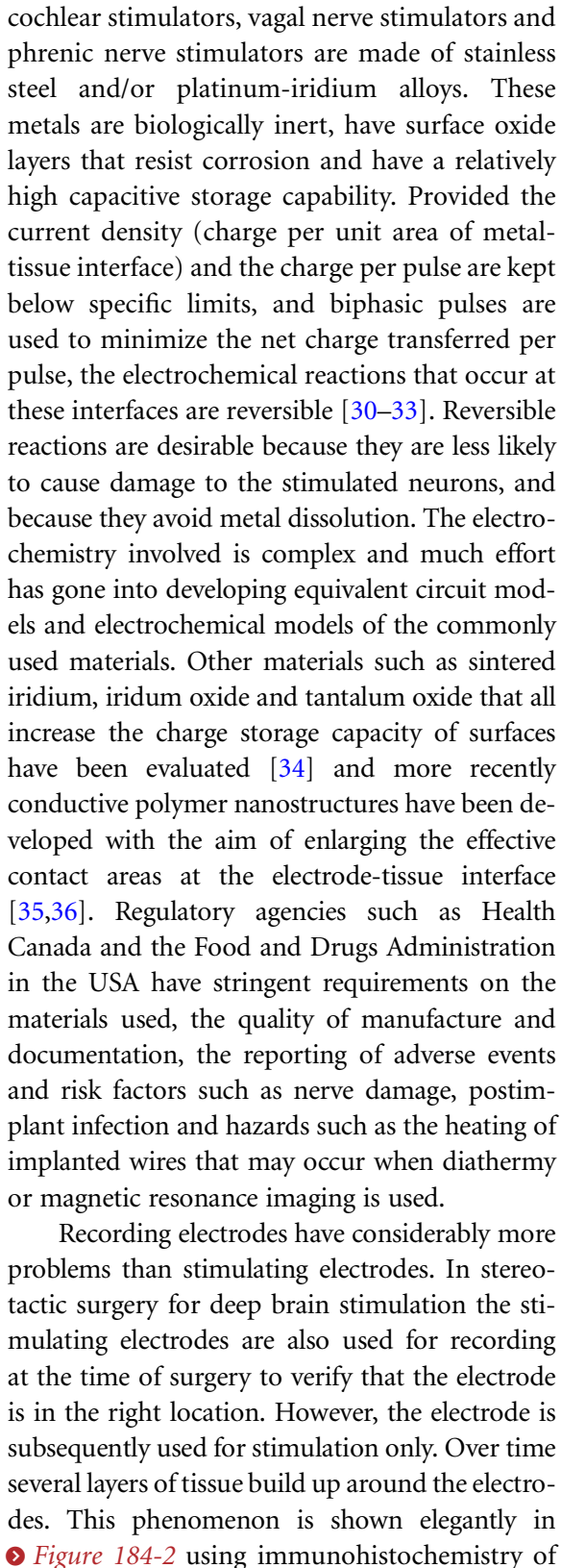
Classically, with reversible Ag/AgCl electrodes the reaction will be

$$Ag + Cl^- = AgCl + e^-$$
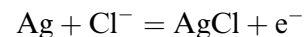
Supplying electrons from the negative pole of a battery will force the reaction to the left and Cl^- will come off that electrode. At the other, positive electrode AgCl will be formed. As long as there is still a coating of AgCl at both electrodes the reaction is reversible but if the coating is dissolved, the Ag will go into solution and the electrode itself will be dissolved. Thus, stimulation pulses should be completely charge-balanced to prevent the coating at one electrode and eventually the electrode itself from being destroyed.

In practice, the electrodes used in commonly implanted stimulators such as cardiac pacemakers,

cochlear stimulators, vagal nerve stimulators and phrenic nerve stimulators are made of stainless steel and/or platinum-iridium alloys. These metals are biologically inert, have surface oxide layers that resist corrosion and have a relatively high capacitive storage capability. Provided the current density (charge per unit area of metal-tissue interface) and the charge per pulse are kept below specific limits, and biphasic pulses are used to minimize the net charge transferred per pulse, the electrochemical reactions that occur at these interfaces are reversible [30–33]. Reversible reactions are desirable because they are less likely to cause damage to the stimulated neurons, and because they avoid metal dissolution. The electrochemistry involved is complex and much effort has gone into developing equivalent circuit models and electrochemical models of the commonly used materials. Other materials such as sintered iridium, iridum oxide and tantalum oxide that all increase the charge storage capacity of surfaces have been evaluated [34] and more recently conductive polymer nanostructures have been developed with the aim of enlarging the effective contact areas at the electrode-tissue interface [35,36]. Regulatory agencies such as Health Canada and the Food and Drugs Administration in the USA have stringent requirements on the materials used, the quality of manufacture and documentation, the reporting of adverse events and risk factors such as nerve damage, postimplant infection and hazards such as the heating of implanted wires that may occur when diathermy or magnetic resonance imaging is used.

Recording electrodes have considerably more problems than stimulating electrodes. In stereotactic surgery for deep brain stimulation the stimulating electrodes are also used for recording at the time of surgery to verify that the electrode is in the right location. However, the electrode is subsequently used for stimulation only. Over time several layers of tissue build up around the electrodes. This phenomenon is shown elegantly in ❯ *Figure 184-2* using immunohistochemistry of

an electrode implanted in a rat brain for 4 weeks [37]. Around the electrode is staining for ED1 in red, which is a marker for inflammation. Outside of this is a staining in green for GFAP, a marker for astrocytes and further out are stains for NeuN and NF which are markers for neuronal cell bodies and neurofilaments. The extent of these layers differs with different electrode types (e.g., silicon arrays or metal wires), whether the electrodes are tethered by lead wires or free to move, the amount of movement (e.g., electrodes in the spinal cord may be subjected to more relative movement than in the brain), the size and density of the array of electrodes.

The build up of these tissue layers and the reduction in neuronal cell numbers will clearly affect the ability to record from relevant neurons. It may not compromise stimulation, because the stimulus level can be increased to stimulate more distal neurons, but a loss of specificity may occur. Despite these potential problems arrays of 100 electrodes have been implanted in the cortex of human subjects [38] who have very limited motor function. Useful recordings have been made for a number of months. As will be described below, the aim of the experiments was to record neural activity in motor cortex and associated areas related to intended movement. These signals can then be decoded to produce movement of a cursor on a screen or to control movements of a robot or the person's own muscles through FES [38–40].

Leg Movements

Liberson et al. [41] first proposed the use of electrical stimulation to treat the condition of *foot drop* that occurs after stroke and other central nervous system conditions. In able bodied people cortical control of dorsiflexor muscles is relatively strong, so flexion of the ankle is often compromised when the cortex or its connections to the spinal cord are damaged. During the swing

▣ **Figure 184-2**

Immunohistochemical staining of the area around an implanted electrode. (a) Shows combinations of the individual stains for ED1 (an antibody that recognizes some leukocyte-associated molecules and is a marker of inflammation), GFAP (glial fibrillary acid protein, an astrocyte specific cell marker), NeuN (a stain for neuronal cell bodies) and NF (neurofilaments) that are displayed in (b) Note that neurons are displaced 100 μm or more from the site of the electrode (indicated schematically as an orange oval at distance 0), which will affect the ability to record from neurons, but may not prevent the ability to stimulate neurons chronically. From Polikov et al. [37]
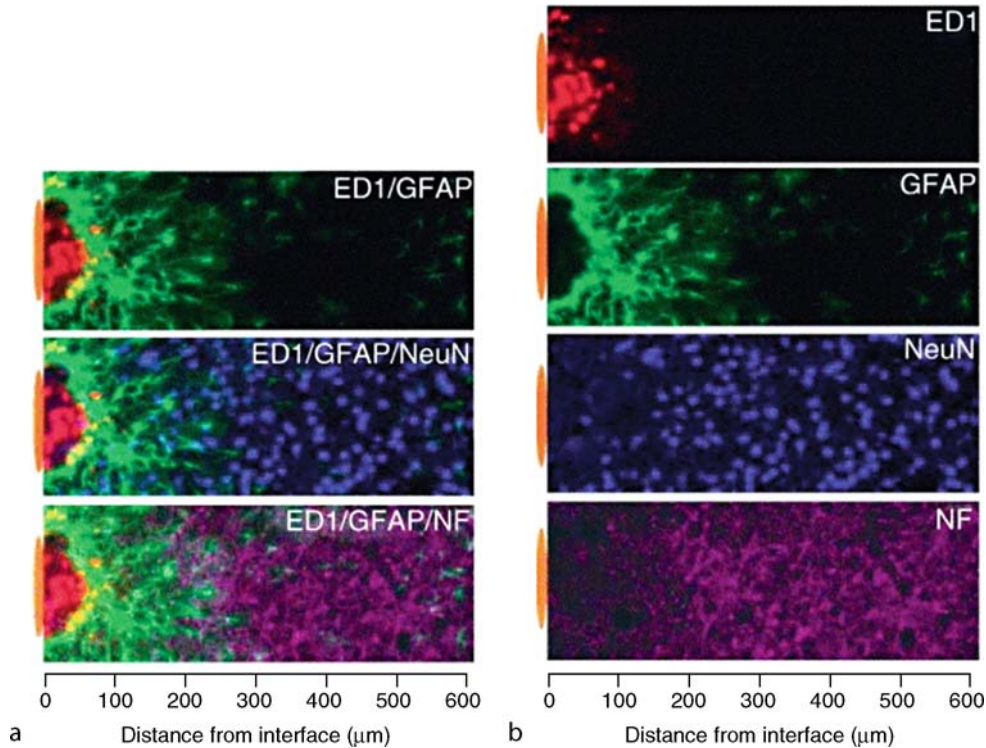
phase of the gait cycle the foot drops and may drag on the ground. Stimulation of the dorsiflexor muscles during swing will lift the foot and assist gait. Recently, the Food and Drug Administration (FDA) in the U.S. has approved several foot drop stimulators and these are available commercially in a number of countries: WalkAide (http://www.WalkAide.com), the Odstock Dropped foot stimulator (http://www.ODFS.com) and the L300 (http://www.Bioness.com). These systems all use surface stimulation, so no surgery is required. The WalkAide and Bioness L300 are shown in ❯ *Figure 184-3*.

Liberson et al. [41] used a heel switch for control of the foot-drop stimulator. Stimulation was turned on when the heel lifted off the ground and turned off when the heel touched the ground again. This system is still used in most foot-drop stimulators, but requires the heel switch to be placed in a shoe and the presence of wires (ODFS) or telemetry (L300) to send the signals to the stimulator. The WalkAide uses a tilt sensor that measures the orientation of the leg with respect to gravity [42]. When the leg is tilted back behind the body at the end of stance the stimulator is turned on and when the leg is tilted forward at the end of swing, the stimulator is turned off. The tilt sensor is incorporated in the stimulator package without the need for external wires or telemetry and can be used with any type of footwear including no footwear (bare feet).

⬡ Figure 184-3

The WalkAide foot drop stimulator (*left*) is about the size of an iPod unit (the size can be judged from the grey, bottom compartment that holds an AA 1.5 V battery). All the electronics including a tilt sensor for control are included in the upper compartment. The L300 foot drop stimulator (*right*) contains an electronics package and cuff on the leg, like the WalkAide, but has in addition an in-shoe wireless foot sensor and a remote control unit (not shown) that fits in a pocket or on a belt. Both devices have a remote clinician interface that allows parameters to be adjusted, and usage data and patient records to be stored, analyzed and printed

Implanted stimulators using open surgery [22,23,43] or BIONS [44] are also under development. The implanted devices offer the potential to have a more reproducible, balanced dorsiflexion during daily use. However, so far little or no additional functional benefit has been demonstrated in terms of speed or effort to walk with the implanted devices compared to the surface stimulation. Recently two types of implantable peroneal nerve stimulator have become available commercially in Europe, the Finetech STIMuSTEP (www.finetech-medical. co.uk) and the Neurodan ActiGait (www.neuro-dan.com). In a recent pilot study 15 individuals with footdrop due to stroke were implanted with the ActiGait system and showed improvements in gait [21]. Technical problems occurred with the stimulators, but these were resolved at follow-ups [21]. Like the original Liberson device, the STIMuSTEP and ActiGait stimulators are triggered from a heel sensor. A new innovation that is still at the experimental stage is to implant a nerve cuff around a sensory nerve and use processed nerve signals to decide when to turn the stimulator on and off [22,45]. The nerve cuff can be implanted together with the stimulating leads of an implanted foot-drop stimulator. A nerve cuff will record signals from all the large fibers in a nerve and is less sensitive to the growth of connective tissue, so it should provide a small, but more stable signal.

Careful comparisons will be required from large numbers of subjects using various surface and implanted devices. Indeed, a cost-benefit analysis is required, not only between implanted and surface stimulators, but between these two classes of device and an ankle-foot orthosis (AFO), a plastic brace that is most commonly prescribed for foot drop. An AFO passively holds the ankle in a neutral position, but has a number

of drawbacks. By bracing the ankle any residual dorsiflexion will be ineffective and the muscles may atrophy. In contrast, a recent study showed [46] that regular use of a foot-drop stimulator over a period of months increases the maximum voluntary contraction in the dorsiflexor muscles as well as the motor evoked potential (MEP) generated by stimulating the foot region of the motor cortex with transcranial magnetic stimulation (TMS). These results are in agreement with other recent studies [47–49] showing remarkable plasticity in cortical function produced by stimulation in adult humans. The mechanism of this plasticity is unknown, but appears to have much in common with long-term potentiation of sensory-motor circuits.

Many injuries and disease processes will affect a variety of muscles in addition to the ankle dorsiflexors. Training programs using modified treadmills that are able to support part of the body's weight [50–52] and robotic manipulators [53] are also effective in patterning the activity of various affected muscle groups and thereby improving walking after more extensive CNS damage. The training is labor-intensive and not effective for people with a complete SCI. Field-Fote et al. used several of these techniques [54] and they are now comparing their relative benefits for walking.

FES systems have been developed that provide limited stepping capabilities for people with complete paraplegia at a thoracic level. The simplest system involves stimulating the quadriceps muscles to lock the knee in extension during stance and stimulating the common peroneal nerve to produce a flexion reflex that brings the leg forward during swing. Four channels of stimulation can therefore produce a basic bipedal gait [55]. Adding additional channels of stimulation such as those activating the gluteal or paraspinal muscle can enhance upright stance [56]. One system has received FDA approval (Parastep; www.sigmedics.com). Fatigue is the limiting factor because of the high level of stimulation to the quadriceps muscle needed to prevent the knee from buckling. The walking is slow and limited to the order of 10 m because of the very high energy consumption.

Limited walking is also possible using existing braces such as the reciprocal gait orthosis (RGO). The RGO braces the ankle, knee and hip so a stiff-legged gait results. A cable links the two legs so forward movement of the trunk on one side leads to the forward movement of the leg on that side with respect to the other. The system is time-consuming to put on and take off and standing up is difficult. However, once the standing position is reached it can be maintained with little energy. An RGO also allows walking in people with complete thoracic spinal lesions with less energy than FES systems. Popovic et al. [57] first suggested the concept of a hybrid system that uses bracing to maintain an upright posture and FES for propulsion. Combining FES with an RGO can enhance the performance, compared to either system on its own [19,58]. New types of braces allow the knee joint to be locked and unlocked (stance control) and only control the joints from the knee down (knee-ankle-foot orthosis, KAFO). A stance control KAFO can be donned and doffed while sitting in a wheel chair and FES can be added to allow easier standing and propulsion for walking [59]. Since the hips are not controlled, trunk stability must be maintained by using the arms on a walker. About half the body weight is typically borne by the arms and fatigue of arm muscles becomes the limiting factor [59]. Several groups have developed systems using implanted electrodes to allow stimulation of more muscles and better control of the stepping movements (e.g., [11,20]). None of these systems has been commercialized and published data are based on a few intensively trained individuals. No studies directly compare the best implanted and the best surface systems, so the relative costs and benefits of surgery are still unknown.

As well as energy cost, control is a limiting factor in more complex systems. The greater the

level of disability the greater the need for sensory feedback, but many of the complex systems incorporate little or no feedback control. For example, control of the Parastep stimulator is by hand switches so control depends on the voluntary activity of muscles above the SCI. As well as initiating each step, the arms are maintaining balance with a walker or forearm crutches, since the legs are operating without feedback (open loop) and this contributes in part to the high energy cost to the upper body. Various types of sensors, in addition to those mentioned above have been proposed including force sensors, accelerometers, goniometers, and gyroscopes [60–62]. However, if the sensors are external, they have to be placed on a daily basis which adds variability and time to put the system on. In a hybrid system sensors can be mounted on the braces, as can the stimulating electrodes and this reduces the donning time and improves reliability. Still another approach is to place recording arrays in the dorsal root ganglia supplying the legs [63]. This provides access to a large variety of sensors and could be implanted together with an intraspinal microstimulation system. However, the system is not yet feasible for human trials due to problems in long-term signal viability (❯ *Figure 184-2*). Future work is clearly needed on appropriate surface or implanted systems to provide enhanced movements and some feedback control to respond to fatigue, external obstacles and other challenges.

Arm Movements

Vodovnik and colleagues in Ljubljana, Slovenia were the first to explore FES control of the upper limb [64,65]. In the late 1970s a therapeutic program for restoring hand function in stroke subjects was implemented at the Rancho Los Amigos Hospital in Los Angeles. Groups of participants performed FES-assisted biofeedback exercises daily [66]. In the mid-1990s, two surface FES devices were developed for quadriplegic people, the Handmaster [67] and the Bionic Glove [68,69]. A similar device, the ETHZ Paracare, was developed at the Swiss Federal Institute of Technology Zurich, Switzerland [70,71].

The Handmaster was marketed in Holland for several years and recently became available in the USA as the Bioness H200. It consists of a hinged splint with in-built electrodes and a separate stimulator. Stimulation is triggered by a push-button switch. The Bionic Glove is a flexible garment with an integral stimulator and electrodes. It is triggered by moving the hand into flexion or extension, boosting tenodesis grasp and release. Both devices have been shown to help restore hand function in quadriplegic people [68,72,73] when used for therapeutic exercise and training. The Bionic Glove was also used as an aid in activities of daily life by some people.

A new version of the Bionic Glove is being tested as part of a study involving in-home tele-rehabilitation in quadriplegic subjects in Edmonton [74]. The device, provisionally called the Hand-E-Stim, is a flexible garment with an integral stimulator the size of an iPod Nano. The Hand-E-Stim is triggered by a wireless sensor like a hearing aid that detects vibrations in the tissues in front of the ear. The user sequentially activates stimulation for hand opening and grasp with small tooth clicks [75]. The device will likely be available commercially in North America in 2009. The Hand-E-Stim has been designed to be used as an orthosis in activities of daily living, as well as for therapeutic training.

Coordinated FES of several muscles of the forearm and upper arm has been tested experimentally in individuals with SCI who have paralyzed elbow, wrist and hand muscles [76]. A programmable multi-channel surface stimulator was used to activate muscles in a sequence that allowed forward reach, grasp and flexion. One of the problems with surface stimulation of large muscles such as biceps and triceps brachii is that

during activity the motor points of these muscles can move several centimeters under the skin. This movement changes the relationship between the stimulating electrode and the motor nerve. Thus, the amount of muscle activation changes as the elbow flexes and extends, which results in problems of control. Nonetheless, encouraging therapeutic results were reported in this study.

An implanted stimulator, the Freehand System, was developed and tested in the late 1980s and 1990s at Case Western Reserve University (CWRU) [77]. The FDA approved it for commercial sale by Neurocontrol Corp in 1997. Freehand systems had been implanted in over 200 individuals by the year 2001. An external radio-frequency control unit activated an implanted device the size of a cardiac pacemaker, which then generated pulse trains and delivered them through epimysial electrodes to the targeted muscles. Signals from transducers monitoring voluntary shoulder or wrist movements were used to control the stimulation of the implanted muscles to produce a variety of hand movements. A multicentre study on 50 of the recipients showed that their hand function improved considerably while using the device. Unfortunately, the Freehand System was withdrawn from the market in 2002 for a variety of reasons that have been analyzed in detail in a Princeton University thesis [78].

A successor to the Freehand System, also developed at CWRU, has been implanted in seven SCI individuals [79]. This device is controlled by electromyographic signals picked up from muscles still under the user's voluntary control. The biceps muscle is activated as well as the muscles eliciting prehension. In a more recent report six subjects were implanted with a second-generation neuroprosthesis consisting of 12 stimulating electrodes, two EMG signal recording electrodes, an implanted stimulator-telemeter device, an external control unit and a transmit/receive coil [80]. Three of the subjects were monitored for at least 2 years. EMG signals

could be recorded from voluntary muscles in the presence of electrical stimulation of nearby muscles. All three subjects had significantly increased pinch force and grasp. At least five tasks in the Activities of Daily Living Abilities Test improved. Each subject was able to use the device at home. Given the ability to stimulate several muscles in the arm and hand in future systems, the control problem becomes significant. Musculoskeletal models of the whole upper extremity are being developed that will allow the synthesis of movements with electrical stimulation to be tested and optimized [39].

Another new development is the Finetech STIMuGRIP [81] which has been implanted in a number of hemiparetic people and the SRS described above, which was implanted in the first human recipient in June 2008. One of the important issues from a reimbursement point of view is the relative efficacy of implanted systems compared to surface stimulators. Implanted systems are more selective in stimulating the desired muscles, and have the potential to be more reproducible in their action from one day to the next, but they are also at least an order of magnitude more expensive. Therefore, their advantages must be analyzed in carefully designed comparative studies, preferably using quantified outcome measures. An important step in this direction is the recent analysis of cost savings of bladder, bowel and upper extremity neural prostheses [82,83]. The cost of the most expensive hand grasp neuroprosthesis would be recovered over the lifetime of the user if the time a personal caregiver was needed was reduced by just 2 h/day [83]. It will be interesting to analyze the time to cost recovery of simpler systems such as the StimuGRIP and SRS once more experience is gained with them.

As the level of a lesion becomes higher, more muscles need to be controlled, but the available voluntary control sites become fewer. One potential solution to this problem is to record neural activity from the motor cortex and use these

signals to control external devices or the person's own muscles through FES. These neural signals may be available even in people who can activate very few if any skeletal muscles. Donoghue and his colleagues [38] have used a 10 × 10 array of microelectrodes inserted in the motor cortex of such patients. Single and multi-unit neural signals could be discriminated from noise on many of the electrodes for a period of several months. Further, these signals could be decoded on-line and used to move a cursor on a screen to operate environmental controls, do email, etc. This is an interesting proof of principle, but the data rates are so slow that the system is unlikely to be accepted in its current form. Velliste et al. [40] reported on experiments using an array of electrodes in motor cortex of monkeys. The signals again were processed on-line and the monkey could use the signals to manipulate a robot to feed itself. Although still slower than the normal feeding movements in these intact animals, the speeds were fast enough that they would be practical for feeding or other tasks. Why better control was obtained in monkeys, compared to humans, remains uncertain. These developments are exciting, but many questions remain about the reliability, practicality and durability of such systems. The number of patients who can not use external systems is limited and the relative benefits of external versus implanted systems in relation to cost remains to be determined.

With the advent of more affordable and convenient implanted systems for less severely disabled individuals the future looks bright for implantable neural prostheses to restore motor function in the arms and hands. Similarly, as described in an earlier section, a range of options has recently become available for neural prostheses for leg function. Thus, although the field has developed more slowly than for heart pacemakers or cochlear prostheses, we are hopeful that widespread clinical acceptance of affordable neural prosthesis implants will occur in the next few years. These successes can then serve

as a springboard for development of more sophisticated and effective devices to assist with a range of arm and leg functions.

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