Department of Veterans Affairs

Journal of Rehabilitation Research and Development Vol. 33 No. 2, April 1996
Pages 145–157

Neural signals for command control and feedback in functional neuromuscular stimulation: A review

J. Andy Hoffer, PhD; Richard B. Stein, PhD; Morten K. Haugland, PhD; Thomas Sinkjær, PhD; William K. Durfee, PhD; Andrew B. Schwartz, PhD; Gerald E. Loeb, MD; Carole Kantor, MS
Cleveland FES Center, Cleveland, OH 44106-3052

Abstract—In current functional neuromuscular stimulation systems (FNS), control and feedback signals are usually provided by external sensors and switches, which pose problems such as donning and calibration time, cosmesis, and mechanical vulnerability. Artificial sensors are difficult to build and are insufficiently biocompatible and reliable for implantation. With the advent of methods for electrical interfacing with nerves and muscles, natural sensors are being considered as an alternative source of feedback and command signals for FNS. Decision making methods for higher level control can perform equally well with natural or artificial sensors. Recording nerve cuff electrodes have been developed and tested in animals and demonstrated to be feasible in humans for control of dorsiflexion in foot-drop and grasp in quadriplegia. Electromyographic signals, being one thousand times larger than electroneurograms, are easier to measure but have not been able to provide reliable indicators (e.g., of muscle fatigue) that would be useful in FNS systems. Animal studies have shown that information about the shape and movement of arm trajectories can be extracted from brain cortical activity, suggesting that FNS may ultimately be directly controllable from the central nervous system.

Key words: *FES, FNS, feedback control, functional electrical stimulation, functional neuromuscular stimulation, natural sensors, neural recording.*

This material is based upon work supported by the Department of Veterans Affairs, Rehabilitation Research and Development Service, Washington, DC; the Paralyzed Veterans of America Spinal Cord Research Foundation; the National Science Foundation; the Whitaker Foundation; the Bakken Research Centre; University of Miami; the United States Public Health Service; Canadian NeuroScience Network of Centres of Excellence; Medical Research Council of Canada; Danish National Research Foundation; Johnson and Johnson Foundation; National Science Foundation; Alfred E. Mann Foundation for Scientific Research; and, the Ontario Rehabilitation Technology Consortium.

Dr. Hoffer is Director of the School of Kinesiology at the Simon Fraser University, Burnaby, British Columbia; Dr. Stein is Director of the Division of Neuroscience at the University of Alberta, Edmonton, Alberta; Dr. Sinkjær is Director for the Center for Sensory-Motor Interaction, Department of Medical Informatics and Image Analysis at the Aalborg University, Denmark; Dr. Haugland is with the same department at Aalborg University; Dr. Durfee is with the Mechanical Engineering Department at the University of Minnesota, Minneapolis; Dr. Schwartz is with the Neurosciences Institute in La Jolla, California; Dr. Loeb is Director of the Biomedical Engineering Unit, Queen's University in Kingston, Ontario; and Ms. Kantor is with Tantalus, Inc., Highland Park, New Jersey.

This paper arose from a symposium that took place at the Engineering Foundation Conference "Neural Prostheses: Motor Systems IV" on July 23–28, 1994 in Mt. Sterling, Ohio.

Address all correspondence and requests for reprints to: Jeanne Teeter, Cleveland FES Center, 11000 Cedar Avenue, Suite 207, Cleveland, OH 44106-3052.

INTRODUCTION

Injuries to the spinal cord or some areas of the brain can cause permanent loss of sensation and voluntary motor function. Currently, regeneration of affected central neurons and restoration of functionally appropriate synaptic connections is not possible. However, even when connections with the damaged control centers are lost, the peripheral nerves and muscles can remain viable indefinitely. This makes possible the partial restoration of voluntary motor functions using functional neuromuscular stimulation (FNS), also called functional electrical stimulation (FES), of paralyzed muscles or their nerves, an approach that has been under development for the past 3 decades.

Current FNS systems provide a means for muscle activation that is controlled by external command signals and that involves very limited feedback. Control and feedback signals are usually provided by external sensors and switches, which pose problems such as, donning and calibration time, cosmesis, and mechanical vulnerability. Implanted sensors and means for conveying their data to external controllers have proven difficult to build with sufficient long-term biocompatibility and reliability. With the advent of methods for electrical interfacing with nerves and muscles (1), the various natural sensors having the potential to provide feedback and command signals are a realistic alternative (2). Feedback can be provided by signals recorded either from nerves (electroneurograms, ENG) or from muscles (electromyograms, EMG), as shown in **Figure 1.** Control signal sources can be derived from the brain, spinal cord, peripheral nerves, or muscles. This paper explores the possibilities for using direct recordings of nerve, muscle, and cortical signals for controlling FNS.

Natural Signals for Use in Feedback Systems

Open-loop FNS systems tend to overstimulate in order to ensure that muscles are sufficiently activated, but this leads to premature fatigue. If sensory feedback were available to adjust stimulation levels, both energy consumption and muscle fatigue would be reduced, and the safety margin could be increased. Feedback could also help alleviate the cognitive burden of open-loop systems, where users have to be closely aware of the level of system operation and must be ready to recognize emerging problems (e.g., slipping of grasped objects). Furthermore, feedback could increase the functional range, making system use possible under a wider variety of conditions (e.g., smooth and rough walking surfaces). The more varied the set of operational circumstances, the more important it is to have feedback that can correct for external or internal perturbations that affect the performance of a task.

Feedback is used very little in current FNS systems. Attempts to use force-sensing resistors in FNS walking systems have slowed the gait, and programmed time delays have been used instead (3). Based on normal EMG data and adjusted through trial-and-error, complex patterns of multiple muscle activation are preprogrammed to substitute for the natural generation of muscle activation patterns (4,5).

Feedback signals applicable to FNS systems may be either nerve or muscle signals from natural sensors, or electrical signals from artificial sensors. Among natural sensors, single neurons have not at present been demonstrated to provide reliable long-term recordings. However, ENG signals recorded from whole nerves with cuff electrodes (see later sections for details) showed greater shape regularity (**Figure 2**) than EMG signals in cats walking on a treadmill (6). A single nerve fiber can transmit 2–3 bits of information in 1 second. As the number of fibers increases, the information increases as the square root of the number (7). To study the effect of synaptic transmission on information processing, nerve impulses recorded from muscle spindle sensory fibers in response to length changes were filtered as would occur in transmission to motoneurons, or as might be done in an FNS system (8,9). As predicted, the filtered signals became progressively cleaner as the number of sensory fibers was increased. (**Figure 3**). Thus, large populations of sensory units are needed to achieve acceptable signal-to-noise ratios.

The relative size of EMGs (order of mV) and ENGs (order of μV) suggests that EMG might be a better natural signal (10). However, ENGs are generated by many more fibers, leading to a more favorable signal-to-noise ratio. Later sections of this paper deal in greater detail with the properties of both ENG and EMG signals for use in control and feedback.

Incorporating Natural Signals into Higher Level Control

To implement higher level control, one can use threshold crossings as criteria for deciding what to stimulate and when to change the level of stimulation. Functional movements can be broken into sets of states, and thresholds can be assigned to mark boundaries between states.

When designing FNS systems for specific functions, it is important to consider whether natural or artificial sensors are better. Neural signals have properties that are useful for control. For slip detection (in grasp or in ambulation), natural cutaneous sensors seem to be ideal (6,11,12). For other functions, they may not be as specific as other natural sensors; for example, muscle spindle signals contain a mixture of length and velocity information that depends on fusimotor tone and may monitor joint angles more closely.

Natural sensors may be more reliable than artificial ones, but a drawback is that their signals can only be obtained by invasive recording systems. On the other hand, the issue of cosmesis plays an important role in the choice of a sensor; on this basis, fully implanted systems are preferred. Contrary to common thinking, perhaps, the least important characteristic is linearity: well-behaved and repeatable sensory signals are not

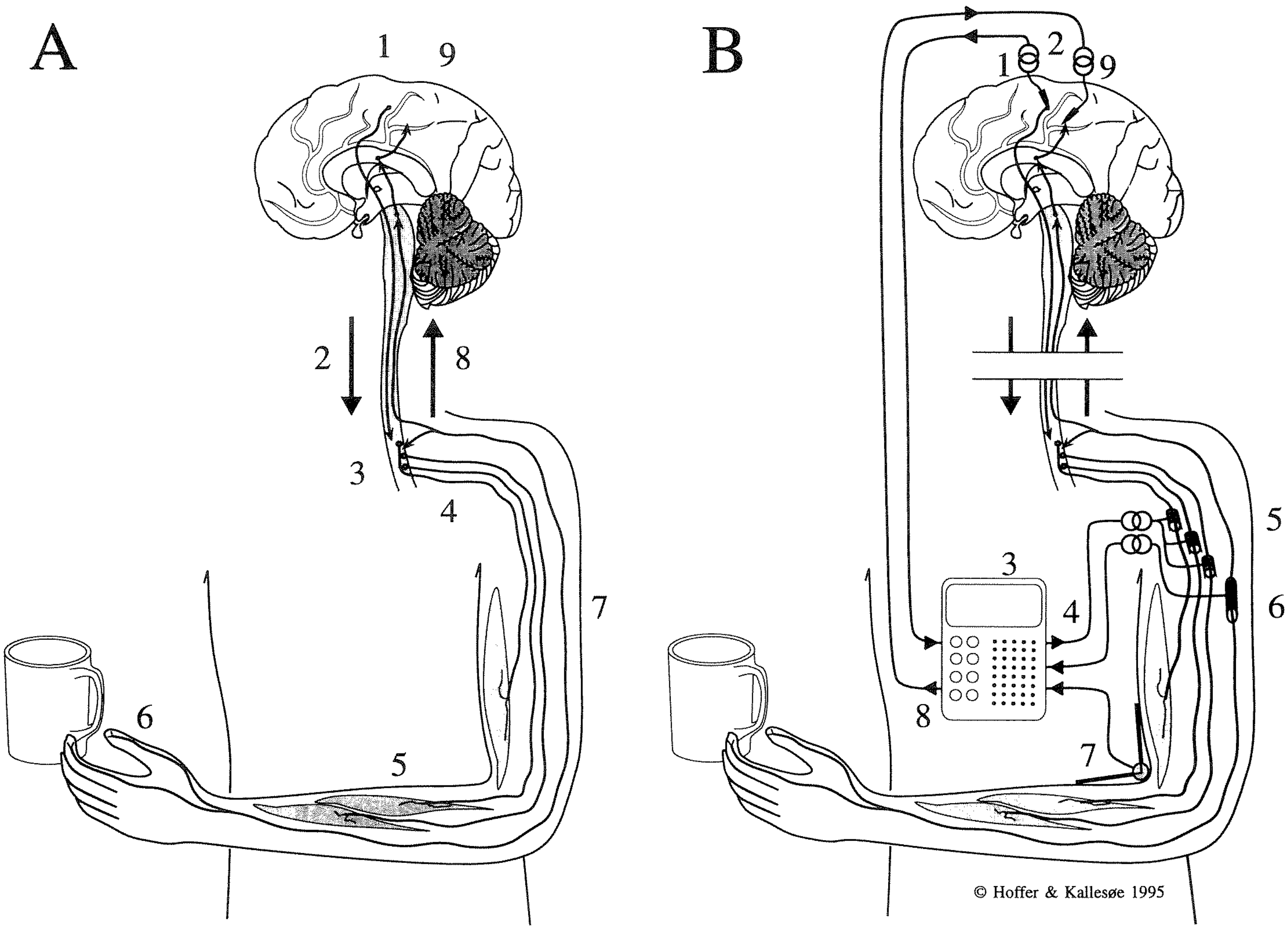

Figure 1.
Schematic diagram of central and peripheral nervous system components involved in the control of voluntary movements.
A. Voluntary action is initiated in various brain areas with the precentral cortex (1) being a primary output source. Cortical motor neurons project along the spinal cord (2) and synapse at various levels of the cord. Peripheral motor neurons have cell bodies in the spinal cord (3) and axons along the spinal roots and peripheral nerves (4) that terminate in neuromuscular synapses in various limb muscles (5). Sensory feedback arises from a variety of receptors, including cutaneous receptors in the fingertips (6). Sensory information is transmitted centrally along afferent nerve fibers (7) that project directly or indirectly onto the motor neurons (3) to produce spinal reflexes. Sensory cells also send ascending branches along the spinal cord (8) that, via relay neurons, project to various sensory areas of cortex, in particular the postcentral cortex (9). Connections among various areas of cortex and subcortical structures (brain stem, cerebellum, basal ganglia) participate in motor control loops. The control of voluntary movements includes sensory feedback from cutaneous afferents, proprioceptive afferents from muscles and joints, and visual, auditory, and vestibular sensory feedback.
B. Lesions of the spinal cord may destroy the axons that carry both descending and ascending information and thus permanently disconnect the body from the brain. This paper considers a variety of methods of interfacing with the central and neuromuscular systems that may be used to restore the voluntary use of paralyzed limbs and reestablish some sensation or perception. The diagram illustrates the recording of cell activity in the motor cortex (1). This signal is telemetered (2) across the skin and fed to an FES controller (3) that is programmed to detect command signals from it. The controller provides the moment-to-moment amounts of stimulation to various muscles as output (4) that is telemetered to appropriate motor nerves or intramuscular electrodes (5). Touch and position information that is generated by cutaneous or proprioceptive receptors travels along large sensory axons and can be recorded with nerve cuff electrodes (6), fed back to the controller and used for closed-loop control of FES. Additional or alternative sensory signals may be obtained with external (7) or indwelling transducers (e.g., goniometers, accelerometers, Hall effect transducers). The controller may also substitute sensory feedback (8) to appropriate areas of the sensory cortex with microstimulation (9) to provide the user a form of sensation about the ongoing movement or motor task. In these regards, the FES controller replaces both the integrative and the communicative functions of the spinal cord.

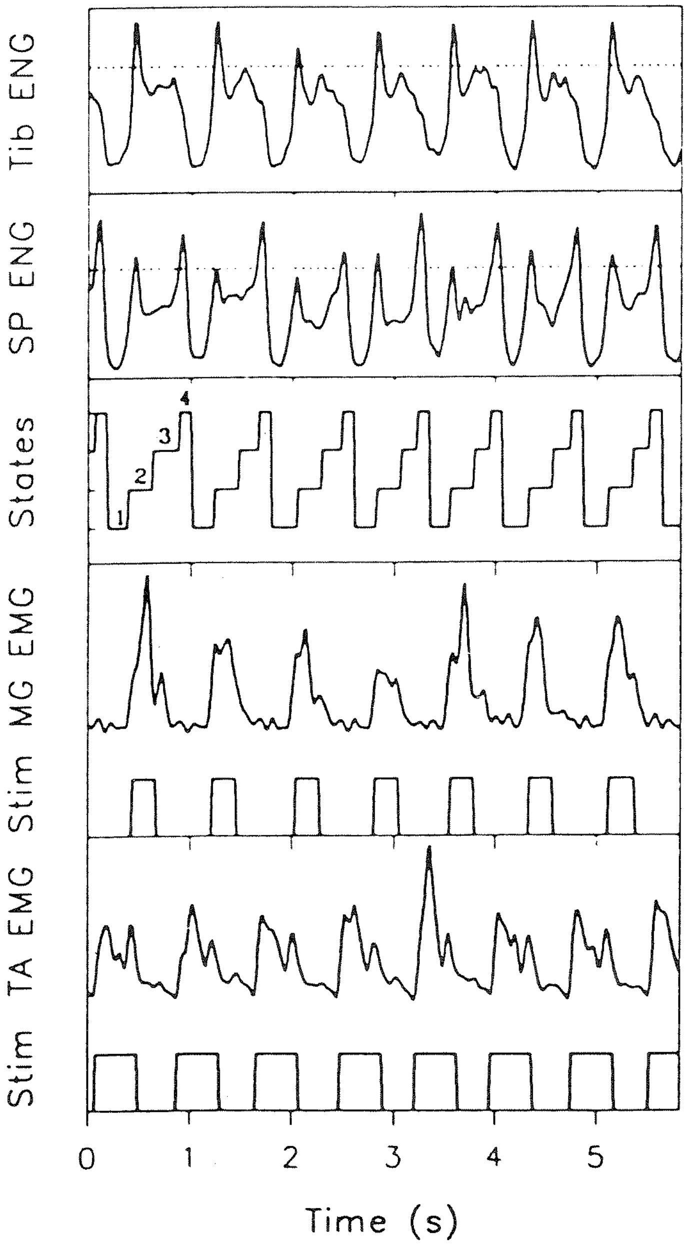

Figure 2.
Signals were recorded from the tibial (Tib) and superficial peroneal (SP) nerves and the medial gastrocnemius (MG) and tibialis anterior (TA) muscles while a cat was walking on a treadmill. Note the greater reproducibility of the neural (ENG) signals than of the muscular (EMG) signals from step to step. Thresholds were set (dotted lines) to determine state changes that could be used to trigger stimulation for FNS, as explained in the original article by Popović and colleagues (47) © 1995 IEEE.

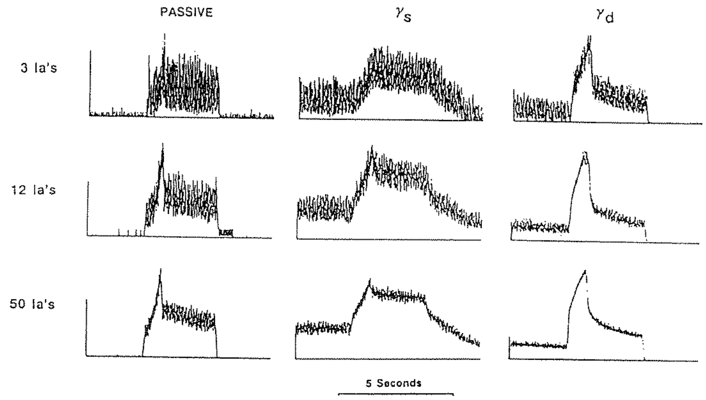

Figure 3.
Responses of primary muscle spindle sensory fibers (Ia) to a 4 mm "ramp and hold" stretch of the muscle in which they lie. The nerve impulses from each fiber were filtered, as would occur in synaptic transmission, to model the response of a motoneuron to these inputs. Responses are shown for populations of 3, 12, and 50 sensory fibers when the spindles had no fusimotor sensitization (passive) or when static (γ_s) and dynamic ($\gamma\equiv$) fusimotor motoneurons were stimulated. Note the reduction of "noise" when more sensory fibers are included. *Reprinted with permission from Vincent (9).*

necessarily linear, and control systems for higher level control do not require linearity.

External sensors have been used in the development of control systems for walking at the University of Alberta. One subject was a 7-year-old girl with paraplegia who was fitted with an FNS system that used force-sensing resistors (FSR) in shoe insoles, and inclinometers at the hip to sense flexion, extension, abduction, and adduction. In the search for signals with reasonable predictive value, an analysis was made of when the FSR signals crossed various thresholds, relative to the subject's hand-switch triggering and to the gait phases. This is illustrated in **Figure 4** for a single FSR under the patient's left heel. A threshold was set (dotted line) and pulses were generated at the times of crossing the threshold. These times were compared to the times when the patient had voluntarily pressed a switch to take a step with the right foot. Both missing pulses (false negatives) and extra pulses (false positives) were observed, as well as variable delays. Use of additional sensors could improve the situation, but the combination of sensors and thresholds would have to be "hand-crafted" for each patient.

Machine learning techniques offer the opportunity to automate and improve this process. In the technique of inductive learning (13), one "training" set of data is used to derive the decision rules and then a second "test" set is used to determine the ability to generalize to new data. Inductive learning was very well able to reduce the number of false negatives and positives to low levels, and to reproduce the timing of the patient

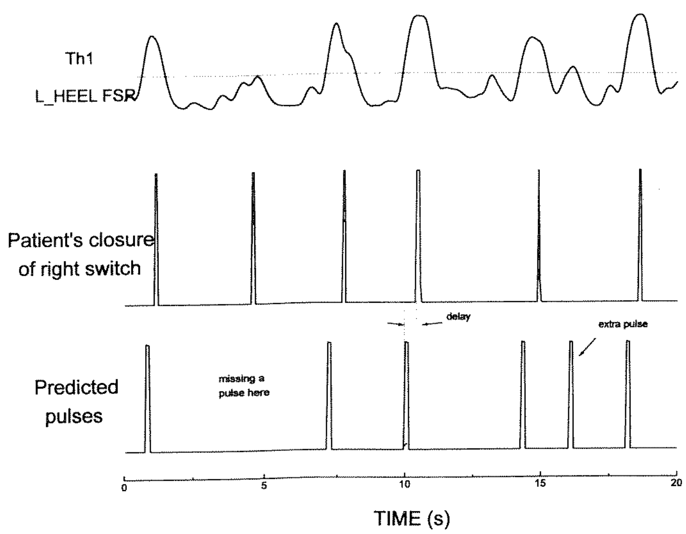

Figure 4.
Use of force-sensitive resistors (FSR) under the left heel and the times of crossing a threshold (Th) to predict the times when a paraplegic patient closed a hand switch to trigger stimulation of a step. Note the lack of reproducibility of the signal from step to step which leads to missing pulses, extra pulses, and time delays[1].

pushing the hand switch (14). Another advantage of inductive learning is that it derives a decision tree that can indicate the relative importance of various sensors, which may not correspond with the relative importance of sensors assumed in the simple threshold system when using hand-crafted rules.

Another method for making higher level decisions is the adaptive logic network (ALN). This is a type of neural network, but it does not involve the probability calculations of the back-propagation algorithm used in traditional neural nets. Rather, it uses binary gates leading to very quick operation once the rule tree is derived. The ALN method was compared to inductive learning and hand-crafted rules in a 43-year-old woman with a unilateral percutaneous FNS system. Timing of signals from FSRs was compared to desired times of stimulation, as indicated by voluntary triggering with a hand switch. The ALN performed well and had some advantages and disadvantages compared to the other methods (14).

In general, function can be improved further when restriction rules are imposed. These provide general limits; for example, that the stimulation period must be between a minimum and a maximum value and the period between successive stimulations must be more than a minimum amount. The ALN with restriction rules has now been successfully used to control the stimulation for the above-mentioned patient on-line in the laboratory (15).

Although these decision-making methods have been tested with externally placed artificial sensors, clearly they will be transferable to natural sensors when they are implemented. Of the three examples given—handcrafted rules, inductive learning, and ALN—the method that performed best combined machine learning, such as the ALN, with restriction rules. Machine learning offers the flexibility and quick generation of rules without the trial-and-error searches required to set up handcrafted rules, while the restriction rules allow the inclusion of features that will ensure safe operation.

Sensory Feedback from Cutaneous Receptors to Improve Control of FNS

Application to Foot-drop

When natural sensory feedback in FNS is considered in the hand, fingers, or sole of the foot, the cutaneous mechanoreceptors in the glabrous skin are the sensors of primary importance. These are of four types, classified as either slow- or fast-adapting (adaptation is the property of decreasing response when a mechanical stimulus of constant amplitude is repeated) and small or large field (16). If a recording method is to be clinically applicable, the ENG sources must be stable and invariant over time and the recording electrodes must be robust.

A whole-nerve cuff electrode that monitors the summed activity of all the axons that innervate an area of skin avoids the problems of specificity of location of the stimulus, and drift and fragility of electrodes that record from only a small number of axons. Whole nerve cuff electrodes are robust and have an excellent record of safety and stability when correctly built and installed (17–21). The ENG, recorded with the nerve cuff, reflects both the recruitment of mechanoreceptors and changes in the firing patterns of active receptors responding to changing mechanical stimuli (22).

In animal studies, it was found that 1) the nerve signal carried detailed information related to the force applied perpendicularly on the cat footpad; 2) it was possible to record it during electrical stimulation of nearby muscles; and 3) the signal could be used for feedback (11,22,23). The next question asked was whether such methods could be used in humans.

In a case study conducted at Aalborg University in Denmark, in a 35-year-old man with spastic hemiplegia

[1] R Dai, A Kostov, and RB Stein, unpublished data.

Journal of Rehabilitation Research and Development Vol. 33 No. 2 1996

and foot-drop, the sural nerve on the affected leg was instrumented with a tripolar nerve recording electrode (24). The sural nerve is a long, purely cutaneous nerve, about 2 mm in diameter (ID), that innervates the side of the foot. It is superficial and easy to instrument. It was judged a good choice for a first test in humans because if damage to the nerve occurred, this would not leave the individual at a great disadvantage. In fact, the sural nerve is often harvested for nerve grafting in surgical repairs of the upper limb. The cuff was implanted during a simple neurosurgical procedure performed under local anesthesia. Care was taken to place the cuff so that it was neither pulled nor twisted by the leadout wires.

The recording cuff (3 cm long, 2.5 mm ID) was made of silicone tubing and contained three circumferential metal electrodes (**Figure 5**). For a detailed review of the nerve cuff electrode technique see Hoffer (21). The insulating cuff resolves the small action currents generated by nerve fibers by constraining the current to flow within a long narrow resistive path. The 3-cm length (large compared to stimulation cuffs) is near optimal for recording from large afferent axons (21,25), allowing differential recording from a center electrode with respect to two end electrodes. Lead wires were percutaneous, with no special connector to the external electronic device (**Figure 6**).

The nerve signals were on the order of microvolts (10). When muscles were stimulated, artifacts dominated the recorded signal as both the electrical stimulus (order of volts) and the compound evoked muscle response (order of millivolts) were picked up. High-pass filtering and bin integration synchronized with the stimulation (10,23) eliminated most of the artifacts and produced ENGs that contained information about foot-floor contact that was adequate for stimulation control.

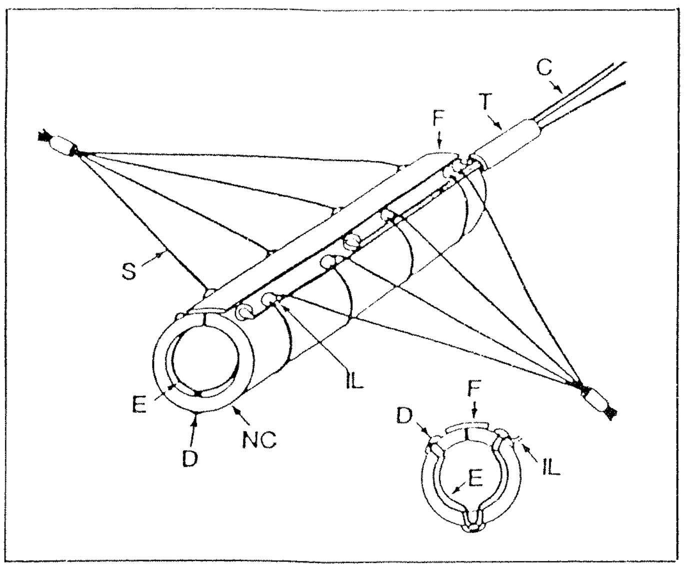

Figure 5.
Schematic diagram of a tripolar nerve cuff recording electrode. The insulating nerve cuff (NC) is made of flexible silicone tubing. Three multistrand stainless-steel wire electrodes (E) are deinsulated and sewn circumferentially to the inside wall. At the points where the wires go through the wall, they are covered with a drop (D) of silicone. Their insulated leads (IL) are gathered in a short length of silicone tubing (T) and run subcutaneously as a cable (C) toward the exit-point in the skin. A silicone sheet flap (F) adhered over the cuff opening ensures a tight seal. Several sutures (S) attached to the outside of the cuff are pulled apart to open the cuff and are individually tied to close the cuff tightly after the nerve is placed inside. *Figure from Hoffer (21) is used with permission of Human Press, Totowa, NJ.*

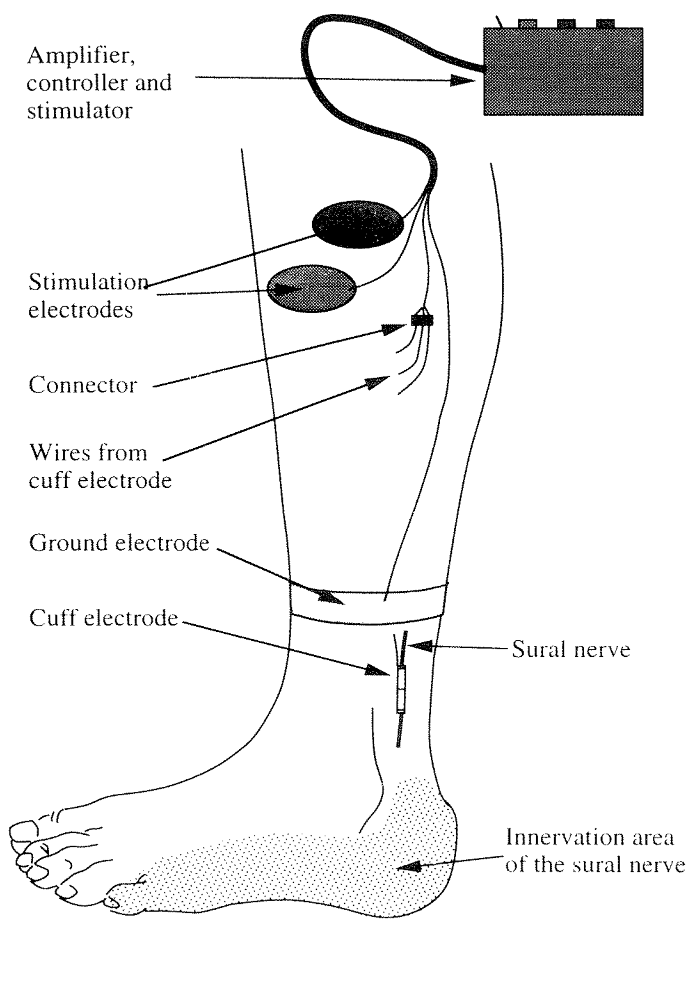

Figure 6.
Apparatus and electrode placement for portable footdrop orthosis. *Reprinted from Haugland and Sinkjær (10) with permission from IEEE Trans Rehabil Eng © 1995 IEEE.*

Perpendicular and lateral forces (up to about 8 N), recorded together with nerve activity, showed features that were essentially similar to analogous recordings in cats (22,24). When a force step was applied, there was a brief spike of activity that settled back to slightly more than baseline level. That is, there was a slight DC component during maintained force, as in the cat data. When the force was removed, there was an ''off'' response, caused by the mechanoreceptors responding to the change in force. Frequency content of the raw neural signal was principally between 500 Hz and 2 kHz, making it possible to reject EMG contamination and 60-cycle hum by high-pass filtering. ENG amplitude varied linearly with the slope of the force application in N/s (10). There was also slip information as in the cat data, suggesting that slip detection may be feasible using neural signals (11).

It was hypothesized that by detecting heel contact and lift, the ENG could be used to distinguish between stance and swing phase and thereby provide a feedback signal for an FNS foot-drop system. In the test setup, ENG was translated into a stimulator control signal (**Figure 7**). The stimulator supplied impulses to surface electrodes and also produced synchronization signals to assist in filtering the EMG artifacts. The amplifier produced a noisy signal (contaminated with thermal noise, muscle signals, and artifacts) which was then high-pass filtered, rectified, and bin integrated (26) in synchrony with the stimulation signal. The bin integrated signal, which was a mostly noise-free nerve signal, was passed into a peak detector, which indicated heel strike. In simultaneous recordings of rectified ENG and heel switch signals during gait, it was more difficult to see where stance ended and swing started than where stance began. Since only heel contact could be determined reliably, a timer was used to estimate heel-off. The FNS foot-drop system, using ENG feedback signals, functioned equally well with the user barefoot and wearing shoes (**Figure 8**); it worked best during a fast and secure gait. The rate of failures was higher when the gait was slow. Failure was defined as either a false positive heel detection (ankle dorsiflexors stimulated when not necessary) or a false negative heel detection (dorsiflexors not stimulated when required). To make such a system functional for everyday use, it will be necessary to eliminate the percutaneous wires, reduce the size and power consumption of both the stimulator and the control unit, and refine the signal processing to remove residual EMG. While the sural nerve provides useful signals, it innervates only the

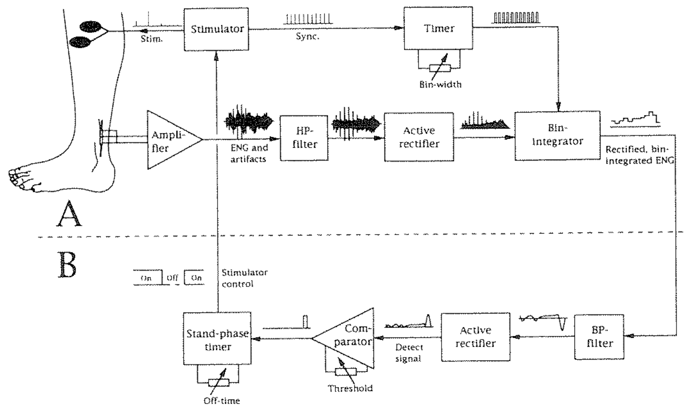

Figure 7.
Block diagram of portable foot-drop orthosis. *Reprinted from Haugland and Sinkjær, 1995 (10) with permission from IEEE Trans Rehabil Eng © 1995 IEEE.*

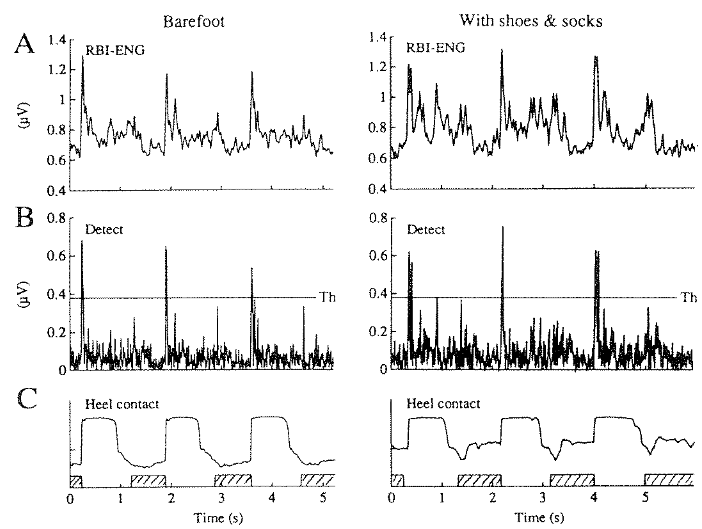

Figure 8.
Data from walking barefoot and with shoes and socks *Reprinted from Sinkjær et al. (48).*

dorsolateral portion of the foot. The group in Aalborg recently placed a nerve cuff on the tibial nerve, innervating the sole of the foot, of an MS patient with foot-drop. This also provided a consistent ''heel-contact'' signal during walking (27).

Application to Grasp

Demonstration that natural sensory feedback can be used to control paralyzed leg muscles has led to exploration of its feasibility for controlling grasping at Aalborg University. The innervation density of cutaneous mechanoreceptors in the fingertips ranges from 10 to 140 units/cm^2 and their receptive field areas range from 11 to 101 mm^2. If ENG could be extracted reliably, hand grasp could be controlled in an FNS hand system. To test this, a 2-cm long cuff electrode of the type described above was installed circumferentially around the radial branch of the palmar digital nerve in a subject with quadriplegia. The cuff lead wires were routed subcutaneously over the wrist to the dorsum of the forearm where they exited and terminated in a 3-pin connector. The palmar digital nerve innervates the lateral part of the second finger.

Reductions in the nerve signal amplitude were observed within a few days after cuff implantation. A surface electrode was used to stimulate supramaximally and a compound action potential was recorded with the cuff electrode. The amplitude of ENG dropped from 110 µV at day 1 to about 60 µV at day 11, and then slowly rose to above 110 µV after 4 months. The observed changes in amplitude were not reflected in the conduction velocity of the nerve signals, which makes the interpretation difficult. However, the nerve may have experienced changes shortly after implantation, similar to those observed in animals (28).

Studies in Sweden by Johanssen and Westling (29,30) showed that the intact nervous system uses slip information to trigger increases in grip force. In animal studies in Canada, Haugland and Hoffer determined that the ENG contains slip information that can be identified (11). Working together with R. Riso of Case Western Reserve University, the Aalborg group studied the nerve signal in the implanted human subject when an object was moved along the skin (31). They recorded rectified and filtered nerve signals during slipping with suede and sandpaper surfaces. Excursion remained constant and velocity varied from 5 to 25 m/s. When the object was slid along the skin there was a velocity-dependent response. Suede and sandpaper caused similar patterns of ENG burst responses as a function of velocity, but with different amplitudes. Both the ON and OFF responses of the receptors were detected when the finger was touched. Microslips, where the held object did not move, were also detectable. Applying percutaneous electrodes developed at Case Western Reserve for stimulating the hand muscles, it was recently demonstrated in this subject that the neural signal can indeed be useful in slip control of a neuroprosthetic system (32).

EMG Signals for Control of FNS

The EMG signal provides an indicator of muscle state that is easy to measure, suggesting that it would be a potentially useful signal for command and control of FNS systems. The simplicity of surface measurement of EMG offers a great contrast to the invasiveness of the nerve cuff sensors described above. However, the unreliability of the EMG signals that result leads to a pessimistic prediction for the method's use in practical FNS systems.

The EMG is the summation of activity of many individual motor unit action potentials. The EMG that is produced through voluntary muscle contraction is the summation of signals from asynchronous motor units and is stochastic. The information content of this signal is reflected in its frequency properties and amplitude distribution. By contrast, the EMG recorded from electrically activated muscles is the summation of signals from synchronous motor units and resembles a giant motor unit action potential. Thus, measurement and processing are different in voluntary and stimulation-evoked EMG. Characteristics of the processed EMG signal are determined primarily by the design and placement of the recording electrodes and secondarily by the distribution of active muscle fibers, amplifier and filter circuits (33). Therefore, the properties of any particular investigator's EMG signal will be unique.

Stimulus artifact is a very important problem when using EMG for feedback signals in an FNS system. There is electrical activity of large amplitude coming from stimulus electrodes, especially surface electrodes. The EMG is much smaller in amplitude and is buried into the artifact. The artifact is grossly affected by the design of the stimulator output stage and the EMG input stage. One method of reducing the artifact is to short the stimulation output after the useful part of the stimulator pulse (34). For this method to work properly, a stimulator with capacitor-coupled output must be used. With transformer coupled output, the output inductance combines with the skin capacitance and causes a ringing which can lengthen the stimulus artifact (35).

In processing EMG, one must pay attention to the particular processing blocks and their order. If the amplifier contains a preliminary high gain stage, as is normally used for voluntary EMG processing, the artifact will saturate the amplifier. Instead, good success has been achieved using an initial DC-coupled buffering stage with gain=1. An appropriate combination of gains, slew rate limiting, and blanking does an adequate job of eliminating the artifact, as is shown in **Figure 9**. Because so much adjustment of amplifier parameters is needed from individual to individual to obtain a clean

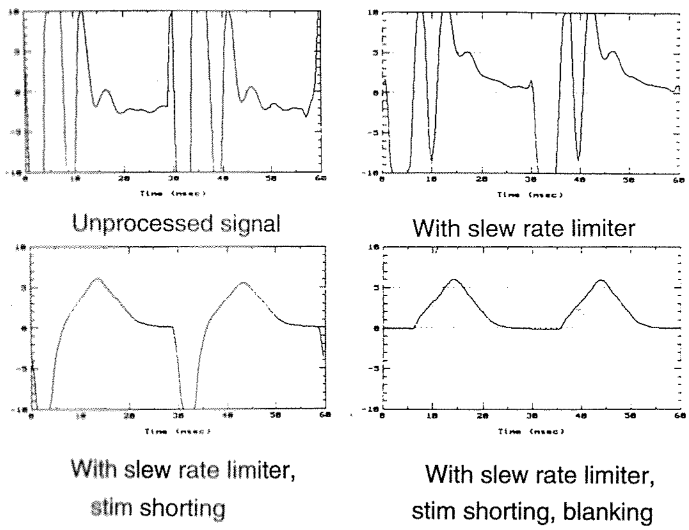

Figure 9.
Stimulus artifact reduction showing signal at various stages in the EMG processor. Artifact reduction is necessary to detect surface EMG when using surface electrodes to stimulate nearby muscles. *Reprinted with permission from N.C. Chesler. EMG as an indicator of fatigue during FES-aided standing of paraplegics (Master's Thesis). Cambridge, MA: Department of Mechanical Engineering, MIT, 1991.*

signal, the presenting author is pessimistic about how practical EMG can be in FNS systems.

Most of these problems do not exist with implanted nerve cuff stimulating electrodes and implanted EMG recording electrodes because the artifacts are much smaller. For example, Solomonow (36) reduced artifact simply by blanking the EMG amplifiers during the stimulus pulse.

To use EMG as a control signal source, one must find a property of EMG that changes with muscle state. One possibility is to use averaged EMG amplitude as a measure of muscle force. There is a trade-off between how many of the EMG spikes are averaged and the frequency response desired in the final signal; the more EMGs are averaged, the farther apart are the time steps at which there will be knowledge of the signal. But, a cleaner signal will result.

Another approach is to look at the frequency content of the signal, which changes with muscle fatigue. A spectral power estimate can be obtained by averaging the EMG spikes and taking the fast Fourier transform (FFT) of the result. The mean or median frequency can then be used for a control signal. Most of the EMG power occurs at frequencies below 300 Hz, with peaks around 100 Hz. Contrast that with ENG signals where the power occurs in the kHz range.

The EMG has been used in FNS research in a variety of ways. Hoshimiya et al. (4) used multichannel EMG in nondisabled subjects to synthesize patterns for open-loop FNS stimulation. Solomonow et al. (36) stimulated muscles isometrically with nerve cuff electrodes and used EMG as a force feedback signal. They were able to vary recruitment through a high frequency nerve conduction blocking technique and demonstrated that the EMG-force relation depended on recruitment strategy in stimulated muscle in the same way it does in voluntary muscle. Graupe et al. (37) have used EMG to control step, stand, and sit in FNS-aided gait by sensing shoulder muscle EMG voluntarily generated by the subject. In addition, they used EMG monitoring to provide signals that compensated for fatigue in quadriceps during FNS-aided standing (37).

EMG as a fatigue indicator was tested in a study at MIT using surface electrodes to stimulate the quadriceps muscle isometrically (35). In 20 nondisabled subjects and 3 subjects with complete spinal cord injury at C6 to T10 levels, isometric knee torque and quadriceps EMG were monitored in response to stimulation. Examination of amplitude and frequency measures of the EMG signal was inconclusive. In some trials, the selected measures tracked force well, but in others there was little or no correlation. Possible reasons for such variable results include the contaminating influence of voluntary contraction and insufficient sampling of motor units with the EMG electrode due to size and position of the electrode.

EMG signals are relatively easy to measure, particularly in surface electrode systems, but the relationship between EMG and force is complex. After a fatiguing contraction, the EMG recovers before force and its utility as a practical indicator in FNS systems may be limited. This study only looked at isometric systems. Once the muscle starts to move, EMG-force relations change, making EMG use very protocol-specific. The situation is more hopeful with implanted EMG electrodes. Perhaps, use of EMG as a binary (on-off) rather than proportional signal would be more practical. It is concluded that use of the EMG signal as a reliable force/fatigue indicator is an open research question.

Relation Between Motor Control and Signals from Brain Cortex

Up to this point, only natural signals from the muscles or the peripheral nervous system have been considered. In this section, the relation between cerebral cortical activity and arm movement is explored. Because signals from single motor cortex units are processed in the central nervous system before they contribute to controlling muscle, the type of information available from these units may be very different from what is observed in the peripheral nervous system.

Arm movements were examined in search of relationships between the kinematics and the goal of the movement. For example, consider the differences between moving your arm and hand to pick up a glass of water, and moving them to swat a fly. It is thought that the cerebral cortex is important in imparting the behavioral aspects of these movements.

Monkeys were studied who had been trained to make arm movements. In the initial studies, single unit cortical activity was recorded while the monkeys moved a handle in one of eight directions uniformly distributed around a 6-cm circle (center-out task). The average discharge rate of the cortical neuron during the reaction and movement time was plotted against movement velocity and direction. For any particular neuron a peak discharge rate was found in a preferred direction, a minimal rate in the opposite direction, and graded rates in between (38,39).

For a neuroprosthetic device, the ideal would be to record a cortical signal, interpret the direction of arm movement from it, and use that to control the device. However, single unit recordings present problems: near the peak discharge rate, small changes in discharge rate occur with large changes in direction. Furthermore, the discharge rate is an ambiguous function with a single cell's discharge rate coding for more than one direction of movement. The most fundamental finding is that the discharge rate is a broad tuning function, covering all directions of movement. Each cell is generally active for all movements. This has been found almost everywhere in the central nervous system. In a population of cells, all cells can be active for most movements; thus, parameter coding is almost always coarse. When one considers more than one cell, they may each have different preferred directions, and thus slightly different tuning functions. The combination of discharge rates can give a better prediction because the effects of ambiguous and coarse coding will be reduced. The more units added, the better the prediction can be.

An algorithm called the population vector algorithm is used. Cells in the motor cortex have different preferred directions (detailed by their tuning functions), which are found to be uniformly distributed across the spatial domain. In a computer-based process, a unit vector is laid down in the preferred direction of each cell. The magnitude of the vector is scaled by the unit's firing rate to obtain the contribution from that cell to the population vector. The summed contributions produce a population vector that can be shown to match the movement direction, with length corresponding to the movement speed.

This method was first tested in studies by Georgopoulos and colleagues (40). For each of the eight directions of the center-out task, a population vector was calculated. These matched closely the direction of movement. This was reproduced in three-dimensional space where population vectors and movement vectors were shown to match (41–45).

Recent studies examined how the movement trajectory is represented in the motor cortex. Monkeys were trained to trace figures—sinusoids, figure eights, and spirals—with a finger on a touch screen. Recordings were made from 349 single cells while the monkeys did the center-out task to determine the cells' preferred directions. Then the monkeys traced one of the figures, while recordings were made from each cortical unit. After signals from each cell were recorded, population vectors were calculated (45).

It was hypothesized that neuronal population vectors represent the trajectory of the hand. Spike data from each cell were aligned to movement onset and divided into 100 bins spread over the duration of the task. For each bin, a population vector summed across all cells was generated. Finger trajectory data, recorded as each neuron was studied, were similarly normalized into 100 values and averaged across experiments in which different cell activity was recorded. As the figures were drawn, the trajectory vectors and the population vectors had nearly identical lengths and directions (cross-correlations >0.8). When the curvature of the shape being traced increased, the speed of the hand decreased. There was also good correspondence between finger speed and length of the population vector. The population vector preceded the corresponding portion of the movement by 120 msec (45). This suggests that the population vector might make a good control signal.

When the population vectors recorded during the drawing of these figures were placed tip to tail, the overall shape was a close match to the finger's trajectory. Thus, the shape of the movement trajectory could be extracted from the cortical activity (45).

In new studies, researchers at Arizona State University and we (ABS) are planning to record instantaneous population vectors from many units simultaneously, and process them through an artificial neural network, which uses a Kohonen feature map to extract trajectory information. In future studies, we plan to use this procedure as a test in animals operating a robotic arm to reach food. Ultimately, these methods may be applicable to control of prosthetic systems, including FNS systems, by human subjects.

DISCUSSION

Many sensors will be required in FNS systems designed to provide complex motor functions such as walking or grasping. If such systems are to be practical, they will ultimately need to be implanted. Implanted systems involve constraints on data transmission in and out of the body, leading to a need for internal signal processing. Sensors and signal processing need to be sufficiently robust (self-tuning, largely pre-set) so that they can be greatly miniaturized. External components will have to involve minimal donning time.

The main sensory modalities used in normal motor control are touch (cutaneous receptors), position (proprioceptors, mostly muscle spindles), and activation (efference copy from the motor neurons themselves by the Renshaw cells which send a signal back into the nervous system to tell the system how much effort is going into the muscles). We often speak of the need for indicators of force and fatigue, but there is no clearly identifiable biological process corresponding to either. There are, of course, the Golgi tendon organs; but these sensors of force in muscle are not used for the types of force feedback we normally think of, and are not consciously accessible as a sense of muscle force.

This discussion will concentrate then on the modalities of touch, position, and activation. They may be used in three types of control: 1) State-based control, such as, in switching from one part of a motor program to another or making a pre-programmed response to a perturbation; 2) Continuous regulation, by mediating adjustments to unexpected but slowly changing conditions of load or muscle fatigue; and 3) Command signals, by initiating the next muscle activation sequence on the basis of volitional control (See **Table 1**).

Touch is used biologically for state switching, and may be most useful in prosthetic systems, as was described above (10,24). Touch feedback indicates when a contact has been made or broken, but is not very good for continuous regulation of motor variables. State switching may also come from position sensors or from the natural proprioceptors. If you have clever state-switching algorithms, as (RBS) described above, you can make use of complex, noisy information.

Table 1
Potential value of natural sensory modalities in three types of control.

	State-based	Regulation	Command
Touch (cutaneous)	++	–	–
Position (proprioception)	+	+	++
Activation (efference copy)	–	++	+
Force	?	?	?
Fatigue	?	?	?

Scale: – (not valuable), + (somewhat valuable), ++ (very valuable).

For continuous regulation, one usually thinks about using proprioception. But there may be a much larger need for the use of activation signals, particularly in prosthetic systems. One of the biggest problems that all FNS systems have is that the recruitment curves for activation of muscle are very steep. Another major problem is that the recruitment curves shift with changes in the position of the limb and electrodes. It would be very useful if EMG signals could provide a moment-to-moment indication of where you are on the recruitment curve, to resolve these very serious problems in control of open loop systems.

For command channels, signals thus far have generally originated from externally mounted position sensors. There has been a long-standing interest in obtaining command signals from the remaining muscles under voluntary control. The field of myoelectric control has been a frustrating area for many of the reasons mentioned above (WKD). It remains to be shown whether EMG will be useful in neuroprosthetic systems.

FNS systems employing a large number of widely distributed sources of command and feedback signals would seem to imply that extensive surgery was necessary to implant and maintain large numbers of long flexible leads among them. An alternative approach that is being developed is an implanted microstimulator that can be injected into muscles and can receive power and data from the outside by radiotelemetry (46). The next generation device under development will also sense and send data out. When multiple devices are used, there will be many opportunities for synergistic interactions. For example, devices may need to communicate with one another; one device may monitor EMG produced in response to another's muscle stimulation; relative positions between devices may provide joint angle information. We may find ourselves with abundant data, but not necessarily from modalities that have traditionally been used for servocontrol. In these cases, intelligent logic or neural network methods may be required to combine signals and so obtain the control variable that is needed at any given time.

Ultimately, these advanced approaches for control of FNS must be proven to be reliable over the long-term, safe, cosmetically acceptable, simple to operate and cost effective.

REFERENCES

1. Hoffer JA, Loeb GE. Electrical and mechanical interfaces with nerve and muscle. Ann Biomed Eng 1980:8:351–60.
2. Loeb GE, Walmsley B, Duysens J. Obtaining proprioceptive information from natural limbs: implantable transducers vs. somatosensory neuron recordings. In: Neuman MR et al., eds. Physical sensors for biomedical applications. Proceedings of Workshop on Solid State Physical Sensors for Biomedical Application. Boca Raton: CRC Press, Inc, 1980.
3. Kobetic R, Marsolais EB. Synthesis of paraplegic gait with multichannel functional neuromuscular stimulation. IEEE Trans Rehabil Eng 1994:TRE-2:66–79.
4. Hoshimiya N, Naito A, Yajima M, Handa Y. A multichannel FNS system for the restoration of motor functions in high spinal cord injury patients: a respiration-controlled system for multijoint upper extremity. IEEE Trans Biomed Eng 1989:36(7):754–60.
5. Marsolais EB, Kobetic R. Development of a practical electrical stimulation system for restoring gait in the paralyzed patient. Clin Orthop 1988:233:64–74.
6. Hoffer JA, Haugland MK. Signals from tactile sensors in glabrous skin suitable for restoring motor functions in paralyzed humans. In: Stein RB, Peckham PH, Popović D, eds. Neural prostheses: replacing motor function after disease or disability. New York: Oxford University Press, 1992:99–125.
7. Stein RB. The information capacity of nerve cells using a frequency code. Biophys J 1967:7:797–826.
8. Loeb GE, Marks WB. Optimal control principles for sensory transducers. In: Boyd IA, Gladden MH, eds. Proceedings of the International Symposium: the muscle spindle. London: MacMillan Ltd, 1985:409–15.
9. Vincent S. Costs and benefits of neuronal ensembles and high firing rates in neuronal systems (Thesis). University of Alberta, Edmonton, Canada, 1990.
10. Haugland M, Sinkjær T. Cutaneous whole nerve recordings used for correction of footdrop in hemiplegic man. IEEE Trans Rehabil Eng 1995. In press.
11. Haugland MK, Hoffer JA. Slip information provided by nerve cuff signals: Application in closed-loop control of functional electrical stimulation. IEEE Trans Rehabil Eng 1994:2(1):29–36.
12. Riso R, Haugland M, Slot P, Haase J, Sinkjær T. Characterization of sensory nerve responses to mechanical skin stimulation recorded by chronically implanted cuff electrodes in man. In: Proceedings of the 20th Annual Meeting of the Society for Neuroscience, 1994:146.8:335.
13. Kirkwood CA, Andrews BJ, Mowforth P. Automatic detection of gait events: a case study using inductive learning techniques. J Biomed Eng 1989:11:511–6.
14. Kostov A, Andrews BJ, Popović D, Stein RB, Armstrong WW. Machine learning in control of functional electrical stimulation systems for locomotion. IEEE Trans Biomed Eng. 1995. In press.
15. Stein RB, Kostov A, Popović D, Armstrong WW, Thomas M. FNS-aided locomotion controlled in real-time by artificial neural networks. Can J Physiol Pharmacol 1995. In press.
16. Vallbo AB, Johansson RS. Properties of cutaneous mechanoreceptors in the human hand related to touch sensation. Human Neurobiol 1994:3:3–14.
17. Glenn WWL, Phelps ML. Diaphragm pacing by electrical stimulation of the phrenic nerve. Neurosurgery 1985:17:974–84.
18. Hambrecht FT, Reswick JB, eds. Functional electrical stimulation: applications in neural prostheses. Biomed Eng Instrum Ser 3. New York: Marcel Dekker, 1977.
19. McNeal DR, Bowman BR. Selective activation of muscles using peripheral nerve electrodes. Med Biol Eng Comput 1985:23:249–53.
20. Krarup C, Loeb GE. Conduction studies in peripheral cat nerve using implanted electrodes—I. Methods and findings in controls. Muscle Nerve 1988:11:922–32.
21. Hoffer JA. Techniques to record spinal cord, peripheral nerve and muscle activity in freely moving animals. In: Boulton AA, Baker GB, Vanderwold CH, eds. Neurophysiological techniques: applications to neural systems, neuromethods. Vol. 15. Clifton NJ: Humana Press, 1990:65–145.
22. Haugland MK, Hoffer JA, Sinkjær T. Skin contact force information in sensory nerve signals recorded by implanted cuff electrodes. IEEE Trans Rehabil Eng 1994:2(1):18–28.
23. Haugland MK, Hoffer JA. Artifact-free sensory nerve signals obtained from cuff electrodes during functional electrical stimulation of nearby muscles. IEEE Trans Rehabil Eng 1994:2(1):37–40.
24. Sinkjær T, Haugland M, Haase J. Natural neural sensing and artificial muscle control in man. Exp Brain Res 1994:98:542–5.
25. Marks WB, Loeb GE. Action currents, internodal potentials, and extracellular records of myelinated mammalian nerve fibers derived from node potentials. Biophys J 1976:16:655–68.
26. Bak MJ, Loeb GE. A pulsed integrator for EMG analysis. Electroencephalogr Clin Neurophysiol 1979:47:738–41.
27. Upshaw BJ, Haase J, Sinkjær T. Human calcaneal nerve activity during locomotion. Fourth IBRO World Congress of Neuroscience, July 1995, Kyoto, Japan.
28. Davis LA, Gordon T, Hoffer JA, Jhamandas J, Stein RB. Compound action potentials recorded from mammalian peripheral nerves following ligation or resuturing. J Physiol 1978:285:543–59.
29. Westling G, Johanssen RS. Factors influencing the force control during precision grip. Exp Brain Res 1984:53:277–84.

30. Johansson RS, Westling G. Signals in tactile afferents from the fingers eliciting adaptive motor responses during precision grip. Exp Brain Res 1987:66:141–54.
31. Riso R, Haugland M, Slot P, Haase J, Sinkjær T. Characterization of sensory nerve responses to mechanical skin stimulation recorded by chronically implanted cuff electrodes in man. Society for Neuroscience, November 1994, Miami Beach, FL, USA.
32. Lickel A, Haugland M, Haase J, Sinkjær T. Restoration of lateral hand grasp using natural sensors. Topical Workshop of RAFT on Control of Ambulation Using FNS. Dec. 16–18, 1994. Enschede, Holland, 1994:27–33.
33. Loeb GE, Gans C. Electromyography for experimentalists. Chicago: University of Chicago Press, 1986.
34. Merletti R, Knaflitz M, DeLuca CJ. Electrically evoked myoelectric signals. Crit Rev Biomed Eng 1992:19(4):293–340.
35. Chesler NC. EMG as an indicator of fatigue during FNS-aided standing of paraplegics (Thesis). Massachusetts Institute of Technology, Cambridge, Massachusetts, 1991.
36. Solomonow M, Baratta R, Shoji H, Ambrosia RD. The myoelectric signal of electrically stimulated muscle during recruitment: an inherent feedback parameter for a closed-loop control scheme. IEEE Trans Biomed Eng 1986:33(8):735–45.
37. Graupe D, Kohn KK. A critical review of EMG-controlled electrical stimulation in paraplegics. Crit Rev Biomed Eng 1988:15(3):187–210.
38. Georgopoulos AP, Kalaska JF, Caminiti R, Massey JT. On the relations between the direction of two-dimensional arm movements and cell discharge in primate motor cortex. J Neurosci 1982:2(11):1527–37.
39. Schwartz AB, Kettner RE, Georgopoulos AP. Primate motor cortex and free arm movements to visual targets in three-dimensional space—I. Relations between single cell discharge and direction of movement. J Neurosci 1988:8:2913–27.
40. Georgopoulos AP, Caminiti R, Kalaska JF, Massey JT. Spatial coding of movement: a hypothesis concerning the coding of movement direction by motor cortical populations. Exp Brain Res Suppl 1983:7:327–36.
41. Georgopoulos AP, Kettner RE, Schwartz AB. Primate motor cortex and free arm movements to visual targets in three-dimensional space—II. Coding of the direction of movement by a neuronal population. J Neurosci 1988:8:2928–37.
42. Schwartz AB. Motor cortical activity during drawing movements. Single-unit activity during sinusoid tracing. J Neurophysiol 1992:68:528–41.
43. Schwartz AB. Motor cortical activity during drawing movements: Population response during sinusoid tracing. J Neurophysiol 1993:70:28–36.
44. Schwartz AB, Kakavand A, Adams JL. Behavioral invariants encoded in motor cortical activity (Abstract). Soc Neurosci Abs 1991:17:308.
45. Schwartz AB. Direct cortical representation of drawing. Science 1994:265:540–2.
46. Loeb GE, Zamin CJ, Schulman JH, Troyk PR. Injectable microstimulator for functional electrical stimulation. Med Biol Eng Comput 1991:29:NS13–19.
47. Popović D, Stein RB, Jovanović K, Dai R, Kostov A, Armstrong WW. Sensory nerve recording for closed-loop control to restore motor functions. IEEE Trans Biomed Eng 1993:40:1024–31.
48. Sinkjær T, Haugland M, Haase J. The use of natural sensory nerve signals as an advanced heel-switch in drop-foot patients. In: Proceedings of the 4th Vienna International Workshop on Functional Electrostimulation, Baden/Vienna, Austria, 1992:134–7. ISBN 3-900928-02-9.