Sensory Nerve Recording for Closed-Loop Control to Restore Motor Functions

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***Abstract*—A method is developed for using neural recordings to control functional electrical stimulation (FES) to nerves and muscles. Experiments were done in chronic cats with a goal of designing a rule-based controller to generate rhythmic movements of the ankle joint during treadmill locomotion. Neural signals from the tibial and superficial peroneal nerves were recorded with cuff electrodes and processed simultaneously with muscular signals from ankle flexors and extensors in the cat's hind limb. Cuff electrodes are an effective method for long-term chronic recording in peripheral nerves without causing discomfort or damage to the nerve. For real-time operation we designed a low-noise amplifier with a blanking circuit to minimize stimulation artifacts. We used threshold detection to design a simple rule-based control and compared its output to the pattern determined using adaptive neural networks. Both the threshold detection and adaptive networks are robust enough to accommodate the variability in neural recordings. The adaptive logic network used for this study is effective in mapping transfer functions and therefore applicable for determination of gait invariants to be used for closed-loop control in an FES system. Simple rule-bases will probably be chosen for initial applications to human patients. However, more complex FES applications require more complex rule-bases and better mapping of continuous neural recordings and muscular activity. Adaptive neural networks have promise for these more complex applications.**

I. Introduction

TECHNOLOGY for functional electrical stimulation (FES) to restore movement in paralyzed limbs has advanced substantially. Implantable electrodes for safe and selective percutaneous stimulation are available Prochazka *et al.* [35] and fully implantable systems will be commercially available in the near future (e.g. Strojnik *et al.* [43], Loeb *et al.* [20]). However, adequate control of FES systems remains a major problem. Redundant muscle groups cannot be controlled in the same way that the central nervous system (CNS) controls them. In addition, a CNS injury results in modified reflexes (e.g., spasticity), so inappropriate contractions may be produced. Finally, the changed patterns of activity are responsible for modifying the contractile properties of different muscle groups. Overcoming these problems requires closed-loop control with reliable, reproducible sensory feedback. Experience in using artificial sensors has proven to be complex with numerous problems associated with mounting, reproducibility, robustness, energy consumption, etc. Webster [47].

Sensing natural signals to control assistive devices was suggested a number of years ago. For example, myoelectric hands and legs have been designed and used with variable success [reviewed by Graupe and Kohn [10]. Published results in Johansson and Westling [17] and Westling and Johansson [48] suggest that cutaneous sensory receptors have some appropriate properties for control purposes. Nerve cuff electrodes have proven to be safe and reliable for recording sensory nerve activity chronically in animals (Gordon *et al.* [9]) and so are a promising source of feedback information. To maximize the amplitude of the neural signal a nerve cuff should be well matched to the size of the nerve and be approximately as long as the wavelength of the action potential. Up to this length the signals increase as the square of the electrode spacing and inversely as the square of the internal diameter of the cuff (Stein *et al.* [40]).

Hoffer and his collaborators [14], [36], [13] correlated neural activity with external perturbations such as touch, slip, and change of contact force in experiments in cats and humans. They succeeded in using a biological sensor in a closed-loop control system for control of the cat's hind limb (Hoffer and Haugland [15]). Semichronic recordings from the palmar digital nerves (Popović and Raspopović [33]) confirmed that a similar correlation can be found in human grasping. In this paper we extend previous studies by using multielectrode recordings for the design of rule-based control, which we feel is an important element in the development of a fully implantable system for severely handicapped humans (e.g., spinal cord and stroke patients) needing a multichannel FES system.

Up to the present, the commonly used technique for control has been hand-crafted rules. Most systems are designed to use a small number of sensors (e.g., force transducers, joint angles, switches, etc.). However, hardware is becoming available to allow integration of larger number of sensors, which could improve the performance of the system. This would make hand-crafting of rules very difficult, if not impossible. Hence, there is a need for identification of invariant features for setting up the control algorithms. Ar-

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tificial neural networks (ANN) have been applied to complex control tasks (Miller *et al.* [24]). The adaptive logic networks (ALN) described in this paper are composed of nodes which, unlike the usual multilayer perceptrons and the backpropagation algorithm (Hecht-Nielsen [12]), realize logic functions AND and OR (Armstrong *et al.* [4]). The input to the network are Boolean, some of them derived by comparing continuous measurements to fixed thresholds. This has the benefit of simple realization in combinational logic circuitry, very similar to the form of a rule-based system. Future control systems can be expected to benefit from the pattern recognition capabilities of ALNs when the dimensionality of the measurement vector of inputs is high. Some initial results of the evaluation of these ALNs for synthesizing signals possibly useful in prosthesis control have been already presented (Stein *et al.* [42], Kostov *et al.* [19]).

Among several applications that can be envisioned with natural sensors one is of particular interest. Control of hand functions in tetraplegic patients may be improved if a closed-loop control is implemented. Use of artificial sensors is almost impossible because of difficulties in their mounting and space limitations. Use of the neural recordings as a sensor of afferent pathways may be also very useful (Hoffer and Haugland [15]) to estimate position on the basis of muscle spindle activity of nonstimulated muscles. In this case motor branches in the forearm should be instrumented with cuff electrodes. Another immediate application is in stroke patients having an implant at the common peroneal nerve to enhance dorsi-flexion during the swing phase of the gait. The sural nerve can be instrumented with a cuff electrode, and recordings of the neural activity used as a trigger to start the stimulation (Sinkjaer *et al.* [37]). An obvious advantage of this technique is that such a system will be the only use of electrical stimulation in stroke subjects that does not require a sole switch or force transducer. The use of cuff electrodes increases the complexity of the system, and it is necessary to include an effective telemetry system, e.g. Charles [6], to avoid leads and cables.

II. Methods

Surgical Procedure: Experiments were performed on adult cats of either sex that showed suitable gait performance on a treadmill after a suitable period of training. Under fully sterile conditions, a gas-sterilized set of electrodes was implanted and a head and/or back connector was attached according to the following protocol approved by a local ethics committee. Prior to surgery the cat was injected intramuscularly with an antibiotic (Ayercillin) to minimize the risk of infection. It was then anaesthetized with 50 μg/kg of Somnitol given intraperitoneally. Regulation of anaesthesia depth was accomplished by regular monitoring of heart rate, respiration rate and the absence of reflexes to noxious stimuli (eye blink and limb withdrawal to paw pinch). Ketamine and atropine were injected intramuscularly and the cat was intubated to allow artificial ventilation, if necessary. During surgery blood pressure and body temperature were monitored, and temperature was maintained with a heating pad. An intravenous catheter was inserted into the cephalic vein for infusing fluids when necessary. The cat was shaved and scrubbed with Betadine in the region of the incision before being draped with sterile sheets. These procedures were carried out by veterinary resource personnel.

Triphasic cuff electrodes (Stein *et al.* [40]) were implanted around several of the following nerves: sciatic, superficial peroneal (SP), tibial, common peroneal, and sural. Epimysial EMG electrodes were sewn to medial gastrocnemius (MG) and tibialis anterior (TA) muscles. After the surgery the skin was closed with sutures and the tracheal tube and venous cannula were removed. When a back connector was used the leads were brought out through the skin and attached to the connector after the surgery. After completing the procedures, the animals were transported to the intensive care unit of the animal care facility. Postsurgical care included one week of antibiotic therapy (Ayercillin), opoid-based analgesia (Buprenorphine) as required, and close supervision by investigators and support stuff in the animal care facilities of the University of Alberta.

Monitoring: Compound action potentials were elicited by stimulation of nerves at 1 Hz with a pulse width of 10 μs, and an amplitude sufficient to have a maximal potential. These measurements were done for all implanted electrodes to verify that normal conduction was preserved and no nerve block occurred as a result of surgical intervention. An increase of peak-to-peak amplitude was often observed in the first few weeks after surgery in parallel with an increase in the electrode impedance (Stein *et al.* [41]), due to the replacement of saline with connective tissue of higher impedance. Impedances were measured with a Hewlett-Packard vector impedance meter (model 4800A) at 10 kHz, since the major component at that frequency is thermal resistance. As expected, phase angles at that frequency were typically small ($< 30°$). No change in conduction velocity was observed, indicating that the nerve diameter had not changed. Neural recordings were stable, and with time, less EMG contamination was recorded (Fig. 1). Results from three cats for more than 200 days showed that relative amplitude of the peak to peak CAP remained constant. In comparison the relative contamination of the CAP by EMG decreased somewhat in all chronic cats.

Signal Processing: Neural recordings in peripheral nerves elicited from cutaneous receptors or muscle spindles are typically in the range of 3 to 10 μV. The frequency spectrum has a peak around 2 kHz with most power concentrated between 1 and 5 kHz (Fig. 2). The thermal noise with nerve electrodes is in range of 1 μV, comparable in size with the neural signals. Peripheral nerves are surrounded by muscles producing electrical signals that are orders of magnitude larger, but have a peak near 200 Hz (Fig. 2(a)) and most power below 1 kHz. This signal is partially suppressed using a good fitting cuff electrode and suturing it tightly shut. In addition, the triphasic nature of the recording produces additional common mode rejection (Stein *et al.* [39]). Nonetheless, for the tibial nerve near the ankle, two distinct peaks are observed in the spectrum (Fig. 2(b)), but high pass filtering at 1000 Hz (fourth order Krohn–Hite filter, model 3700) can eliminate virtually all EMG contamination (Fig. 2(c)). With other nerves (SP,

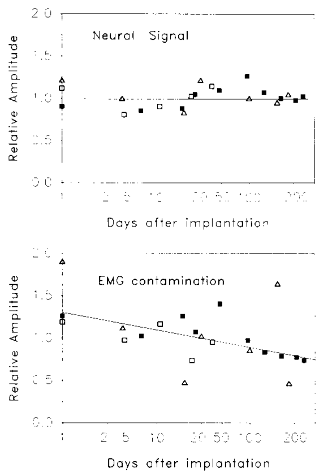

Fig. 1. The amplitudes (top) of compound action potentials were measured with triphasic cuff electrodes on three different nerves: sciatic, tibial and superficial peroneal in three legs (different symbols) of two chronic cats. The potentials were elicited by stimulating the sciatic nerve and recording from the other two nerves which are branches of it and by stimulating the branches while recording from the sciatic nerve. The four values relative to the mean for each nerve over the entire recording period were averaged and plotted. The amplitudes of EMG contamination on the other nerves (bottom) were also averaged in the same way when the sciatic nerve was stimulated maximally. The EMG contamination could easily be distinguished because of its greater latency and slower time course.

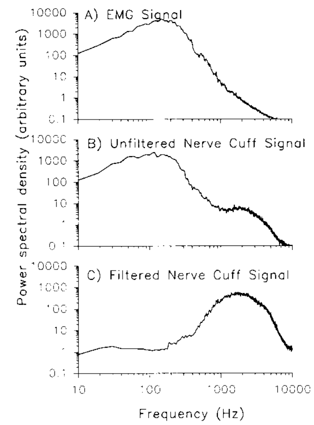

Fig. 2. (a) Power spectral density calculated from EMG recorded from medial gastrocnemius (MG) muscle, while a chronic cat walked on a treadmill. (b) Corresponding spectra are shown from a nerve cuff on the tibial nerve without filtering. (c) After high-pass filtering with a cutoff at 1000 Hz. Note that without filtering the peak spectral density of the EMG is approximately 100 times that of the neural signal (10 times the amplitude), but the ratio can be reversed using a fourth order filter.

sural) the location near major muscle groups may require a cutoff frequency of 1500 or even 2000 Hz or more.

Circuit Design: When stimulating a nerve or muscle, such as would occur in FES, a much larger artifact is produced, which is best eliminated by using a blanking circuit (Hoffer and Haugland [15]). To measure the average output from the nerve, the signal is rectified and low-pass filtered, using a combination of *RC* and Paynter filters. Although these techniques have been used for a number of years with rack-mounted equipment, the challenge was to develop a portable system with small-enough size, weight and power consumption that it could be used conveniently in an FES system.

The electronic circuit is composed of the following cascaded blocks: low noise and low input impedance preamplifier, bandpass filter, amplifier, blanking circuit, rectifier, paynter filter, controller, and stimulator.

We made the preamplifier for nerve recordings using INA110 Burr Brown instrumentation amplifier. To increase the signal to noise ratio, and to match the low impedance typically observed with nerve cuff electrodes to the high impedance that is suitable for FET preamplifiers, a miniature step-up transformer (turns ratio of 20) was employed (PICO 24400). INA110 instrumentation amplifier was set to have the gain of 200, thus the total gain of the preamplifier was A = 4000.

The output of INA110 is fed to the fourth-order bandpass filter and than to a two stage precision operational amplifiers OP77 allowing the selection of gain to 75, 400, 1500, and 2250×10^3. Total amplification was on the order of 10^4 for EMG signals and 10^{6} for nerve signals. The blanking circuit (realized with two 14066 CMOS analog switches) switches the amplifier to ground just before and after the stimulation. To prevent the output from decaying, the low-pass filtered signal is sampled and held for the period of the blanking. This blanking circuit allows recordings whenever there is no stimulation applied.

Rectification is accomplished with a full wave rectifier. Smoothing of the rectified signal effectively creates a signal similar to the envelope of the neural or muscular electrical activity. This signal can be amplified and offset.

The controller can be either a microprocessor based circuit or a preprogrammed analog circuit. The simple solution for the stimulation used in this project uses thresholds, thus the simple controller can be reduced to a comparator. This controller triggers a biphasic, charge balanced stimulator.

CMOS technology was used widely to reduce power consumption. The total power consumption is currently on the order of 9.5 mW and the conventional printed circuit boards occupy a volume of 8 cc per channel. Special attention was given to the layout of components and realization of ground loops to maximize signal to noise ration. Further details can be found in Nikolić and Popović [26] and Nikolić *et al.* [27] or obtained from the authors.

Rule Based Control of FES System: The control architecture system adopted in the present system is based on the finite state approach to the control of prosthetic and orthotic devices. The implementation of the finite state approach is based on the use of a rule-based control. Rule-based methods belong to nonanalytical control systems and have an *if-then* structure (Popović *et al.* [31]). The cyclic motor activity is presented as a sequence of discrete events. Each of these discrete events is associated with a unique sensory pattern. A sensory pattern occurring during particular motor activity is recognized with the use of artificial and/or natural sensors. The specific discrete event is called the state of the system by analogy to the state of finite-state automata. A recognized sensory pattern during a specified state of the system initiates corresponding functional movement.

Rules in a rule-based control system were initially handcrafted. This involved human expertise and it was very appropriate for simple systems having a limited number of states. The expertise for detecting rules can be classified as a pattern recognition problem. An alternative to hand-crafting event detection rules is to use rule induction methods, developed for machine learning in artificial intelligence (Andrews *et al.* [3] and Veltink *et al.* [46]). The following steps are included in this method: 1) collection of examples, termed the training set; 2) each example in the training set is described in terms of a fixed number of attributes; 3) each instance in the training set has an associated class value; and 4) the algorithm seeks to characterize each class value in terms of its attribute values. The development of artificial neural networks gives a new tool for computer generation of rules for control of an assistive device.

Adaptive Logic Networks: If the transfer function between input and output of a system is not known and cannot be described in an analytical form, artificial neural networks can approximate the required mapping. One of the interesting mappings for control of a prosthesis is that between afferent neural signals and activation of muscles of the same or the opposite limb. We have attempted to reproduce this mapping using adaptive logic networks. In our experiment the inputs to an ALN were derived from neural recordings presented in the first two traces of Fig. 4 (tibial nerve and superficial peroneal nerve). Each input signal was quantized to a certain number of levels and each level was converted into a Boolean vector in such a way that close levels resulted in vectors which were close in Hamming distance (Smith and Stanford [38] and Armstrong [4]). The required output of the system was Boolean (trace 2 in Fig. 6), namely a thresholded EMG signal from trace 1 in Fig. 6 (medical gastrocnemius EMG). The output signal has been considered as an indication of the muscle activity.

The adaptation procedure involved selecting the node functions in the ALN based on sequence of presentations of inputs and outputs (Armstrong and Gecsei [5]). The adaptation algorithms are available in the form of the ATREE 2.7 ALN Simulation Package (ftp from menaik.cs.ualberta.ca [129.128.4.241] in file pub/atre27.exe—binary mode). The software runs under Microsoft Windows 3.x and contains extensive documentation on ALNs.

A system for multiple-input real-time prosthesis control must be implemented with the greatest economy of hardware means. If, for example, a standard microprocessor is used, the cost of the system is greatly reduced. The usual type of artificial neural networks requires numerical calculations, preferably done in hardware rather than simulated arithmetic. The ALN offers the advantage of having only logical calculations in its nodes. An additional significant advantage occurs in that simulation of logic functions can be executed in a "lazy" fashion. For example, if the first input to an AND gate is 0, then the other input is irrelevant, so the subtree producing it need not be simulated. This produces significant increases in speed that are not available to backpropagation type networks. Since the results of ALN training are combinational networks, inexpensive parallel hardware realization is possible using off-the-shelf programmable logic devices.

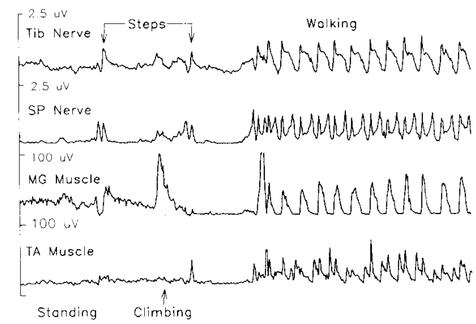

Fig. 3. Recordings from superficial peroneal (SP) and tibial (TI) nerves as well as medial gastrocnemius (MG) and tibial anterior (TA) muscles of a chronic cat. This 20 s recording was selected to indicate the distinct rhythmic activity in peripheral nerves and ankle muscles in various behaviors. Further details in the text.

Input vectors form a domain, while the output vectors form the codomain. An ALN can learn to recognize patterns, but sufficient training is required. The result of the training, when domain and codomain are presented to the ALN, is one or more trees composed of logic gates (NAND) which approximate the transfer function between the input and the output. Continuous input and output are treated by encoding them into Boolean vectors. In later use, once the training is completed, only the domain is presented to the ALN, and it generates the codomain.

III. Results

We concentrated in this study on controlling a single joint by stimulating ankle extensors and flexors and processed signals using custom build electronic circuits (see Methods). Fig. 3 shows typical records from the SP and tibial nerves as well as ankle extensor (MG) and flexor (TA) muscles of a chronic cat implanted previously. The signals cover a period of 20 s, where the cat stood for 3 s, made one step, stood again for about 2 s, climbed up on the wall of the treadmill with its forepaws while balancing on its hindlegs (note the strong burst of activity recorded in the MG muscle), took another step and then stood for 2 s before starting to walk rhythmically on the treadmill with a cycle time of $T \approx 750$ ms. There is a characteristic double burst with each step in the SP nerve and the single burst in the tibial nerve. Furthermore, although the signal amplitudes are much smaller, the processed electroneurograms (ENG) are much more reproducible in amplitude from step to step than the corresponding EMG signals.

The ENG signals were also used to design a simple state diagram, based on setting thresholds on the filtered nerve records (dashed lines on Fig. 4). The single peak in tibial nerve activity occurs each time the cat's paw hits the ground and corresponds closely to the time when the MG activity begins. Thus, the first rule in Fig. 4 is *if* the tibial nerve

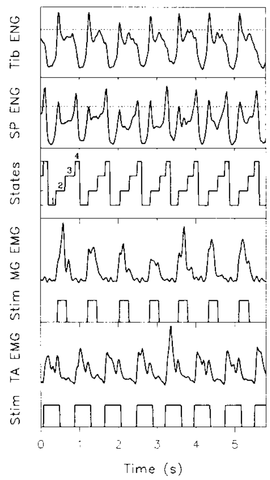

Fig. 4. Threshold levels were set (dashed lines) to determine instants when processed electroneurograms (ENG) recordings from tibial and superficial peroneal nerves (upper two traces) crossed above these preset levels. Threshold crossings activated transitions in a simple rule-base with four states (third trace), as explained further in the text. Entering state 2 turned on circuits for a fixed period of time (trace 5) that correspond on average to the duration of supra threshold activity of MG EMG (trace 4). Similarly, entering state 4 turned on a circuit for a period (bottom trace) corresponding to the suprathreshold period of the TA EMG (trace 6). Note that the circuit operated reliably with no false positive or negatives.

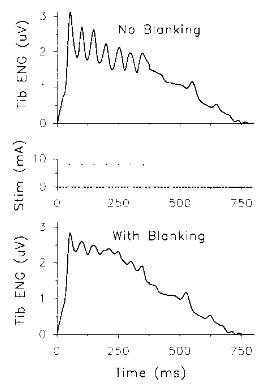

Fig. 5. When the rule-base was connected to a stimulator and stimuli were applied (middle trace), artifacts were generated on the processed tibial ENG (top trace), even with filtering, but this could be greatly reduced (bottom trace) by switching in a blanking circuit.

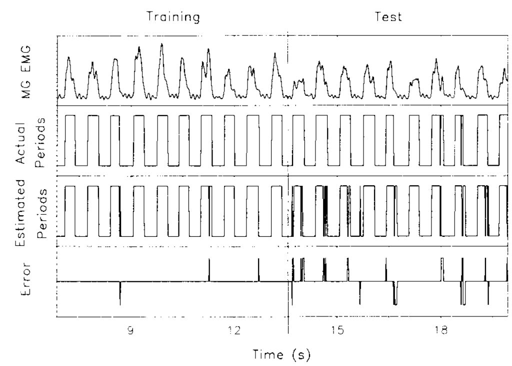

Fig. 6. To match the activity pattern of an ankle muscle, based on the signals from the tibial and SP nerves, we used an adaptive logic networks (ALN). Neural signals (upper two traces in Fig. 4) were presented to the ALN as the input, together with an output which consisted of pulses (second trace in this figure) representing periods when MG EMG exceeded a threshold value. The data on the left half of the figure were used in training of the ALN to reconstruct the binary representation of the EMG signal. The reconstructed result is in the third trace, and the error is in the fourth trace. The right half of the figure presents test data, where reconstruction was not as successful as in training. Further details in the text.

signal exceeds threshold, *then* activate the MG stimulus for a period corresponding to the average duration (307 ms) that the MG activity remains above a threshold level (not shown). As mentioned above, the SP nerve shows two peaks, one corresponding to the time when the paw hits the ground and another, larger peak when the paw is lifted off the ground. The latter peak corresponds to the onset of the TA activity. Thus the second rule is *if* the SP activity exceeds its threshold *and* the tibial nerve is below its threshold, *then* activate the TA stimulus for a period equal to the average duration (420 ms) that the TA activity remains above a threshold level. These two simple rules are sufficient to reproduce the basic structure of the alternation between the flexors and extensors controlling the cat's ankle joint.

In the example shown in Fig. 3, no stimuli were actually applied. However, with the blanking circuit it was possible to stimulate the muscle to generate substantial force without affecting the sensory nerve recordings. Recordings from the tibial nerve for one gait cycle are shown in Fig. 5, and for tibial nerve and superficial nerves when the cat is walking in Fig. 4. Hoffer *et al.* [14] obtained similar results but used a computer to process the signals. A number of years ago (Hoshimiya *et al.* [16]) developed an analog circuit for this purpose. The results shown here utilized a small battery operated circuit (see Methods) that a patient could use practically on a daily basis.

The rule base of Fig. 4 only matches two features of the normal EMG signals, their onset and average duration, but it does so reliably (no false positives or false negatives) if input signals (neural recordings) are of a certain amplitude and shape as described earlier. It may be desirable for some purposes to match the exact timing for each step profile. We examined if adaptive logic networks (ALN) could perform this function in two series of trials. The first served to find the number of samples from the past required for training and an optimal region in the prediction error plane. The second produced the actual results of training in that region. Based on previous experience with similar signals (Kostov *et al.* [19]), a decision was made to quantize neural signals to 128 levels and encode them with 14 bit. In contrast, the EMG signals were quantized to 2 levels and encoded with only 1 bit. The best training results were obtained using the original neural signals plus two copies of the same signals delayed 80 and 160 ms and two copies of differentiated neural signals with the same delays. The artificial neural trees contained 1536 nodes, the majority vote of seven trees being taken to compute an output bit, the number of presentation (epochs) of the training set to the learning algorithm during training of every tree was 20.

The results from the ALN are presented in Fig. 6. The top trace is the EMG signal. The second trace shows the actual intervals when EMG activity was above the threshold. This binary representation of the MG EMG was presented as a target signal (codomain) to the ALN during the training. The neural recordings, shown in the first two traces in Fig. 4, were used as input to the ALN (domain). The third trace presents the reconstruction of the ALN after learning. To compare the actual timing with that predicted by ALN, we included the errors in the bottom trace. The errors during training were very few (below 1%, left half of Fig. 6), but increased considerably when we tested the performance of the ALN against data that it had not seen before (over 8%, right half). This error can be reduced if different filtering is applied to the original signals. However, the present paper is not intended to be a test of neural networks but merely to demonstrate that ALN applied to neural and muscular signals may be appropriate for determination of invariant features in the locomotion and instead of hand crafting rules for control.

IV. Discussion

Up to the present only simple control algorithms have been applied to FES systems for restoring gait, such as hand switches to initiate each step or preprogrammed sequences. These sequences are based on recorded average EMG patterns in normal individuals (Marsolais *et al.* [21] and [22] and McNeal *et al.* [23], Thoma *et al.* [44]). Direct, computer control of electrical stimulation was proposed (Chizeck [7]), emphasizing muscular properties and a discrete model for restoring functional locomotion. The use of feedback can improve the performance of FES systems. One approach relies on sensing natural signals (e.g., EMG activity) for control (Graupe [11]), while the other uses artificial sensory feedback (Crago *et al.* [8]). A few closed-loop systems relying on rule-based control have been tested with artificial sensors (Andrews *et al.* [1]–[3], Joonkers and Schoute [18], Mulder *et al.* [25], Phillips [28], Phillips [29], Popović *et al.* [30], [32], Veltink *et al.* [45]).

To prove our hypothesis that multielectrode recordings from sensory nerves can provide satisfactory information for a rule-based control, we used a chronic cat model. The ankle joint of the cat's hind limb was selected for several reasons: 1) many characteristics of peripheral nerve recordings and muscle signals are well described in the literature (e.g., Stein *et al.* [39], [40]); 2) a percutaneous system consisting of several nerve-cuff and epimysial recording electrodes can be easily installed; 3) cats can be trained to perform simple motor tasks, such as walking on a powered treadmill and 4) long term recording and stimulation has proven to be effective. In this model system recordings from two sensory nerves (TI and SP) were sufficient to trigger the appropriate periods of stimulation reliably to ankle flexor and extensor muscles. The exact timing of activity could be learned using an ALN. It responded reasonably well to testing with a distinct data set, but it remains to be seen how widely the ALN can generalize before it is implemented in a patient. Until these questions are resolved, the simpler rule bases operating from artificial or natural sensors seem to offer the most promise for patient application in the near future.

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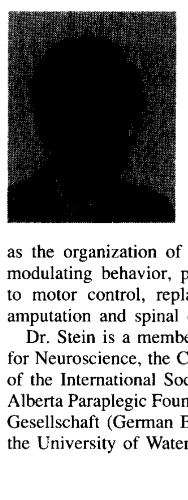

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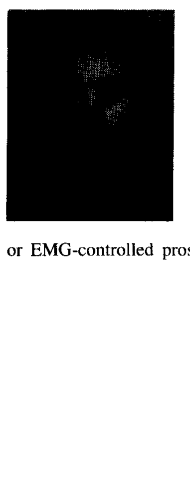

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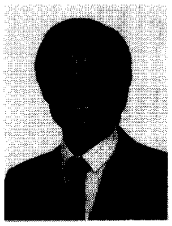

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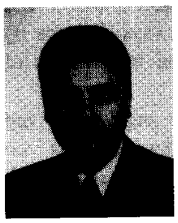

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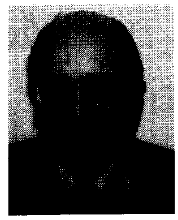

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