Communications

Artifact-Free Sensory Nerve Signals Obtained from Cuff Electrodes During Functional Electrical Stimulation of Nearby Muscles

Morten K. Haugland and J. Andy Hoffer

Abstract—**Restoration of the voluntary use of paralyzed limbs using functional neuromuscular stimulation (FNS) is limited by complex muscle properties and unpredictable load behaviors; closed-loop control of FNS would improve performance but requires reliable sensory feedback modalities. Sensory nerve signals recorded by cuff electrodes provide accurate information about forces acting on the skin in anesthetized animals [5]; however, nerve cuff signals are very small (approximately 10 μV), and during FNS they become contaminated with large stimulation artifacts and synchronous EMG potentials from nearby muscles. We show in this study that it is possible to record neural signals from the cat tibial nerve without interference from distributed stimulation of four calf muscles surrounding the recording electrode by use of high-pass filtering and synchronized bin-integration. Nerve signals sampled in this way retained all the information about footpad contact force that was normally obtained in the absence of muscle stimulation. We propose that this approach has wide applicability for rehabilitation of paralyzed people with neural prostheses.**

I. Introduction

As implantable stimulators for functional neuromuscular stimulation (FNS) have become available, the need for implantable sensors has also become apparent [4], [13]. Information from natural sensors in the skin, muscles, an joints would be valuable if it could be recorded reliably in the presence of noise and stimulation artifacts. Tripolar, balance nerve cuff electrodes have been found to be suitable for recording whole-nerve signals an provide stable signals over long term (reviewed by Hoffer [6].

In order for nerve signals to be applicable for the control of FNS in paralyzed humans, it is necessary to demonstrate that the original nerve information can still be recorded when nearby muscles are activated with FNS. The reason for this concern is that when muscles are stimulated electrically, two kinds of artifacts contaminate the signals recorded from peripheral nerves: one is the current passed through the stimulation electrodes that is also picked up by the recording electrodes, and the other is the compound electromyographic (EMG) potentials that are elicited in the stimulated muscles. Removal of artifacts in physiological signals have been the concern of many studies. The simplest method is to blank the artifact by shunting or opening the input to or the output from the amplifier during the artifact (e.g., Knaflitz and Merletti [9], Babb *et al.* [1], [12]). Other more sophisticated methods have also been used, such as adaptive filtering (Widrow *et al.* [14], Laguna *et al.* [10]) and subtraction of a template artifact (McGill *et al.* [11]).

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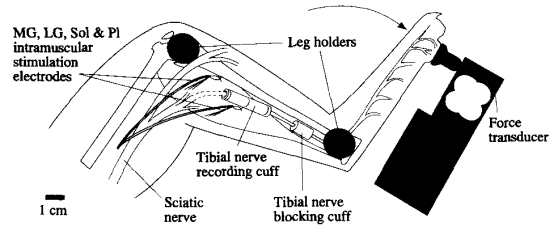

Fig. 1. Experimental setup. See text for details.

In the present study, we wanted to investigate whether artifacts produced by stimulation of muscles surrounding a nerve cuff electrode could be removed effectively enough for the remaining nerve signal to be useful as possible control/feedback signal in an FNS system. The concern was the small amplitude of the nerve signal (5–10 μV) compared with the stimulation pulses (several volts) and the evoked muscle signals (10–100 mV). An anaesthetized animal preparation was used to model a paralyzed human limb, and neural activity was recorded from the tibial nerve during electrical stimulation of the surrounding ankle plantar flexor muscles. The force produced on the footpad by the contracting muscles generated neural activity in the tibial nerve that could be recorded by the cuff electrode [5]. Preliminary results of this work have been published (Hoffer *et al.* [7], Haugland *et al.* [4], Hoffer and Haugland [8]).

II. Methods

The hindlimb of anaesthetized cats was used as experimental model of a paralyzed human arm, with the central footpad serving as a model of human fingertips. Five adult cats (4–6 kg) were chronically implanted using aseptic techniques under gas anaesthesia (Halothane in a mixture of oxygen and nitrous oxygen) and were allowed at least 3 days to recover before the first experiment. This period also allowed the devices to stabilize within the leg. The cats were maintained in accordance with the guidelines of the National Institutes of Health (USA) and the Canadian Council for Animal Care.

Bipolar intramuscular stimulation, a more selective method than stimulating via monopolar electrodes with respect to a common ground, was used in order to reduce unwanted cross stimulation of other leg muscles. Bipolar stimulation electrodes were implanted in the four ankle extensor muscles: medial (MG) and lateral gastrocnemius (LG), soleus (Sol), and plantaris (Pl). Each electrode consisted of two Teflon-coated 40-strand stainless steel wires (Cooner 634), with the ends deinsulated for 15 mm (Fig. 1). The electrodes were inserted deep in the proximal half of the muscle belly, about 2 cm apart, and secured to the muscle fascia with a 3-0 Mersilene suture.

A tripolar nerve recording cuff (2.2 mm inside diameter, 30 mm long, as described by Hoffer [6]) was implanted on the tibial nerve 4–5 cm above the ankle, distal to the muscular branches. The electroneurographic (ENG) signal recorded from the tibial nerve in this location is dominated by sensory activity from the plantar aspect of the foot and the footpads [3].

A nerve-blocking cuff (as described by Hoffer [6]) was implanted 10–15 mm distal to the recording cuff on the tibial nerve. By infusing lidocaine sodium (2% solution) through a catheter that coursed subcutaneously from a backpack connector to the blocking cuff, conduction in the nerve could be blocked transiently and reversibly without disturbing the experimental setup.

Recording sessions were performed approximately every 10 days. At the beginning of each recording session, the cat was anaesthetized with an intravenous injection of Thiopentothal (8–10 mg/kg), intubated, and maintained with Halothane in a mixture of nitrous oxide and oxygen. To remove contributions to the ENG signal arising from hair receptors, the whole foot was shaved and treated with depilatory cream, followed by a thorough wash and application of moisturizing cream.

With the cat lying on a heated table, the implanted leg was secured by two pairs of cupped clamps (Fig. 1) that allowed the ankle joint to move without damaging the skin. A hard rubber disc-shaped probe (14-mm diameter), mounted on a force transducer, was placed so that the central footpad pushed perpendicularly on it when the ankle was plantarflexed. Except for the clamps that held the ankle and the probe, nothing else touched the foot during the recordings.

Cuff electrode impedances were about 3 kΩ (measured at 1 kHz). The cuff signal was amplified in two stages. The first stage was a low-noise preamplifier (QT-5B, Leaf Electronics) set at a gain of 100 and bandwidth from 65 Hz to 10 kHz. The second stage was a general purpose differential amplifier (Bak Electronics) with a gain of 1000 and bandwidth from 50 Hz to 10 kHz.

III. FNS System

In order to record ENG signals while surrounding muscles were stimulated with FNS as would occur in a clinical implementation, we constructed a closed-loop FNS system that stimulated the four plantarflexor muscles in such a way that the force profiles applied on the footpad were comparable in size and frequency content to natural inputs.

The stimulation frequency was constant, 25 Hz for each of the four muscles. Simulation pulses were applied sequentially to each of the four muscles. If each muscle was stimulated alone, this frequency caused unfused tetani with considerable ripple (in the order of 20% of the maximum force) except for the soleus (which is a slower muscle than the others). Ripple was reduced to an acceptable level (below 5%) when the four muscles were stimulated in sequence, which made the muscles generate force with an effective frequency of 100 Hz rather than 25 Hz. Force amplitude was controlled by modulating the duration of each stimulation pulse (pulsewidth modulation). Rectangular, monopolar, regulated current pulses ranged in duration between 0 and 255 μs with a resolution of 1 μs.

Feedforward control of the footpad force proved difficult, because of rapid fatigue combined with the complex geometry and mechanical properties of the muscles, ankle joint, and footpad. To improve performance, a closed-loop controller was implemented so that the force sensor provided a feedback signal to the computer (based on Chizeck *et al.* [2]). The four muscles were controlled as one functional unit rather than separately.

IV. Results

Although tripolar cuff electrodes provide effective shielding from asynchronous EMG and other external signals, there was considerable pickup of stimulation artifact and the compound action potentials (CAP) from the stimulated muscles. Two methods were used to reduce the amplitude of artifacts in the ENG signal: 1) high-pass filtering and 2) synchronization of sampling and stimulation.

1) In high-pass filtering, the frequency distributions of ENG and EMG recorded by tripolar nerve cuff electrodes are largely nonoverlapping (reviewed by Hoffer [6]). Much of the EMG contamination of the cuff signal could therefore be filtered out with a sharp high-pass analog filter set at 1000 Hz (Ithaco model 4302, set at 80 dB/decade).

2) Because the times of stimulation were known and both the stimulation artifact and the EMG CAP's were of known duration, it was possible to reduce the artifact pickup substantially by only sampling the cuff signal at the end of each interpulse interval; i.e., the sampling was locked to the stimulation, resulting in a nerve signal sampling frequency of 100 Hz.

Fig. 2, top trace, shows four superimposed recordings of the cuff signal during FNS. Each record was 40 ms long and thus included four stimuli, one for each muscle. The four traces are synchronized to the time of stimulation of the soleus muscle. Each time a muscle was stimulated (labeled "Stim"), the first event in the cuff signal was the stimulation artifact: a narrow spike that varied in amplitude depending on the stimulation intensity and which muscle was stimulated. This was followed by passive conduction into the cuff electrode of the compound EMG volley generated by the stimulated muscle: a slower large amplitude wave (labeled E+N). The shape of this EMG burst depended on the stimulus intensity and the muscle being stimulated, and this was very repeatable. The neural signal itself was the higher frequency signal, 5–10 μV in amplitude, onto which these artifacts were added.

The data in Fig. 2 were obtained by recording the cuff signal on an FM tape recorder while the computer controller the stimulation intensity using external force feedback, as described above. The signal was then replayed and sampled at a high rate (20 kHz) to produce the figure. For large positive artifact amplitudes, the amplifier saturated (marked AS) and for large negative amplitudes the tape recorder overloaded (marked TO), causing the signal to be zero during the overload.

The effects of filtering the nerve signal are seen by comparing the top and bottom traces in Fig. 2. In the top panel, filtered between 65 Hz and 10 kHz, the pickup of the EMG volley showed up very clearly. In the bottom panel, the same signal was further high-pass filtered at 1 kHz, and the EMG contamination was largely removed.

We attempted the approach of removing the remaining artifact by shunting the source electrode or switching the gain of the amplifier to a low value (as described by, e.g., Knaflitz and Merlertti [9], but this approach failed because the switches created more noise than was removed when applied to the high gain ($>$ 100 000) amplifiers used in this study. Instead, we blanked out the stimulation artifacts by sampling the ENG only during periods in between artifacts. This was done by means of a rectifier/integrator (Bak RBI-1) that had an adjustable integration period and was reset in synchrony with an external clock, supplied by the stimulator. It was thus possible to integrate the ENG signal in bins that lasted 3–4 ms and that included only data from the last part of each interstimulus interval (horizontal bars labeled "sample" in Fig. 2). The ENG signal occurring during each of these periods was rectified and integrated, resulting in a single value for each bin that was sampled by the computer just before each new stimulation pulse was elicited.

The validity of this method of noise suppression was demonstrated by the following experiments. The muscles were first stimulated in a quasi-sinusoidal manner to generate a force profile shown by the solid trace in Fig. 3 (top panel), ranging between about 0.5 and 5 N (similar to typical loads on the footpad during standing or walking). The application of this force profile on the footpad gave rise to the ENG signal shown by the solid trace in Fig. 3, bottom panel (labeled "Normal"). Without changing the setup, conduction in the nerve was then blocked by infusion of 0.4-ml lidocaine into the blocking cuff.

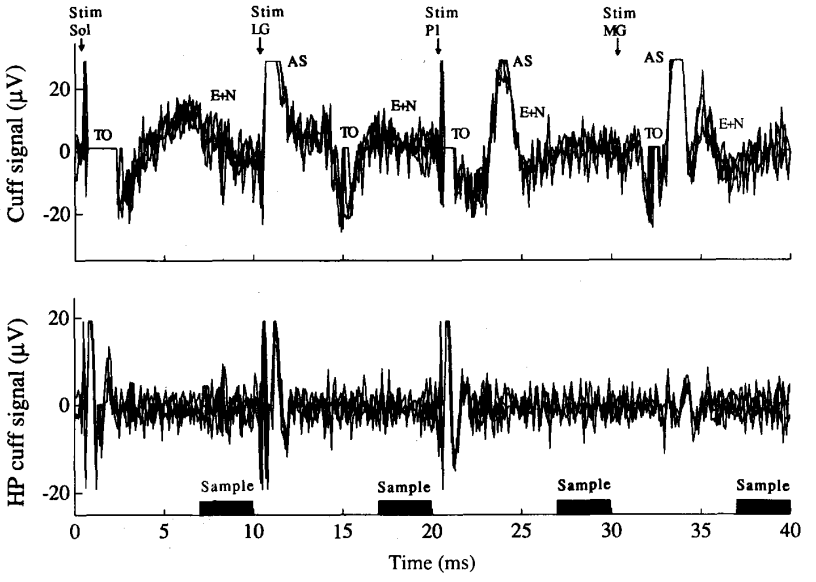

Fig. 2. Raw signal recorded from a tibial nerve cuff while four nearby muscles were stimulated. Four sweeps of data are shown superimposed. In the top trace the bandwidth was 65 Hz–10 kHz, and the compound EMG volleys (labeled "E+N") were clearly present. In the bottom trace the signal was further high-pass filtered at 1000 Hz and the EMG pickup was practically removed. Horizontal bars show the periods during which the cuff signal was bin-integrated and sampled by the computer. AS=Amplifier saturation, TO=Tape overload. Further description in text.

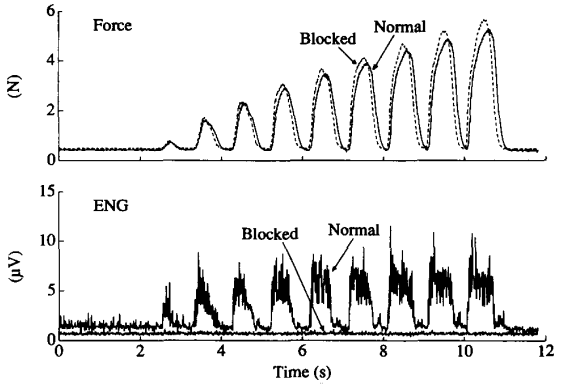

Fig. 3. Validation of the artifact removal method. Solid traces show the force applied on the footpad (top) and the normal sampled ENG (bottom) during quasi-sinusoidal stimulation of the four plantarflexor muscles. The dashed traces show a similar trial, after sensory nerve conduction between the foot and the recording cuff was blocked. Further description in text.

After about 30 min, the ENG signal recorded from the tibial nerve was insensitive to touching and squeezing of the foot, demonstrating that afferent nerve conduction between the foot and the recording cuff was completely blocked. The stimulation of the four muscles was then repeated, and a similar force profile was generated, shown by the dashed line in Fig. 3 (top panel). The sampled nerve cuff signal was now reduced to the flat dashed line in Fig. 3, bottom panel (labeled "Blocked") i.e., the cuff signal contained neither ENG, nor stimulation artifacts, nor EMG. Since the only difference was the blocking of the nerve distal to the recording cuff, this experiment showed that for the normal unblocked nerve situation, the sampling method removed all introduced artifacts and sampled only ENG activity.

The result was the same in both of the two cats that were tested by the nerve block method. After the experiments, the blocking cuff was flushed with saline. During periodic stimulation in the following 2–3 h, the ENG signal amplitude returned to normal (not shown).

V. Discussion

This study has shown that it is possible to record the very small peripheral nerve signals (in the 5–10-μV range) in a cutaneous nerve with an implanted nerve cuff electrode, during functional neuromuscular stimulation of four surrounding muscles, which generated large evoked signals from the stimulation electrodes and from the muscles (in the 10–100 mV range). The technical problems of pickup of compound EMG and stimulation artifacts were solved by appropriate filtering and by synchronizing the stimulation and sampling epochs. The techniques are not new—synchronized bin-integration is in essence a blanking technique (previously used by, e.g., Knaflitz and Merletti [9], Babb *et al.* [1]. Minzly *et al.* [12])—and use of high-pass filtering to separate neural and muscular components is a standard technique (see, e.g., Gordon *et al.* [3]). However, it has not previously been shown that the huge amounts of artifacts produced by an FNS system operating on muscles surrounding the nerve can be rejected by a tripolar cuff electrode and relatively simple means of data processing. The applications of the ability to record peripheral neural information during FNS are potentially very wide and the technique is immediately usable.

In human FNS systems, it may be expected that artifact problems could be smaller than those presented here, because larger distances could be available between muscle stimulation electrodes and nerve cuff recording electrodes. On the other hand, in practical human FNS applications the number of muscles stimulated may have to be considerably larger than four, which would increase the amount of artifacts. Based on the duration of compound EMG signals observed in this study, as long as at least 10 ms elapse between any two

stimuli, it should be possible to sample noise-free ENG by this method. This can be obtained if subsets of muscles are stimulated simultaneously, rather than in a random or time-distributed manner. Furthermore, in human applications the stimulation frequency can probably be lower than the 25 Hz used in the cat experiments, thereby leaving longer artifact-free periods available for sampling between stimuli.

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Frequency Content of Whole Body Gait Kinematic Data

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***Abstract*—We analyzed the frequency content of camera image plane data for markers on each body segment during gait. For each segment, we determined the low-pass filter cutoff frequency that balanced the amount of signal distortion and the amount of random noise passed. The frequency content of the data as measured by the cutoff frequency differed for different body segments, being higher for the lower limb segments and lower for the head. This result supports the hypothesis that the frequency content of the kinematics of segments decreases caudal to rostral.**

I. INTRODUCTION

Filtering of gait kinematic data is used to improve the signal-to-noise ratio (SNR) of the displacement information and time derivatives estimated from position data. The frequency content of normal gait is generally considered to lie within a narrow band, with the upper limit between 4 and 6 Hz [11], [22]. The band-limited nature of gait was also described in terms of the limited number of Fourier coefficients needed to generate functions that faithfully reproduce gait data [6]. Fourier coefficient reconstruction of displacement data is generally done using the number of coefficients corresponding to a frequency content of less than 15 Hz. Force plate data have been shown to have a broader spectrum, between 15 and 75 Hz [2], [10], that must be reflected in the kinematics of the foot or footwear, but would not necessarily appear in foot marker kinematic data.

The narrow-band nature of gait allows use of low-pass filters to improve the signal-to-noise ratio (SNR) of gait kinematic data. An appropriate cutoff frequency is selected to reject high-frequency noise with least attenuation of the actual kinematic data. Since the frequency spectrum of noise overlaps the frequency content of gait data, a compromise is necessary in selecting the cutoff frequency. Assuming that noise has a broad spectrum compared to the signal spectrum, Winter [12] proposed an approach to determine the optimum low-pass filter cutoff frequency that balances the root mean square error (RMSE) due to noise passed and the RMSE due to true signal attenuated. Applying the RMSE method to digitized kinematic data, he concluded that a cutoff frequency near 6 Hz could be applied to kinematic data before estimating derivatives, although he did note some variation in the frequency content of the different body segments.

Cappozzo [4], [5] has suggested that the frequency content of the kinematics of various body segments may be different. Specifically, he has proposed that dynamics are increasingly damped from caudal to rostral so that the head motions have a narrower frequency content than the lower body. The purpose of this communication was to test the hypothesis that kinematic data of different body segments have different frequency content by using the RMSE method to define the upper frequency bound of useful kinematic data for each body segment.

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