Feasibility of Using Peroneal Nerve Recordings for Deriving Stimulation Timing in a Foot Drop Correction System

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ABSTRACT
The objective of this research was to demonstrate the potential of using peroneal nerve activity to derive timing control for stimulation in foot drop correction and to attempt recording and stimulation through the same electrode. Two subjects were implanted with cuff electrodes on the peroneal nerve. An input domain was derived from the recorded electroneurogram (ENG) and fed to a detection algorithm based on an Adaptive Logic Network (ALN) for predicting stimulation timing. A switching circuit was furthermore built for switching between stimulator and recorder for combined use of the cuff electrode. The detection was successful, but the accuracy depended on the signal to noise ratio of the recorded ENG. The switching circuit successfully allowed for simultaneous recording and stimulation through the same cuff electrode. We conclude that the peroneal nerve can potentially be used to record sensory information for derivation of a stimulator control signal in a foot drop application, while at the same time being stimulated to activate foot dorsiflexors.

KEY WORDS: adaptive logic networks, foot drop correction, functional electrical stimulation, natural sensors.

INTRODUCTION
The classic foot drop FES orthosis, originally introduced by Liberson (1), activates dorsiflexors in the affected leg of hemiplegic users by stimulating the peroneal nerve using surface electrodes. The stimulation is timed with a heel switch so that the stimulation is active in the swing phase of gait, and thereby clears the toes of the ground. Several systems based on this principle has been commercially available for many years, such as the Odstock Dropped Foot Stimulator (2). The practical use of these devices have been somewhat limited because of problems related to the use of the systems (3). The surface electrodes need to be precisely located to produce the required motor output, a task which is difficult for the average hemiplegic user. The surface stimulation may be painful for some, and might cause skin problems. The heel switch is also difficult for some users to position in the shoe, and the switch can be a frequent source of malfunction due to the mechanical stress. Finally, these devices are typically mounted on or just distal to the knee, which is cosmetically undesirable for some.

In order to solve the problems caused by the use of surface stimulation electrodes, a number of...
implantable stimulators have been developed over time (4–7). These are typically single or multiple channel stimulators somehow attached to the peroneal nerve through a nerve electrode and are activated through a telemetric link. However, all of these implanted stimulators are controlled using a heel switch. Different control strategies have been applied in order to solve the problems related to the heel switch, such as the use of gyroscopes (8), accelerometers (9), or tilt sensors (10). All of these control the stimulation based on properties indirectly related to the foot to floor contact. From a control point of view, it would be desirable to measure a parameter directly related to the foot to floor contact. To eliminate some of the problems associated with the use of external sensors (eg, donning and doffing), it would be desirable to use implantable sensors. However, implantable heel switches or force sensors do not exist, which is why it has been suggested that afferent neural signals, such as those recorded from the sural or calcaneal sensory nerves just proximal to the ankle be used as a sensory feedback signal in an implanted system (11, 12).

In previous work (13), we have shown the potential of using sural ENG for providing stable control for timing stimulation. We used a combination of simple signal processing and an adaptive logic network followed by restriction rules, and showed that the derived stimulation timing was accurate and reliable. The sural ENG was recorded by a chronically implanted nerve cuff electrode placed on the sural nerve a few centimeters proximal to the ankle and connected to an implanted neural telemeter (14). Along with an implanted multichannel stimulator and a cuff electrode implanted on the peroneal nerve proximal to the knee, the system allowed the subject to walk barefooted with no external devices mounted below the middle of the thigh. However, the implantation procedure was extensive as it required two separate devices (two cuff electrodes, two sets of cables, and two electronic devices). It would therefore be of great practical value if the extent of the surgery could be reduced to implantation of a single device, preferably localized to a smaller region of the leg. To do this we have tested whether a cutaneous nerve signal from the foot could be recorded from the common peroneal nerve proximal to the knee using the same electrode as was used for stimulation to dorsiflex the foot. The common peroneal nerve branches off the sciatic nerve distal to the knee, curves laterally around the knee before it divides into the deep peroneal and superficial peroneal nerves (15). The deep peroneal branch mainly innervates the tibialis anterior and other muscles (foot dorsiflexors and inverters) around the ankle and foot. It also contains some fibers from cutaneous receptors around the first and second digit as well as afferent fibers from proprioceptive sensors in the muscles. The superficial peroneal branch mainly innervates the foot everters, but it also contains fibers from cutaneous receptors on the lower leg and the dorsal side of the foot and all the toes as well as fibers from proprioceptive sensors.

We mainly tested the quality of the signal for detection purposes by using it as an input to the detection system previously reported (13), and compared the performance to that of using the sural nerve signal. We also investigated whether the recorded signal contained significant properties from proprioceptive sensors using recordings of the nerve activity and of ankle angle. Using hemiplegic subjects we assumed that the content of muscle efferent signals were negligible. Finally, we tested if it was possible to record the peroneal nerve signal concurrently with stimulating, using the same cuff electrode.

**METHODS**

**Instrumentation**

As part of the procedure for testing the implantable foot drop stimulator developed at our center, two hemiplegic patients were instrumented with a 12-polar nerve cuff electrode on the common peroneal nerve a few centimeters proximal to the knee (7). The electrodes had four tripolar sets of contacts with the end contacts connected on the cuff. The eight lead wires to the electrode were taken out through the skin for an initial 8-week period, while the electrode was allowed to stabilize around the nerve. During this period stimulation parameters and stimulation channels were tested for suitability to produce a useful dorsiflexion movement of the foot. After the 8 weeks, a two-channel stimulator with the chosen parameters was...
built and the percutaneous wires were replaced with the stimulator placed subcutaneously on the middle of the thigh. This procedure allowed direct electrical access to the cuff in the initial period, which made the present study possible.

Subject A, a 57-year-old male, 6 years post stroke with a right side hemiplegia and foot drop, was instrumented with a 26-mm long, 4.6-mm inner diameter (ID) cuff electrode on the peroneal nerve, (Fig. 1). Due to mechanically evoked noise problems (at heel strike) only one of the four channels were used for recording. This was thought to give a lower signal to noise ratio than if all four channels had been connected to form an interface to the entire circumference of the nerve.

Subject B, a 35-year-old female, 5 years post stroke, with a left side hemiplegia and foot drop was instrumented with a 20-mm long, 4.6 mm ID cuff electrode on the peroneal nerve (Fig. 1). All four channels were connected during recording sessions to form a single channel resembling the configuration of a tripolar cuff electrode. Subject B also had a 30-mm long, 2.8-mm ID cuff electrode placed on the sural nerve just proximal to the lateral malleolus on the left ankle, with lead wires across the knee to an implantable neural telemeter positioned subcutaneously on the lateral side of the thigh approximately 25 centimeters proximal to the knee (Fig. 1). The implantable telemeter was powered through a magnetic field and transmitted the amplified ENG through the skin using a frequency modulated carrier signal (14).

Recording System

For the gait detection experiments, the peroneal ENG was recorded using a battery powered neural amplifier from Microprobe Inc. (Carlsbad, CA), with a gain of 80 dB, followed by an additional amplifier stage of 20 dB. Before rectifying and integrating into 10 ms bins, the ENG was further filtered with an 8th order Chebyshev type I band pass filter with an 800-1300 Hz pass band, i.e., a total gain of 100 dB in the pass band. The sural ENG was recorded using the receiver device which on total gave a gain of 97.6 dB, followed by the same type of band pass filter and bin integration. All ENG for these experiments was recorded without stimulation for foot drop. The rectified and bin-integrated ENG is referred to as RBI ENG, and the amplitude indicates the average nerve activity in the bin.

In order to create a target signal for training the detection system and for calculating performance measures, force sensitive resistors (FSRs) were mounted under the affected foot at the heel and the lateral metatarsal. The target signal was derived by adding the heel and lateral metatarsal signals followed by a hysteresis threshold. This gave a binary signal representing the phases of gait with a high level being stance and a low level being swing.

The ENG and FSR signals were recorded by letting the subject carry the ENG amplifier and the band pass filter in a light backpack connected to a data acquisition computer with a long cable.
To compare the features of the recorded sural and peroneal ENG during gait, mean stride profiles were calculated for the RBI ENG and FSR signals by normalizing the gait cycles in time, based on the transitions in the target signal.

Analysis of Possible Proprioceptive Information

We also investigated whether the peroneal nerve signal contained considerable content originating from proprioceptive sensors during gait. To this end, ENG signals were recorded as described above while ankle angular data were obtained by means of a flexible electrogoniometer (Penny & Giles, UK). Foot switch data were used to determine the duration of the stance and swing phases, respectively. These analyses were only carried out on data from subject B.

The recorded ENG signals were sampled at 10 kHz and band-pass filtered between 1.6 kHz and 1.9 kHz (10th order digital Butterworth filter). The filter boundaries were based on maximum signal to noise ratio estimates for proprioceptive information. More specifically, filter boundaries were chosen based on the ENG power correlating maximally with angular data during the swing phase (during which cutaneous information is at a minimum).

For this portion of the study, stretches of data were chosen for analysis if they had no FES. The time series were segmented to separate swing data from stance signals. During the stance phase no proprioceptive information was expected due to the strong cutaneous signal. With regard to the swing phase data, high correlations were to be expected between several ENG features and the recorded angular data if there was strong proprioceptive information in the nerve signals. To explore this possibility, several features were studied in the time and joint time-frequency domains (including higher order spectral analysis). The following features were analyzed:

(a) Rectified and bin-integrated ENG amplitude vs. time (bin size = 20 ms),
(b) Intra-bin amplitude variance vs. time,
(c) Bin peak frequency vs. time,
(d) Most varying bin-to-bin frequency band (based on density) vs. time,
(e) Maximum bin-to-bin frequency band power change vs. time,
(f) Bin's first AR coefficient vs. time,
(g) Bin 3rd order cumulant vs. time,
(h) Maximum bin autocorrelation vs. time, and
(i) Bin 4th order cumulant vs. time.

Then, the new time series obtained from the above features were compared to ankle angular trajectories by estimating the temporal correlation coefficients between the features’ and the angles’ histories. The same was done using angular velocity and angular acceleration data. Further, average correlation coefficients (CRC) were estimated for each feature for both swing and stance phase data.

Detection System

As initially described by Kostov and the authors (16) and previously reported on in detail (13), we used ALNs with preprocessing and feature extraction to create a control algorithm, followed by restriction rules to minimize errors. The input domain was created by 11 previous samples (covering one second in total) held in a buffer with low pass filtered RBI ENG from the input. The low pass filter and the previous samples configurations were optimized for each nerve using batch processing as previously reported (13, 16). The output was calculated at a sampler-per-sample basis and each output sample was then sent to the restriction rules which would produce the detection system output meant for driving a stimulator directly (that is, by substituting a heel switch). The actual output from the ALN was real valued and was made Boolean using a hysteresis threshold to represent stance or swing as the target signal. Restriction rules are a set of algorithms which force the output from a sensor system to be within reasonable limits, for example, a series of strides lasting only a fraction of a second are clearly unphysiologic and can be ruled out. The restrictions used in this study were based on simple fuzzy logic algorithms, and are thoroughly described by the authors (17).

Two measures of performance were used to evaluate the performance of the detection system by comparison to the target signal. First, the ALN performance was calculated as the percentage of correctly detected samples on the output of the
ALN and indicated how accurately the ALN was able to predict the state of the foot (stance or swing). This measure gave an overall impression of the performance of the ALN without including the effect of the restriction rules. Then, the distribution of the detection accuracy was calculated and plotted as the relative detection time in the detection system output and compared to the corresponding event in the target signal. This was calculated for each gait event.

Switching Circuit

To make it possible to record and stimulate through the same cuff electrode, an electronic circuit was built that switched the electrodes to the stimulator while a pulse was issued, and to the amplifier in the interpulse intervals (Fig. 2). In this experiment an implantable amplifier identical to the one implanted in subject B was used to provide a complete galvanic isolation barrier and to simulate a possible implantable system. The switches were made using opto-coupled field effect transistors, as these produced very little switching noise (no charge injection through a gate capacitance). The stimulator produced charge-balanced, current-controlled pulses (1 mA, 0–255 µs pulse width) and was controlled by a PC which also sampled and processed the amplified nerve signal. The stimulation frequency was 25 Hz, and the combined stimulation artifacts and switching noise were about 10 ms in duration, which gave 30 ms of ENG data between each stimulation pulse. The 30 ms of ENG was rectified and integrated into one number representing 40 ms of activity.

One downside to stimulating and recording from the same nerve, is that the nerve will be unable to conduct afferent sensory signals after being stimulated, due to the refractory period of the nerve fiber. When taking into account the 54.5 m/s average peroneal conduction velocity measured from the ankle to the knee joint (18), the 2 ms absolute refractory period of the nerve fiber (19), and the approximately 0.45 m distance from the cutaneous innervation area to the location of the cuff, the nerve cuff should be unable to record any cutaneous information for 18.5 ms after a stimulation pulse. This is only the case though, if all sensory fibers were activated by the stimulation pulse, which is thought to be unlikely. However, the effect expected to decrease the recorded signal levels to some degree.

Recording and stimulating through the same cuff was only done with subject A while sitting. The technique was tested by recording ENG while mechanically activating the cutaneous mechanoreceptors in the innervation area of the peroneal

![Figure 2. Schematic illustrating the switching circuit for recording and stimulating through the same cuff electrode. Only one of the four channels available in the cuff were used for this experiment for simplicity.](image)
nerve while applying stimulation and comparing to when not applying stimulation.

RESULTS

Signal Properties

In both subjects the peroneal cutane mapping was found by mechanical stimulation of the skin to originate from sensors on the dorsal side of the foot, ranging from the toes to the ankle. Similarly, it was determined that the sural ENG contained mainly cutane sensory information originating from the lateral side of the foot sole ranging from the heel to the fifth digit.

From the analysis it was clear that the various ENG features investigated were at best only poorly correlated with joint angular information. The best correlation (0.217, feature e) was found between swing phase angular velocity and ENG maximum bin-to-bin frequency band power change vs. time. However, even in this case the correlation was found to be very weak. Further, this value decreased to −0.029 when stance phase data were included. For all other combinations of ENG features and angular variables, the absolute correlation coefficients were well below 0.2 for the isolated swing phase data and below 0.07 for data including the stance phase.

Based on another study (20), the absolute correlation between the bin integrated signal (feature a) and joint angles should be near 0.8 if proprioceptive information was present. However, in the present study, feature a yielded an absolute correlation coefficient of 0.087 at best (during swing). Further, all other ENG features displayed a poor correlation with joint angular data, thus clearly indicating that any proprioceptive information in the peroneal signals was very weak.

The mean peroneal ENG stride profile recorded from subject A is illustrated in Fig. 3, and the mean peroneal and sural ENG stride profiles are illustrated in Fig. 4. Both are presented with the averaged FSR signals which throughout this paper were mapped into force resembling signals using the logarithmic properties of the FSR. It was clear that the peroneal nerve signal was modulated in synchrony with the gait cycle and had features similar to the sural nerve signal. However, the background activity and the amplitude of the modulation was different. Calculating signal to noise ratios (S/N) based on the mean ENG profiles, taking the peak to peak value of the modulation as signal and the background activity as noise, gave the following numbers: peroneal ENG subject A: 1 dB; peroneal ENG subject B: 6 dB; sural ENG subject B: 14 dB. The differences in S/N ratios could be due to a number of factors: 1) the amplifier used for recording sural ENG had better characteristics and was implanted with shorter lead wires than the external amplifiers; 2) the length of the cuffs were different, that is, the longer cuff implanted on the sural nerve in subject B gave larger signal amplitudes than the peroneal cuffs (21); and 3) only one out of four channels were used to record from the peroneal cuff in subject A which was expected to give a lower modulation level.

Detection Properties

It was then investigated if the detection algorithm for extracting a stimulator control signal could be applied to the peroneal nerve signals. Two data sets were recorded for each cuff electrode. One data set was used to train a detection algorithm for each cuff electrode, while the other data set was used to evaluate the detection performance of each trained ALN. Figure 5 shows a section of the evaluation based on the peroneal ENG
recorded from subject A, while Figure 6 show a section of evaluating sural and peroneal ENG from subject A.

The average ALN performance when using the sural ENG as an input was 96.8%, while when using the peroneal ENG it was 73.4% for subject A and 92.5% for subject B. When visually inspecting the signals in Figure 5 it can be seen that ENG did not appear as uniform from stride to stride as the signals presented in Figure 6. The binary detection output in Figure 5 also indicated that the detection was not as confident (decimal output differing considerably from the binary output) as when using the ENG recorded from subject B. In the signals presented in Figure 6, it was apparent that the detection based on the sural ENG gave the most confident output, but also that the detection based on the peroneal ENG appeared correct even if the decimal output was less confident. The detection accuracies are presented in Figure 7, showing that the detection based on ENG from subject B was more accurate than that based on
ENG from subject A. It appeared that the accuracies of using sural or peroneal ENG from subject B as an input was comparable to those previously reported (13), but the accuracy of using peroneal ENG recorded from subject A seemed somewhat erratic.

Switching Circuit

The results from recording and stimulating through the same cuff are illustrated in Figure 8. The plots are arranged in four columns with stimulation activity, ankle movement, peroneal RBI ENG, and skin touch indicated in four rows. The first column shows the background activity in the peroneal nerve without electrical stimulation or skin touch. The second column shows the peroneal response to skin touch without any stimulation. The third column shows the ankle movement as a result of stimulating the nerve, with a slightly elevated level of nerve activity as a result of the deformation of skin at dorsiflexion. The fourth column shows the ankle movement and the nerve activity resulting from both stimulation and skin touch. It is evident from Figure 8 that the recorded nerve signal was modulated with the skin touch in the innervation area, both with and without electrical stimulation of the nerve. However, it must be noticed that the signal to noise ratio is considerably lower (about 14 dB) than in the data recorded during walking. The reason for this was mainly the fact that the nerve signal was filtered with a 1000–4000 Hz fourth order band pass filter, which removed less noise than the filter applied during walking. The difference in band width gave a decrease in signal to noise ratio of 7.8 dB. An additional source of thermal noise was in the internal resistance (400 Ohms) in the switching components shown in Figure 2, which furthermore decreased the signal to noise ratio with about 1 dB.

DISCUSSION

We have demonstrated that techniques used to detect gait events from sural nerve activity potentially can be applied to common peroneal nerve activity recorded proximal to the knee. We found that the peroneal ENG in both subjects was modulated with the gait cycle as previously reported (13). When investigating if the signal contained information from proprioceptive sensors, we found that the signal was at best poorly correlated with ankle movement. This poor correlation could be due to the proprioceptive information being of more complex nature than estimated. However, considering the similarity of the amplitude levels when recording during walking and when physically stimulating the dorsal side of the foot, we find it most likely that the signal recorded from the peroneal nerve mainly contained information from cutaneous sensors.
We tested the detection technique using peroneal activity recorded during gait from two subjects. In subject B, where we also recorded sural activity, the detection performance using the peroneal activity was comparable to that of using sural activity. In subject A the performance was lower and seemed unfit to control stimulation in an FES application. The poor detection performance was presumably caused by a poor signal to noise ratio (1 dB), thought to be due to some problems with the electrode, which led us to only be able to record from one of the four sets of contacts around the circumference of the nerve. It must therefore be a prerequisite for successful detection that the signal to noise ratio be higher than that observed in subject A.

We also demonstrated that we were able to extract sensory ENG from the peroneal nerve while stimulating through the same cuff. We constructed a circuit that successfully switched between the recording and stimulation equipment using opto-coupled field effect transistor technology, so we could record the nerve activity in the interval between stimulation pulses. We envision that it would be possible to base an implantable foot drop orthosis on one implanted cuff electrode located on the peroneal nerve. Implanting a device that switches between stimulation and recording with a two-way telemetry link would enable an external device to decide whether to activate stimulation or not. This technique would lead to an orthosis with no external devices located below the knee, using sensory information as a control source and a multichannel stimulator for a balanced motor output.

We believe that future work in this area should be focused on developing an optimized circuit for switching between stimulation and recording on the cuff electrode. This is a critical factor for being able to record the neural signal with an appropriate signal to noise ratio. Once this is completed, new implantable devices and new experimental algorithms can be developed and tested.

**CONCLUSION**

We have shown that sensory information recorded from the peroneal nerve potentially can be used to detect heel strike and foot lift-off for control of a foot drop stimulator. We found that if the signal to noise ratio was adequate (6 dB or higher) the detection accuracy was comparable to that found when using sural nerve activity as an input to the detection algorithms.
We also demonstrated that it was possible to stimulate a peripheral nerve and to record natural sensory information from this nerve with a single cuff electrode using a switching circuit multiplexing between stimulating and recording equipment. This may be very useful in a number of applications where natural sensory feedback can add performance to a neuroprosthesis.

REFERENCES