Control of FES Thumb Force Using Slip Information Obtained from the Cutaneous Electroneurogram in Quadriplegic Man

Morten Haugland, Andreas Lickel, Jens Haase, and Thomas Sinkjær

***Abstract*— A tetraplegic volunteer was implanted with percutaneous intramuscular electrodes in hand and forearm muscles. Furthermore, a sensory nerve cuff electrode was implanted on the volar digital nerve to the radial side of the index finger branching off the median nerve. In laboratory experiments a stimulation system was used to produce a lateral grasp (key grip) while the neural activity was recorded with the cuff electrode. The nerve signal contained information that could be used to detect the occurrence of slips and further to increase stimulation intensity to the thumb flexor/adductor muscles to stop the slip. Thereby the system provided a grasp that could catch an object if it started to slip due to, e.g., decreasing muscle force or changes in load forces tangential to the surface of the object. This method enabled an automatic adjustment of the stimulation intensity to the lowest possible level without loosing the grip and without any prior knowledge about the strength of the muscles and the weight and surface texture of the object.**

***Index Terms*— Functional electrical stimulation (FES), hand grasp, natural sensory feedback, nerve cuff electrode, neural prostheses**

I. Introduction

MOTOR function of a paralyzed hand can be partially restored by means of functional electrical stimulation (FES) [1]–[3]. The present systems are feedforward-controlled, and the user relies on his/her experience and on visual feedback. When picking up an object, the user has to estimate the appropriate stimulation intensity, which produces enough force to hold the object. This is difficult and causes many patients to use excessive force [4]. Further, the system is not able to adapt to slow changes in force output of the stimulated muscles caused by, e.g., fatigue, electrode drift, and length-tension properties of the muscles at different hand orientations.

To deal with these problems, it has been proposed to use force sensors, position sensors, or combinations thereof to provide closed-loop control of the grasp. This provides more linear control of grasp force and grasp opening [5]. Several investigators have looked into the possibilities of using artificial sensors for measurement of finger position and grasp force (e.g., reviewed by [6] or [7]). However, a problem remains: for clinical use, no sensors are available, being cosmetically and practically acceptable for mounting on the fingertips and/or joints. Another unsolved problem is that the user still has to estimate the force necessary to hold a given object. Even if perfect control of grasp force can be achieved, objects can still slip out of the grasp, and the user may still use excessive grasp force. This is also a problem for robotics. Designers of robotics and mechanical prostheses are attempting to mimic the capabilities of the human hand to adjust the grasp force to a given object by the use of slippage sensors [8]–[10].

Able-bodied persons make extensive use of the cutaneous sensors at the tips of the fingers for automatic grasp adjustment [11]. The density of sensors in the fingers is very high compared to the rest of the body. The density of low-threshold cutaneous mechanoreceptors at the tip of the fingers has been reported to be as high as 241 units per square cm [12], some of which have large overlapping receptive areas. Therefore, a large number of receptors innervated by large myelinated nerve fibers can be expected to respond to even very small mechanical events on the skin. Several studies have been performed using single unit neurography, investigating the activity from cutaneous receptors in the human skin during passive movement of the skin or fingers [13]–[17]. These studies describe four types of cutaneous mechanoreceptors, which differ mainly in their rate of adaptation to a remaining stimulus and in the size of their receptive fields. In general, the receptors are sensitive to indentation and stretch of the skin. Other studies have investigated the receptor activity during active manipulation of objects [18]–[20]. These studies show a usage of cutaneous receptor activity in automatic control of grasp force during lifting tasks, making the motor output adapt to the present weight and surface friction of the lifted object. Further slips across the skin were shown to elicit automatic motor responses that increased the grasp force. The present paper is investigating whether such function may be implemented in an FES system by use of activity recorded from a cutaneous nerve.

The recording technique used in the studies of single unit activity is not suitable for chronic use. An electrode used in several chronic animal studies of nerve activity as well as in a few human studies is the nerve cuff electrode [21]–[28]. This electrode records a weighted summation of the activity in the whole nerve around which it is implanted. It is thus less specific than single unit electrodes, but provides a stable signal even after long time of implantation. In a C6 tetraplegic

Manuscript received August 18, 1998; revised December 8, 1998. This work was supported by the Danish National Research Foundation.

M. Haugland, A. Lickel, and T. Sinkjær are with the Center for Sensory-Motor Interaction, Aalborg University, Aalborg DK-9220 Denmark (e-mail: mh@smi.auc.dk).

J. Haase is with the Department of Neurosurgery, Aalborg Hospital, Aalborg DK-9000 Denmark.

Publisher Item Identifier S 1063-6528(99)02223-5.

spinal cord injured (SCI) patient, we have earlier shown the possibility of recording activity from cutaneous receptors which is related to slip events from this type of electrode [29]–[31].

For the control of functional electrical stimulation (FES), the idea of using natural sensors as a source for feedback signals has been pursued by in number of investigations. Previous studies showed such a reliable usage of cuff electrodes to detect nerve signals from cutaneous nerves that they could be applied for feedback in an animal model of a slip-controlled grasp [21]. In that study the hindlimb of an anesthetized cat was used as a model of a finger holding an object by FES. The recorded nerve signal from the tibial nerve was shown to contain information about slips across the skin of the paw. This information could be used to control the stimulation intensity and to compensate for slips. In two other studies, the cutaneous information from the sural nerve [23] and the calcaneal nerve branch of the posterior tibial nerve [28] was used for the correction of human footdrop. Peaks in the nerve signal marked the occurrence of heel-to-floor contact and could be used for the control of a peroneal stimulator dorsiflexing the foot during the swing phase of gait. Other investigators have used intrafascicular electrodes for the recording of signals from muscle afferents [32]. They showed the possibility of providing closed-loop control of ankle angle in an animal study by stimulating the ankle extensors against a load. As feedback they used the muscle spindle activity from tibialis anterior and lateral gastrocnemius muscles recorded with dual-channel, intrafascicular electrodes in the tibial, and common peroneal nerves. Popovic *et al.* [33] have used nerve cuff recordings in combination with an artificial neural network to determine the timing of events in the gait-cycle of walking cats. Recordings from the phrenic nerve have been used for synchronization of a respiratory pacemaker [34].

In the present study, we have focused on developing an experimental system for the control of thumb force in lateral grasp (key grip). The system should use the slip-related nerve responses recorded with a cuff electrode as previous animal experiments have proven feasible [21]. Possibly any other useful property of the nerve signal should also be applied, such as the relation between skin stretch velocity and nerve signal amplitude [35]. The present paper describes experiments performed to develop an algorithm for the control of the grasp force involving the nerve signal. This also includes a description of the performance of the system in a controlled experimental setup. The work has been described in more detail by Lickel [36].

II. Methods

A. Subject

The volunteer was a 25-year-old tetraplegic male with a complete C5 spinal cord injury (two years postinjury). He had no voluntary elbow extension, no wrist function, and no finger function, and used a splint to keep his wrist in position. He had partial sensation in the thumb, but no sensation in the second–fifth fingers.

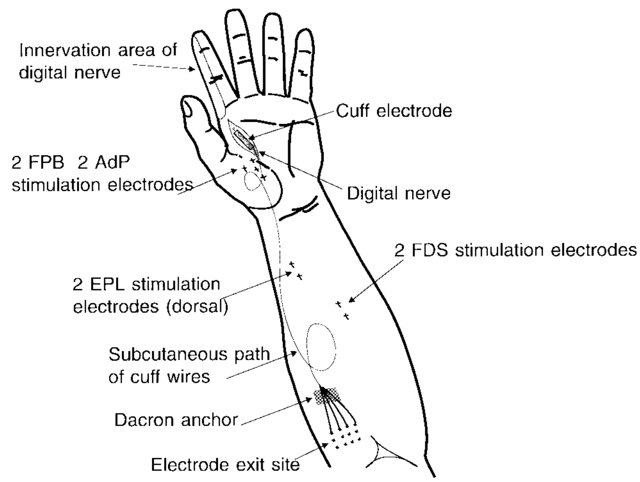


Fig. 1. Implanted devices in left arm of tetraplegic subject. The innervation area of the digital nerve is also shown. Crosses mark the active site of the stimulation electrodes: extensor pollicis longus (EPL), flexor pollicis brevis (FPB), adductor pollicis (AdP), and flexor digitorum superficialis (FDS). All wires from cuff and stimulation electrodes were routed subcutaneously to a common exit site.

The study was approved by the local Ethics Committee and conducted in accordance with the Declaration of Helsinki.

B. Implanted Devices

The subject was implanted with a nerve cuff electrode on the volar digital nerve to the radial side of the index finger branching off the median nerve (from hereon referred to as "the digital nerve," Fig. 1). The implantation was carried out during general anesthesia. The cuff was a 2-cm-long silicone tube (produced by dip-coating a mandrel with an RTV silicone adhesive, MED-1137 from NuSil) with an inner diameter of 2.6 mm and a wall thickness of approximately 0.8 mm. It had three circumferential, multistranded stainless steel electrodes (Cooner wire, AS 634) placed on the inside of the tube; one at each end and one at the center as described by Hoffer [25]. Further a reference electrode (same material) was mounted on the outside of the cuff with MED-1137. This was done partly for practical reasons as it removed the need for an external reference electrode, and partly because it reduced pickup of external noise when the reference electrode was placed close to the cuff. The main length of the wires to the cuff was straight, and the wires were lightly twisted together in a single bundle. They were routed from the proximal end of the cuff to the dorsum of the hand, where a small loop was made for stress relief, and further subcutaneously over the wrist and around to the ventral side of the forearm, where another loop was made (see Fig. 1). The last four centimeters before they exited through the skin were coiled separately, to anchor them in the skin. The wires were tied to a subcutaneous "anchor," a 3 $\times$ 3 cm piece of Dacron mesh, just distal to the start of the coiling, to reduce the risk that external pull on the wires would spread to the cuff and the nerve.

The condition of the nerve interface was tested by impedance measurements every day of experiment. The condition of the nerve was tested by a standardized neurophysiological test, which incorporated measurement of compound action potentials (CAP) generated at the tip of the index finger and recorded at a location in the forearm (see [37] for details).

During the same surgical session, eight intramuscular electrodes (NEC, [38]) were placed in or close to the following muscles (two in each): extensor pollicis longus (EPL), flexor pollicis brevis (FPB), adductor pollicis (AdP), and flexor digitorum superficialis (FDS). Activation of these muscles was sufficient to produce a lateral grasp, i.e., finger flexion combined with thumb extension and flexion/adduction. As the purpose was to produce lateral grasp, no electrodes were implanted for the activation of finger extensors. The electrodes were implanted according to the method described by Scheiner *et al.* [39]. A probe with an insulating sheath was inserted for search of a location at which the desired muscle could be fully recruited without undesired cocontraction of adjacent muscles. The probe was then removed and the remaining sheath used for guiding the cannula containing the electrode into the right position.

Prior to the surgery the muscles were conditioned by means of surface stimulation for four hours per day in eight weeks according to the protocol of Stein *et al.* [40] (see also [41]).

C. Stimulation Apparatus and Generation of Grasp

The stimulator and control system was developed at our Center. The system consisted of an eight-channel stimulator controlled by a PC via the parallel port on a pulse-to-pulse basis. The pulses were charge-balanced, constant current pulses with a cathodic phase of fixed amplitude (15 mA) and variable pulse width in the range from 0 to 255 μs in steps of 1 μs. The anodic phase was a capacitive discharge with a current limit of 0.5 mA. The stimulation frequency for each electrode was 20 Hz (i.e., the interpulse interval, IPI, was 50 ms), and the electrodes were activated sequentially with a delay of 1 ms between two successive channels. This “stacking” of the stimulation pulses for the different channels made the tails of the stimulation artifacts overlap. This reduced the total duration of the stimulation artifacts in worst case to a period of 13 ms (7 ms between the eight pulses and 6 ms of tail from the last artifact). As the IPI was 50 ms, this left 37 ms of artifact-free nerve signal before the next pulse was issued.

To be able to control the individual muscles with a single control signal, a template was made to map the control signal into pulse widths for each muscle. By using the control (or command) signal as an index in the lookup table the sequence and balance of the activated muscles was determined (see also [42]). The template was generated in the following manner. For each muscle, the pulse width was controlled manually to determine the threshold for muscle activation as detected by palpation. It was then increased until either saturation of force or spillover to other muscles occurred. The two resulting pulse widths defined the range to be used for the specific muscle. For the thumb flexor/adductor muscles, a recruitment curve was generated. This was done by measuring the grasp force with a force transducer (described below) while ramping up the pulse width from threshold to the chosen maximum value over a period of 10 s. A piecewise linear function was fitted manually to these data to produce an inverted recruitment curve and make the force output linear. The template was designed so that the grasp went from completely open to completely closed when the command signal was increased from 0 to 100 points. The transition from thumb extension to thumb flexion occurred at a command signal of approximately 20 points. To get a smooth transition, the extensor and flexor/adductor muscles were allowed to cocontract in the command range from 10 to 30 points. Finger flexors were fully activated at a command level of ten points (i.e., before thumb flexion). This made the fingers form a base for the pressing down of the thumb.

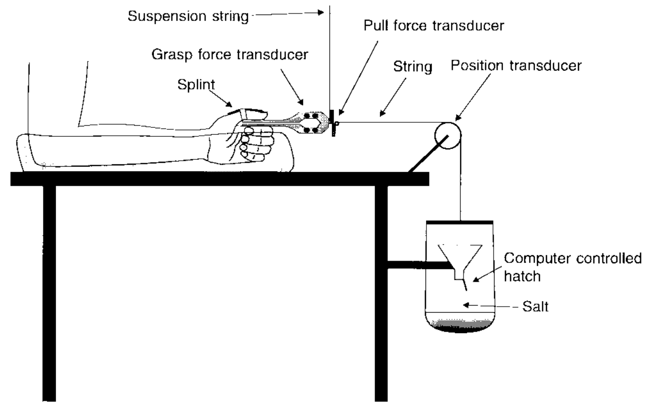


Fig. 2. Experimental setup showing the hand of the subject on the table, holding the force transducer. The pull force was provided by a mass hanging at the end of a string which was led over a pulley and attached to the force transducer. An extra weight could be added at a predetermined time by means of a computer controlled solenoid release mechanism.

D. Experimental Setup for Slip Experiments

The experimental setup is illustrated in Fig. 2. The subject was sitting next to a table with his arm stabilized in a vacuum cast. His lower arm was supported, and the 2nd–5th fingers were kept in a flexed position. A flat force sensor, 7 cm square and 1.5 cm thick, with exchangeable surface texture was placed between the thumb and the flexed index finger. The design of this sensor ensured correct measurement of grasp force, independent of the position of the center of pressure. The force sensor could be pulled out of the grasp by a weight hanging in a string over a pulley on the edge of the table. A force sensor placed in line with the string measured the pull force. A potentiometer placed in the pulley measured the position of the object. Because of the weight of the object (220 g), it had a tendency to rotate (fall down) when pulled out of the grip. To avoid this movement we suspended it with a string from the far end of the object to the ceiling, which resulted in a practically linear motion when the object slipped out.

The force sensor could be equipped with one of three different surfaces: silk, suede, or fine sandpaper (600 grit). These three materials have very different surface frictions and have been used in previous investigations of cutaneous mechanoreceptor activity [11], [18].

Since no surgery was done to stabilize the interphalangeal joint of the thumb, we kept the thumb stiff in a slightly flexed position by means of an aluminum splint mounted with adhesive tape to the back of the thumb.

E. Signal Conditioning

The signal from the cuff was amplified with a low-noise preamplifier (ADT-1 from Micro Probe Inc.) at a gain of 10 000. During stimulation of nearby muscles the signal contained stimulation artifacts and was contaminated with the strong, evoked EMG potentials generated by the stimulated muscles. As stimulation artifacts and residual EMG pickup were synchronized to the stimulation, it was possible to remove them by using only the part of each interpulse interval, which was not contaminated with stimulation artifacts and EMG. Further, EMG contains lower frequencies than the nerve signal, and can be reduced by highpass filtering (e.g., [26]). A previous description of this method was made for an anesthetized animal model [22]. This method produced a signal that represented the envelope of the nerve signal without including artifacts. In our setup, the nerve signal was filtered through an analog fourth-order filter (Krohn–Hite) set at bandpass between 1 and 4 kHz. The resulting signal was sampled at 10 kHz and digitally rectified and integrated in blocks of samples from each interpulse interval (bin-integration). This processing was done with a digital signal processor (a TMS 320C25 placed on a board in the PC), and the results of the integration were sent to the PC for further processing.

F. Slip Detection Method

The method for slip detection and following compensation was basically the same as the one earlier reported in an animal study [21]. The rectified and bin-integrated signal (RBI-ENG) was filtered through a differentiator, which subtracted a delayed and low-pass filtered version of the signal (representing the background activity) from the nondelayed RBI-ENG, low-pass filtered at a higher frequency. This procedure was computationally very simple, and adequately enhanced peaks, removed interference from slow changes in background activity, and smoothed the signal to reduce the small natural variations in the combined noise and nerve signal. Detection of slips was then done by comparing this signal to a threshold value.

As will be detailed in Section III, reducing the stimulation intensity while the subject was holding the force transducer could provoke slips. It was relatively easy to find a set of parameters (the two filter frequencies and the delay of the background signal) by trial and error, which made the system capable of catching the object when it slipped. However, to set these parameters to their optimal values, we analyzed 105 slips in 15 trials off line (a total of 300 s of data including 35 slips with each of the three different surfaces). As cost function for the optimization, we chose to use the ratio between the lowest and highest value of the threshold values, which caused proper detection of all slips in the test data without false or missing detections. Maximizing this normalized range of possible settings for the threshold value was assumed to maximize the robustness of the system. Optimization was then done by letting a computer step through a large array of possible combinations of parameters, run the test data through a filter using each of the combinations of these parameters, and calculate the range of threshold values which caused proper detection for each set of parameters. The parameters were tested in the ranges $0 < F_{\rm low} < 5$ Hz, $F_{low} < F_{high} < 10$ Hz, and 0 < delay < 20 samples (500 ms), taking 20 values of each parameter with linear spacing. The set of parameters resulting in the largest threshold range was found to be: $F_{\rm low} = 0.3$ Hz, $F_{\rm high} = 2.5$ Hz, and delay $=$ 10 samples (250 ms). For these values, the threshold could be increased with 83% relative to the minimum value (from 0.12 to 0.22 μV) without missing any of the 105 slips in the test data and without making any false detections. A threshold was chosen to be in the middle of this range to further reduce the risk of false and missing detections.

G. Compensation for Slips

When a slip was detected, the command signal (the input to the grasp template functions) was increased to stop the slip. To stop the slip as effectively as possible, but at the same time with just the necessary force, it was important to generate a fast and strong increase in force—the longer time the object was allowed to slip, the more difficult it was to catch it—partly because it moved out of the grip, and partly because it usually gained momentum. Two methods were used, either alone or in combination. One was to momentarily increase the stimulation intensity (for AdP and FPB) to maximum and turn it back down immediately after to avoid too high a force. The other was to briefly increase the stimulation frequency. Once the slip of the object had been stopped, the command signal was set to a level slightly higher than the level before the slip to ensure a better grip.

When a reaction to slip had been issued, the nerve signal was ignored for 500 ms to avoid further reactions to the same slip and to avoid interpreting the increased nerve activity caused by the increased grasp force as a new slip. Without it, the system would sometimes "wind up" to maximum command level if just a single slip was detected. The precise duration of this period was not critical.

III. RESULTS

A. Implanted Stimulation Electrodes

One of the main difficulties encountered in the project, was the placement and stability of the stimulation electrodes. The electrodes themselves have been failure free—none of them broke during the two-year implantation period—but the placement of them was less than optimal. For some electrodes, the recruitment curve was very steep and variable and for other electrodes, the usefulness was limited by cocontraction of adjacent muscles. To get useful hand function from this, it was necessary to adjust the template for muscle activation in the beginning of almost every session.

B. Nerve Cuff Electrode

At the present time, the implanted nerve cuff electrode has worked without problems for more than 820 days. Impedance measured at 1 kHz from the reference electrode on the outside of the cuff to each of the internal electrodes was followed during this period. During the first week the impedance rose from 800 Ω for each of the end electrodes and 1200 Ω for the center electrode to 1300 Ω for the end electrodes and 1700 Ω for the center. In the following period the values increased slowly to the present level of 1600 and 2100 Ω, respectively.

The cuff is a little too large to fit well in the hand of our subject. This has caused the skin in the palm covering the distal end of the cuff to get progressively thinner. To avoid breakage of the skin, we opened the skin about 200 days after implantation, and with a suture pulled some more tissue over the cuff. Use of a slightly shorter cuff (2–3 mm shorter) and/or a cuff with a thinner wall should remove this problem.

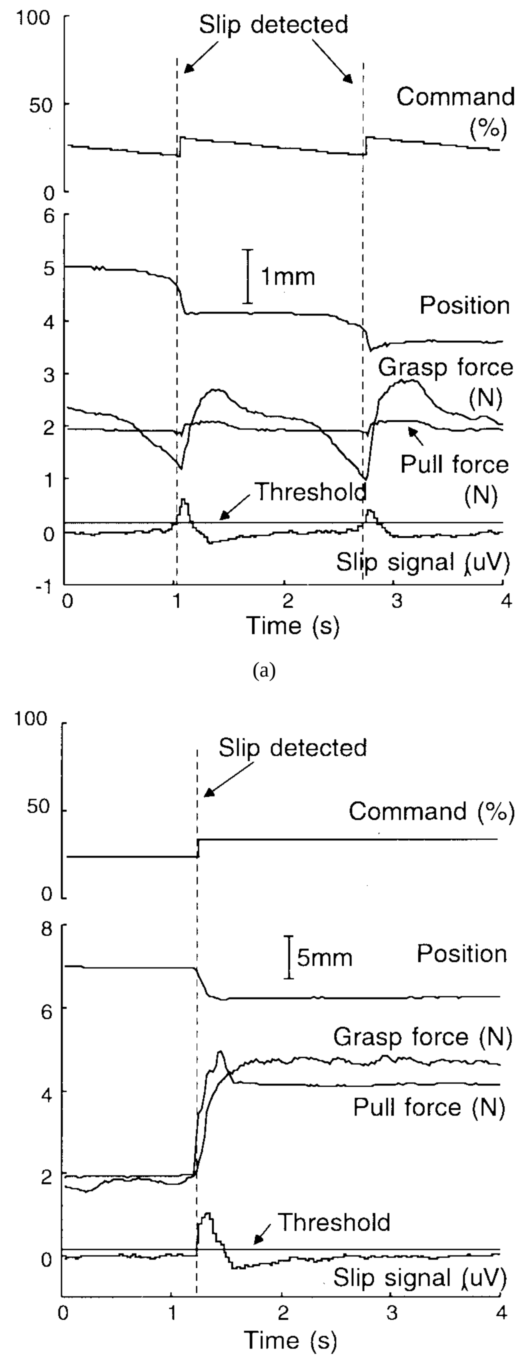
The integrity of the nerve was examined with a standard neurophysiological test [43]. Compound sensory nerve action potentials were elicited in the nerve by supermaximal, electrical stimulation of the second finger with surface (ring) electrodes. This volley was then picked up by the cuff electrode and by surface electrodes placed above the median nerve proximal to the wrist. The proximity of the recording electrode to the median nerve was verified by stimulation through the electrode, and the precise location of it was selected to get constant threshold for activation of motor fibers (3–4 mA). The amplitude of the compound action potential (CAP) was measured together with the conduction velocity and compared to results from similar measurements taken from the first and third finger. Over a period from 10 days before the implantation to 822 days after, the conduction velocity did not show any significant change (69 m/s, std. 2 m/s). The peak–peak amplitude of the CAP's did vary more (surface = 7.0 μV std. 0.9 μV, cuff = 150 μV std. 24 μV), but showed no apparent trend over time.

C. Slip Experiments

Experiments were conducted to investigate if the system was able to react and stop a slip, which was either caused by a decrease in grasp force or by an increase in external load force.

1) Slips Caused by a Decrease in Grasp Force: The force transducer was placed in the hand of the subject by the experimenter, stimulation started, and the object was then held only by the thumb and index finger of the subject. The initial stimulus intensity was chosen so that the grasp force was well above the force necessary to produce a secure grasp. The command signal was then slowly linearly decreased until a slip occurred. An example of this is shown in Fig. 3(a). Two slips were detected as the "slip signal" (calculated from the processed nerve signal) exceeded the threshold. Increasing the command with a predetermined amount (ten points) compensated for the slips. This produced an increase in grasp force, which was enough to stop the slip. Trials of 20 and 30 s were run, and the procedure of reducing the stimulation intensity to cause a slip and then compensate for it was repeated until either the object was lost or the trial ended.

2) Slips Caused by an Increase in External Load Force: Increases in external load force can be caused by weight changes of the object (e.g., a cup being filled) or if an object is being manipulated (eating with a fork, inserting a disk into a computer). Alternatively, it can be caused if an active object is being held (an umbrella in the wind or a fishing rod when a fish bites). To investigate if sudden increases in pull force could be compensated for, the following experiment was performed. While the force sensor was held in the grasp as described above, but with a constant stimulation intensity, additional weight was added at the end of the string providing the pull force. The extra weight was added by letting a solenoid open



(a)

(b)

Fig. 3. (a) Slip caused by slow reduction of grasp force and (b) slip caused by an increase in pull force (increasing weight from 200 to 400 g). A slip was detected when the "slip signal" exceeded the threshold and compensated for by increasing the command with ten points. In both cases the surface was sandpaper.

the bottom of a container with salt to make the salt fall into another container at the end of the string (see Fig. 2). The size of the aperture in the container could be adjusted to change the rate of change in load force. Trials of 10 s were performed while the stimulus intensity was kept constant (when no slip was detected). The initial stimulus intensity was high enough to produce a grasp force that could hold the object, but low enough that when extra weight was added, the object would slip out of the grasp if nothing was done to prevent it. The extra weight was added after 5 s. This caused a detectable slip, and the controller increased the command signal. Fig. 3(b) shows such an event where the weight was increased from 200 to 400 g (the data were aligned so that the salt was released at 1 s). The 200 ms delay from release of the salt until the pull force increased was caused by the opening mechanism and the time it took for the salt to drop the approximately 15 cm. The rate of rise in pull force was set to be about 10 N/s, but the kinetic energy picked up by the salt during the fall added some extra force when the salt landed in the other container. This caused the rate of rise to be larger and also resulted in an overshoot in the pull force as can be seen in the figure.

To make sure that the compensation was the actual cause of the stopping slip, experiments were performed in which the system alternatingly did and did not react to the slip. An example of this is shown in Fig. 4 in which six such trials are shown superimposed (three with and three without compensation). In these experiments, the reaction to a slip was a single stimulation pulse with maximum intensity followed by a fixed increase in command level of 15 points. As can be seen in all trials using compensation, the increased stimulation managed to stop the slip whereas in all the noncompensated trials the object slipped out.

Notice that in the noncompensated trials, the object kept slipping until it was out of the grasp. This caused the slip signal to last longer compared to the compensated trials. The compensated trials generally gave a higher peak value in the nerve signal, which was caused by the sudden increase of the grasp force.

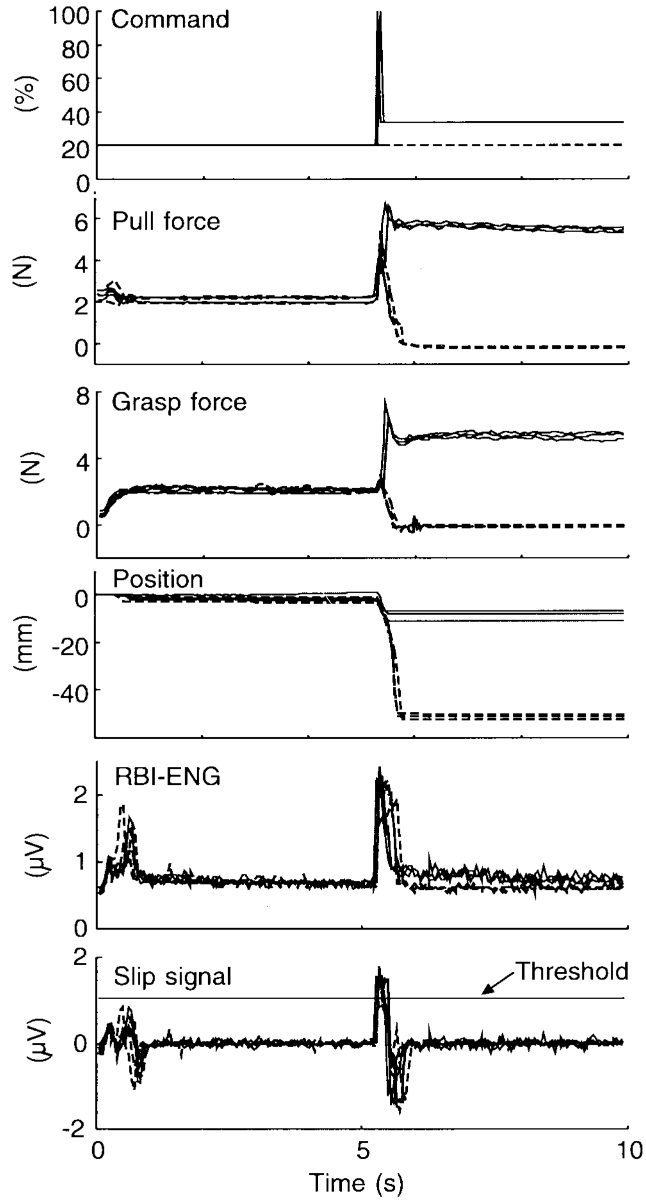


Fig. 4. Experiments with increasing load force where alternately the system *did* (solid traces) and *did not* (dashed traces) react to slips detected in the nerve signal. RBI-ENG is the rectified, bin-integrated electroneurogram. The computer released the load after 5 s, and a small delay was introduced by the release mechanism. The initial change in position and responses in the nerve signal (and slip signal) were caused by the experimenter releasing the object into the grasp of the subject when stimulation started.

D. Threshold for Slip Detection

A central parameter in the slip detection algorithm was the threshold for detection. As mentioned in Section II, we investigated 105 slips under various conditions, 35 with each of the three surfaces. All slips were caused by slowly decreasing grasp force, and in all cases the system was able to catch the object. These data were used for testing the range within which the threshold could be set without missing or false detections. With the optimal filter settings for generation of the slip-signal, the threshold could be changed from 0.12 to 0.22 μV and still detect all slips at the same sample as was originally detected. Therefore no difference will be found in the performance of the system for any threshold setting within this range. This was also tested in a randomized series of experiments with a sudden increase in load force (as described above). The surface material was suede, and the load force was changed from 2 to 4 N. Five different threshold settings were tested: 0.10, 0.13, 0.16, 0.19, and 0.22 μV in a total of 19 tests. This resulted in average slip lengths of 4.6, 4.6, 4.5, 4.5, and 4.0 mm, respectively, i.e., they showed an apparent decrease in slip length with increasing threshold, which would be a surprising result. However, the differences were statistically insignificant ($p = 0.54$ for a *t*-test comparing the slip lengths for the two lower thresholds with the slip lengths for the two higher thresholds). We therefore conclude that the choice of threshold is not important within this range.

E. Increase in Stimulation Intensity After Slip Detection

Another central parameter is how much to increase the stimulation intensity once a slip has been detected. To investigate

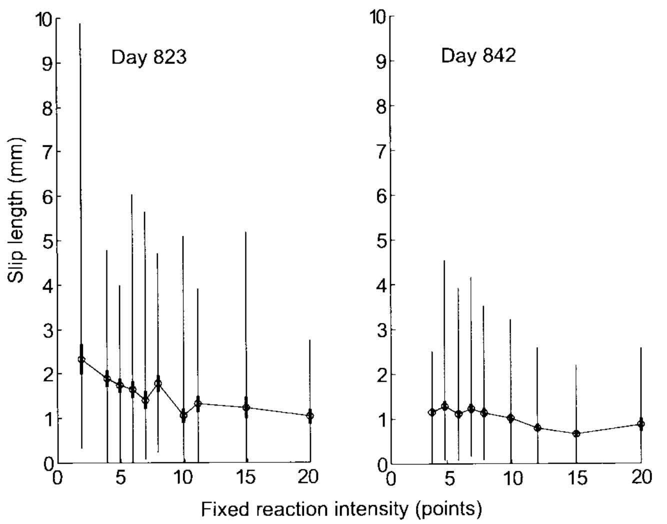


Fig. 5. Slip distances for different increments of the command signal for slips caused by decreasing grasp force at two different days. The surface of the object was suede and the pull force was 2N. The "o"s indicate the average slip distances for approximately 30 slips with each reaction intensity. The thin vertical bars show the range and the thick vertical bars show the standard error.

the significance of this parameter we performed a series of experiments with decreasing grasp force [as described above, Fig. 3(a)]. The fixed increase in command signal that was issued as response to a slip was varied in a randomized series of experiments. Approximately 30 slips were recorded with each of ten different increments of the command signal after slip. The slipped distance was measured and averaged. The results are plotted in Fig. 5. It shows data from two different days of experiments, where the overall conditions were the same (suede surface and a pull force of 2N). The markers show the average values, the thin vertical bars show the range of slip distances, and the thick vertical bars show the standard error. As expected, the slip length decreased with increasing reaction, but it can also be seen that the effect is somewhat smaller than the variability in the individual slips. The large variability of the slip lengths were attributed to the nonlinear properties of the immediate friction between the fingers and the object. Also, in spite of attempts to linearize the recruitment curves for the stimulated muscles, there were still nonlinearities in the relation between the command signal and the generated force. As the object did not always slip at the exact same command level, the differences in background force and force increments may have introduced some variability as well. These differences not only result in different slip lengths from slip to slip but also on the overall performance from day to day, as can be seen from the different average values in the two graphs of Fig. 5.

F. Improved Slip Detection and Reaction

Stimulation at 20 Hz (i.e., with an interpulse interval of 50 ms) made the reaction time from the start of a slip until a reaction of the stimulator relatively long. In worst case it could become even longer than the interpulse interval since the bin-integration averaged the nerve signal over a bin. And if a slip started late in a bin, the result might not change

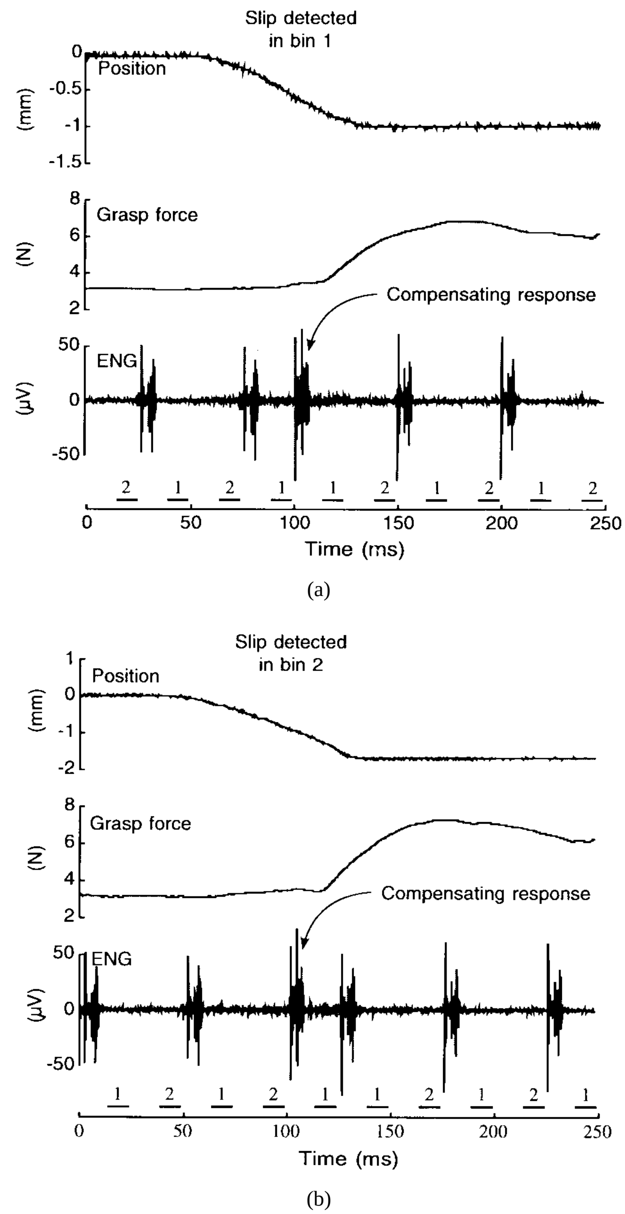
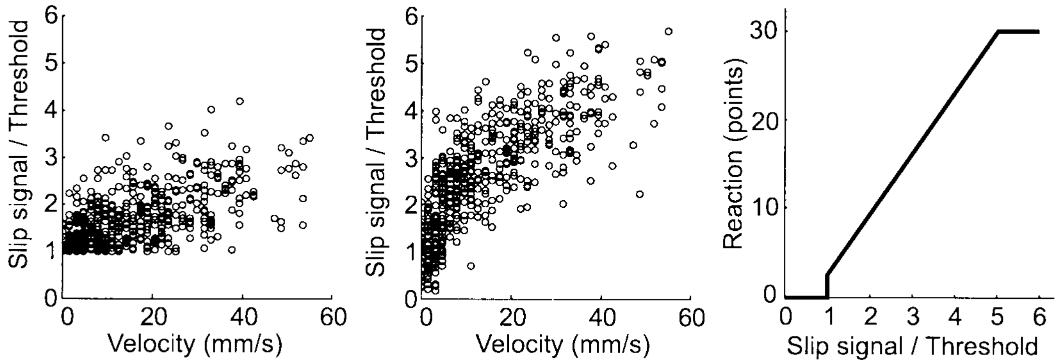


Fig. 6. Illustration of the reaction to slips occurring in the first (1) or second (2) part of an interpulse interval. In (a) the slip occurred in bin 1, and the reaction came as a pulse 25 ms after the latest pulse whereafter stimulation was continued with an IPI of 50 ms. In (b) the slip occurred in bin 2, and the reaction came 50 ms after the latest pulse. To get a comparable increase in force as in situation A, the following IPI was reduced to 25 ms whereafter stimulation was continued with IPI = 50 ms.

enough for the slip to be detected. Then the reaction could not appear until the next pulse was issued more than 50 ms later. This delay in combination with the delay in neuromuscular transmission and the delay produced during force build-up in the muscle fibers and mechanical movement of the fingers might give the object time to build up enough momentum to be impossible to catch. To reduce the reaction time of the system, the nerve signal was bin-integrated twice during an interpulse interval (IPI) as illustrated in Fig. 6. The 50 ms IPI was divided into two 25 ms bins (denoted as "1" and "2" in Fig. 6), of which the data in the last 12 ms were rectified and integrated (because stimulation artifacts lasted up to 13 ms). In Fig. 6, the movement of the object started at approximately 50 ms. The



(a) (b) (c)

Fig. 7. (a) Slip signal amplitudes at the first sample that crossed threshold and (b) the sample immediately after, related to the velocity of the sliding object at the same time. The surface was suede. (c) Chosen relation between reaction intensity and amplitude of slip signal. Below threshold there was no reaction. Above threshold the reaction would increase linearly with the amplitude of the nerve signal until a limit was reached at five times threshold.

nerve signal responded by increasing the amplitude, although not immediately. The amplitude was increased enough for the slip to be detected in the period from 75–100 ms. In Fig. 6(a), this period fell just after a regular stimulation pulse was elicited (bin 1), and the reaction could be issued already 25 ms after the latest regular stimulation pulse, i.e., 25 ms faster than with an IPI of 50 ms. In Fig. 6(b), the slip was detected in bin 2, i.e., just before a pulse would normally be issued. Thus, in this case the system was not faster than it would have been with an IPI of 50 ms. The short IPI *before* the reaction in situation A caused extra muscle force to be generated (due to the higher instantaneous stimulation frequency). To avoid very different force responses in the two situations, the IPI *after* the reaction in situation B was also reduced to 25 ms.

To test the performance of this method, 60 trials were performed, alternatingly using the double-bin integration (40 Hz), as described above, and using the single-bin integration (20 Hz) as previously used. All other system parameters were kept constant. The surface of the object was suede. The subject held the object with fixed stimulation intensity, until a slip was provoked by increasing the pull force from 2.2 to 4.2 N. At slip detection, the command was increased from 33 to 63 points, causing the grasp force to increase from 2.8 to 6.3 N. We evaluated the differences in slipped distance between the two methods. Three of the 20 Hz trials had to be excluded as the object was not caught and another three had to be excluded due to mistakes in the experimental procedure. In the remaining 24 trials using 20 Hz update of nerve information, the slipped distance was on average 4.76 mm, ranging from 1.64 to 9.43 mm. For the 30 trials using 40 Hz update of nerve information, the average slip length was 2.86 mm, ranging from 1.2 to 6.38 mm. This was a significant difference (student's t-test, $p < 0.0002$) and showed the benefit of using the increased update rate.

G. Nerve Signal Dependent Reaction Intensity

The amplitude of the nerve signal at the time of the slip may be useful as an indicator for the necessary intensity of the reaction as the nerve signal increases approximately linearly with the speed of the slip [35]. Further a fast slip requires a stronger reaction.

To investigate if the nerve signal contained information about the velocity of the sliding object, a large number of slips (573) were produced experimentally. This was done in the type of experiment shown in Fig. 3(a). The command signal was decreased with ten points per second until the decreasing grasp force caused a slip. Each slip was detected, the system reacted correspondingly, stopped the slip, and the process started over again. The value of the slip signal that caused each slip to be detected (i.e., the first value above threshold) was plotted against the velocity (=differentiated position signal) of the object at that time. This is shown in Fig. 7(a). It can be seen that there is some correlation between the slip signal and the velocity of the object. However, this correlation is more pronounced when the sample immediately after the first value is considered. This is shown in Fig. 7(b). The reason for the better correlation we attribute to the precise timing of the starting slip. For the first sample, the slip may have started at any time during the period in which the raw signal was integrated, giving rise to some variation and to a smaller amplitude of the signal. For the second sample, the slip is in progress during the whole integration period and a larger and more stable value for the slip-related nerve signal is obtained.

It was decided to make the increase in stimulation intensity (the reaction) depend linearly on the amplitude of the slip signal. Gain and offset of the relation were selected by trial and error, and the relation that was used is shown in Fig. 7(c). It was linear from a minimum reaction at threshold of five points to a maximum of 30 points at five times threshold.

This "adaptive" method was then tested against a number of trials with fixed reactions. The order of the trials were randomized to reduce the effect of time-dependent changes in performance. On average 30 slips were produced and successfully compensated for, for each of ten different fixed reaction intensities. The results are illustrated in Fig. 8. The average slip length for these slips are shown as "o"s, showing also the standard errors with vertical bars (same data as left graph in Fig. 5). For comparison, single trials resulting from the adaptive method are shown with "x"s. The advantage of the adaptive method shows by most of the points (26 of 36) being below the average curve for the fixed reactions. This can be interpreted either as if it is possible to produce shorter slips

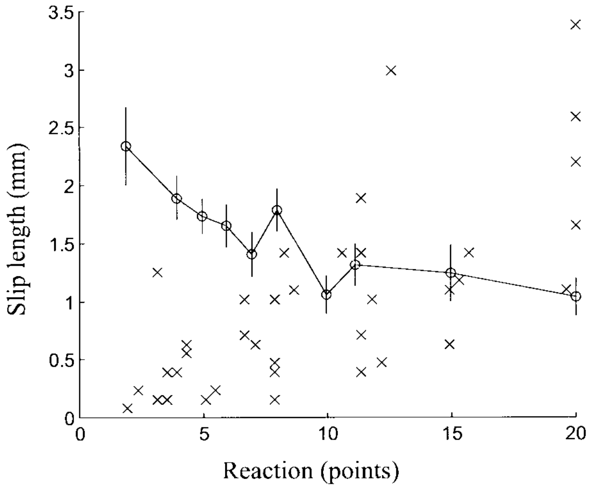


Fig. 8. Average slip distances for different fixed reaction intensities (o) compared with slip distances for individual slips as resulting from the adaptive method (x). The bars show the standard error for the average values.

for a given average reaction intensity, or as if a given average slip length can be obtained with lower stimulation intensities.

To show the potential of the method for controlling the stimulus intensity with slip-related nerve information, we performed an experiment which was a combination of decreasing stimulation intensity and changing pull force. The idea was that by continuously decreasing the stimulation intensity, it might be possible to probe the minimally necessary stimulation intensity. Thereby, the system could automatically adapt to the task at hand. To produce a changing load, we hooked up a leaky container at the end of the string that provided the pull force. A trial was started with a minimum pull force of 2 N (see Fig. 9), and after a few seconds, water was added into the container, which made the pull force increase. This usually caused the occurrence of one or more slips, which again caused the system to increase the stimulation intensity and stop the slips. As soon as no more water was added, the leak in the container caused the pull force to slowly decrease. As no more slips occurred, the constantly decreasing stimulation intensity returned the grip force to a level suitable for the lower pull force. The method hereby managed to let the grasp force follow the pull force as it can be seen by the dashed trace in Fig. 9, which is the grasp force smoothed with a 4-s long Hanning window.

H. FES Generated Grasp Compared to Normal Grasp

To give some numbers on the performance of the system, we asked five neurologically intact subjects to perform similar tasks as described above. They were asked to hold the force transducer in a lateral grasp, with as low force as possible, without slips in the static phases and without losing the object when the extra weight was added. They were blindfolded and wore hearing protection to reduce the influence of feedback information other than the cutaneous input. For the FES system, the initial stimulation intensity was set low enough for the object to slip out of the grasp if there was no reaction to the slip.

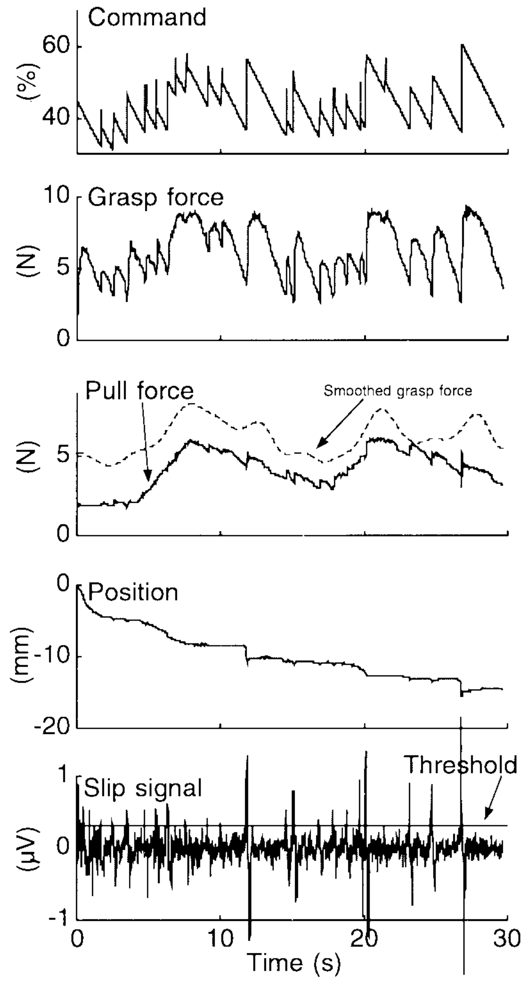


Fig. 9. Combined increase in load force and decrease in grip force. While the stimulation was slowly decreased, the load was increased by pouring water into a container hanging in the string providing the pull force. The container had a hole in the bottom so the water leaked out. See experimental setup in Fig. 2.

Results for rapid increases in pull force (approximately 10 N/s) are shown in Table I. For low weights and rough surface, the spinal cord injured subject with the system performed equally well as healthy subjects. For higher weight increases and for more slippery surfaces, the system performed less well. For the most difficult tasks, the system gave up whereas the healthy subjects could still do it although they did acknowledge the difficulty of it. The inability of the system to catch the slippery objects was attributed to the weaker muscles of our subject as the slips were always detected properly. The responding increase in force was simply not sufficient.

Results for slow increases in pull force (0.25 N/s) are shown in Fig. 10 and Table II. Five recordings of 20 s each with three different surfaces (silk, suede, and sandpaper) were run in randomized order. The object was each time placed in the subject's grasp by the experimenter, and the load force increased linearly from 2 to 5 N. For the FES system, the reaction intensity was set to a range between five and 30 points depending on the amplitude of the nerve signal. The grasp force, load force, and object position from one healthy

TABLE I
SLIP DISTANCES FOR DIFFERENT SURFACES AND LOADS WHEN A WEIGHT WAS ADDED. FOR COMPARISON WE ALSO SHOW THE SLIP-DISTANCES FOR FIVE NORMAL SUBJECTS SITTING IN THE SAME SETUP. THEY WERE ASKED TO HOLD ON TO THE OBJECT AND CATCH IT WHEN THE WEIGHT WAS ADDED

	345g→521g	345g→646g	345g→734g
SCI-subject, Sandpaper range	1.2±0.1mm 1.03, 1.36	2.4±0.6mm 1.4, 3.81	3.5±0.5mm 2.96, 4.84
SCI-subject, Suede range	4.12±0.35mm 3.86, 4.37	slipped out	slipped out
Healthy subjects, Sandp. range	1.3±0.7mm 0.19, 2.93	2.4±1.2mm 1.51, 5.27	2.8±0.9mm 1.40, 4.94
Healthy subjects, Suede range	3.7±2.9mm 0**, 7.88	6.4±3.7mm 1.12, 12.05	6.6±3.0mm 1.45, 12.71

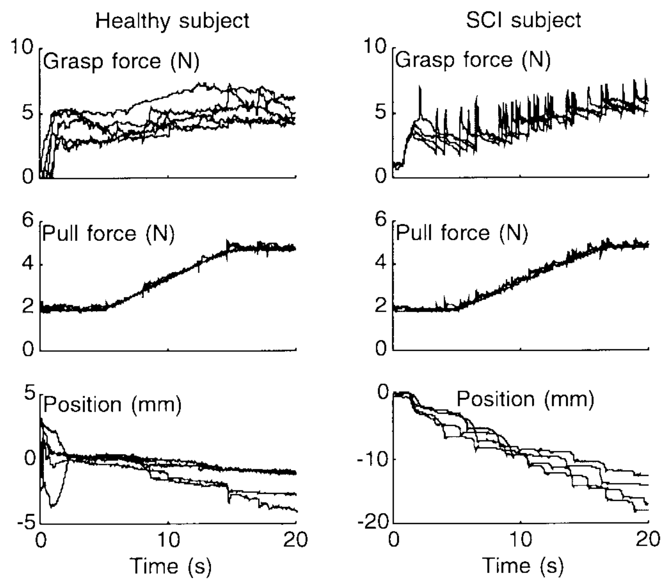


Fig. 10. Comparison of performance of healthy subject and SCI subject. The load was slowly increased from 200 to 500 g, and the healthy subject was instructed to use as little force as possible without loosing the object. The surface of the object was suede. The range of grasp forces was comparable, but the object slipped further for the SCI subject using the FES system.

TABLE II
AVERAGE FORCE (±1 STANDARD DEVIATION) USED BY THE DIFFERENT SUBJECTS THROUGHOUT FIVE TRIALS OF THE TYPE SHOWN IN FIG. 10. THE SUBJECTS WERE INSTRUCTED TO KEEP AS LOW FORCE AS POSSIBLE

	Silk	Suede	Sandpaper
Healthy #1	1.89±0.37	1.94±0.33	1.63±0.31
Healthy #2	1.87±0.16	1.65±0.25	1.73±0.21
Healthy #3	1.04±0.25	1.11±0.27	0.77±0.26
Healthy #4	1.44±0.27	1.49±0.38	1.10±0.24
Average	1.56±0.43	1.55±0.42	1.3±0.46
SCI-subject	1.92±0.17	1.40±0.04	1.02±0.09

subject and from the paralyzed subject are shown for five runs in Fig. 10.

All healthy subjects used a force (averaged over the whole trial) which was comparable to the force produced by the paralyzed subject with the FES system (see Table II). This was the case for any of the three surfaces used. The differences between subjects were likely caused by differences in surface friction of the skin.

An important difference, however, was that the FES system had to produce small slips each time the minimum force was probed whereas healthy subjects usually could determine the minimum force without visible slips occurring. The latter can be seen in the five superimposed traces for the healthy subject in Fig. 10 (left). Only four times during these five 20-s trials did the object make sudden slips. The slips indicated that the subject was using a force very close to the threshold for slippage, but only very few slips occurred in all five subjects. As the FES system required repeated slips in order to track the minimum force, the object slipped further compared to the healthy subjects. In order to deal with this in a future functional system, one will probably have to make a tradeoff between how long an object can be held and how fast the grasp force can follow load-changes.

IV. DISCUSSION

It was demonstrated that signals related to mechanical events on the skin of the index finger could be recorded from a nerve cuff electrode implanted on a digital nerve in the palm of a tetraplegic subject. These signals most likely originated from cutaneous mechanoreceptors, as these are very numerous in the fingers and transmit information via large myelinated fibers. There was possibly some contribution from joint receptors, but it was not attempted to quantify the influence of this. It was further shown that the amplitude of the signal could be reliably recorded even during electrical stimulation of several muscles in the forearm and hand.

The amplitude of the nerve signal clearly signaled phasic events on the skin, such as changes in contact force, skin stretch and slips across the skin. In a controlled setup, this information was used to initiate compensatory reactions in stimulation intensity when an object held in a lateral grasp started to slip. This allowed the system to automatically catch the object. This was shown for slips caused by a decrease in internal muscle force as well as for slips caused by an increase in external load force tangentially to the skin. The performance of the system with respect to the slipped distance when extra weight was added, was comparable to the performance of healthy subjects performing a similar task.

The ability to compensate for slips could also be used for other purposes than simply catching objects that unexpectedly started to slip. In this paper the minimally necessary stimulation intensity to hold a given object was probed by deliberately introducing a slip which was then immediately compensated for. By slowly decreasing the grasp force until a slip occurred, the minimum stimulation intensity could be measured as the level used just before the slip was detected. The stimulation intensity could then be set to a level just above that. This produced a secure grasp with minimum grasp force. Repeated application enabled automatic adjustment of the grasp force so that it was within a range comparable to the forces produced by healthy subjects performing the same task. This was done automatically and required no interaction of the user. For a slow increase in load force the FES generated grasp force was on average comparable to the force produced by healthy subjects. However, healthy subjects performed better with respect to slipped distance, as the function of the FES system depended on small slips to make the grasp force follow a changing load force. This is not necessary for the natural system. This difference may turn out to be a significant limitation for the use of the method in clinical systems.

We have demonstrated the function of a system that simply sampled the rectified bin-integrated nerve signal at a rate of 20 Hz and increased the stimulation intensity with a fixed amount when a slip was detected. However, by sampling the processed nerve signal at the double rate and using the nerve signal amplitude for determining the strength of the reaction, the system became faster and on average it resulted in shorter slips.

A number of open questions still exist. The correct amplitude of the reaction to compensate for a slip depends on many parameters, such as the reason for the slip (whether it is an external or internal disturbance, is the object being pulled away from the grasp or is it perhaps being pushed further into the grasp, is the object rotating, etc.), the weight of the object, the immediate friction between skin and object, the fatigue-state of the muscles, etc. Any one of these parameters is practically impossible for the system to know. The method of using the amplitude of the slip signal for estimation of an appropriate response seems as an intuitively correct method: the faster the slip, the stronger the reaction. However, how to set the relation between slip signal amplitude and response is still not known and was done here by trial and error. Also, there may be a dilemma as a smooth surface has been shown to give a weak slip signal, but may require a stronger response.

A problem which is encountered when slip compensation is to be used for functional tasks, is the fact that cutaneous receptors respond not only to slips, but also to increasing and decreasing forces applied perpendicularly on the skin [14], [15], skin stretches produced by movements of the fingers without any contact with an object [16], and stretches produced by changing load forces on an object in contact with the skin [35]. No present method can distinguish between these events based on the signal from this type of cuff electrode, and it is unlikely that a robust method for this can be found without additional information. The only way to deal with this at the moment is to make use of the slip information in the stable periods of the grasp, i.e., when no manipulation or strong movements are going on. However, we have done some preliminary tests of the subject lifting, moving and replacing an object without giving serious problems to the algorithm (not shown here).

Based on the experiments described in the present paper and the general experience with this system, we propose that slip information will be useful in the following two tasks: 1) for simple compensation of unexpected slips during more or less static hold phases and 2) for determination of the optimal stimulation intensity in the initial phase of a grasp-and-lift task. The latter can be made so that slips occur if the initial force is not strong enough to hold the object. This can trigger an increase in the stimulation intensity. If the initial force is too strong, the system automatically decreases the stimulation intensity until a slip occurs, and the minimally necessary stimulation intensity is then known. This may further be used to implement an on/off control of a hand neuroprosthesis so that the user needs only to ask for "grasp" or "release," instead of having to guess the correct level of stimulation and set this by means of wrist, shoulder, or ramp control (see [44] for a discussion of different control methods). This may make it easier for the user to control the device as no precise tuning of the stimulation intensity has to be performed when lifting different objects. When a task requires varying levels of stimulation intensity, e.g., eating with a fork, it might be possible to set a fixed level of stimulation intensity, enough to hold the fork alone. When the fork is either stabbed into some food, or is put into the mouth, a stronger force is needed. This can be obtained by letting the system increase the force strongly when slips occur and perhaps after one second with no slips, decrease back to the previous level.

Adding feedback from natural sensors to an FES system for handgrasp could potentially be very beneficial. However, in a clinical system it is still a question if it is worth causing a slip and risk loosing the object to get the information about the minimum stimulation intensity. In the data shown in the present paper, the stimulation was repeatedly adjusted during a whole trial, which could cause the object to slip several times over a period of 20 s. This is likely not acceptable in a functional system and was only included here to show the features of the method. Instead we suggest to adjust the stimulation intensity during the pick-up phase as described above and then only one time after picking up an object. After that the intensity will be held at a constant level until either the object slips unexpectedly (due to fatigue or external disturbances) or until the user commands the stimulator to release the object. Exactly how to do these things and whether or not they are beneficial, can only be tested with a person who uses an FES handgrasp system for everyday activities.

ACKNOWLEDGMENT

The authors wish to thank the subject involved in this study for his hard work and patience.

REFERENCES

[1] N. Hoshimiya, A. Naito, M. Yajima, and Y. Handa, "A multichannel FES 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.,* vol. 36, pp. 754–760, July 1989.

[2] R. H. Nathan, "Control strategies in FNS systems for the upper extremities," *Crit. Rev. Biomed. Eng.,* vol. 21, no. 6, pp. 485–568, 1993.

[3] B. Smith, P. H. Peckham, M. W. Keith, and D. D. Roscoe, "An externally powered, multichannel, implantable stimulator for versatile control of paralyzed muscle," *IEEE Trans. Biomed. Eng.,* vol. 34, pp. 499–508, July 1987.

[4] R. R. Riso, A. R. Ignagni, and M. W. Keith, "Cognitive feedback for use with FES upper extremity neuroprostheses," *IEEE Trans. Biomed. Eng.,* vol. 38, pp. 29–38, Jan. 1991.

[5] P. E. Crago, R. J. Nakai, and H. J. Chizeck, "Feedback regulation of hand grasp opening and contact force during stimulation of paralyzed muscle," *IEEE Trans. Biomed. Eng.,* vol. 38, pp. 17–28, Jan. 1991.

[6] P. E. Crago, H. J. Chizeck, M. R. Neuman, and F. T. Hambrecht, "Sensors for use with functional neuromuscular stimulation," *IEEE Trans. Biomed. Eng.,* vol. BME-33, pp. 256–268, Jan. 1986.

[7] J. G. Webster, "Artificial sensors suitable for closed-loop control of FNS," in *Neural Prostheses: Replacing Motor Function After Disease or Disability,* R. B. Stein, P. H. Peckham, and D. J. Popovic, Eds. New York: Oxford University Press, 1992, pp. 88–98.

[8] R. Masuda, K. Hasegawa, and K. Osako, "Slip sensor of industrial robot and its application," *Elec. Eng. Japan,* vol. 96, no. 10, pp. 219–226, Oct. 1976.

[9] A. Mingrino, P. Dario, and A. Sabatini, "A hand prosthesis with slippage control by tactile sensors," in *Proc. 15th Int. Meeting IEEE Eng. Med. Biol. Soc.,* 1993, pp. 1276–1277.

[10] R. D. Howe and M. R. Cutkosky, "Touch sensing for robotic manipulation and recognition," in *Robotics Review 2,* O. Khatib *et al.,* Eds. Cambridge, MA: M.I.T. Press, 1992, pp. 55–112.

[11] R. S. Johansson and G. Westling, "Roles of glabrous skin receptors and sensorimotor memory in automatic control of precision grip when lifting rougher or more slippery objects," *Exp. Brain Res.,* vol. 56, pp. 550–564, 1984.

[12] R. S. Johansson and Å. B. Vallbo, "Tactile sensibility in the human hand: Relative and absolute densities of four types of mechanoreceptive units in glabrous skin," *J. Physiol.,* pp. 283–300, 1979.

[13] Å. B. Vallbo and K.-E. Hagbarth, "Activity from skin mechanoreceptors recorded percutaneously in awake human subjects," *Exp. Neurol.,* vol. 21, pp. 270–289, 1968.

[14] M. Knibestöl, "Stimulus-response functions of rapidly adapting mechanoreceptors in the human glabrous skin area," *J. Physiol.,* vol. 232, pp. 427–452, 1973.

[15] ——, "Stimulus-response functions of slowly adapting mechanoreceptors in the human glabrous skin area," *J. Physiol.,* 245, pp. 63–80, 1975.

[16] M. Hulliger, E. Nordh, A.-E. Thelin, and Å. B. Vallbo, "The responses of afferent fibers from the glabrous skin of the hand during voluntary finger movements in man," *J. Physiol.,* vol. 291, pp. 233–249, 1979.

[17] Å. B. Vallbo and R. S. Johansson, "Properties of cutaneous mechanoreceptors in the human hand related to touch sensation," *Human Neurobiol.,* vol. 3, pp. 3–14, 1984.

[18] R. S. Johansson and G. Westling, "Signals in tactile afferents from the fingers eliciting adaptive motor responses during precision grip," *Exp. Brain Res.,* vol. 66, pp. 141–154, 1987.

[19] G. Westling and R. S. Johansson, "Responses in glabrous skin mechanoreceptors during precision grip in humans," *Exp. Brain Res.,* vol. 66, pp. 128–140, 1987.

[20] V. G. Macefield, C. Häger-Ross, and R. S. Johansson, "Control of grip force during restraint of an object held between finger and thumb: Responses of cutaneous afferents from the digits," *Exp. Brain Res.,* vol. 108, pp. 155–171, 1996.

[21] M. K. Haugland and J. A. Hoffer, "Slip information obtained from the cutaneous electroneurogram: Application in closed-loop control of functional electrical stimulation," *IEEE Trans. Rehab. Eng.,* vol. 2, pp. 29–36. 1994.

[22] ——, "Sensory nerve signals recorded by implanted cuff electrodes during functional electrical stimulation of nearby muscles," *IEEE Trans. Rehab. Eng.,* vol. 2, pp. 37–40, 1994.

[23] M. K. Haugland and T. Sinkjær, "Cutaneous whole nerve recordings used for correction of footdrop in hemiplegic man," *IEEE Trans. Rehab. Eng.,* vol. 3, pp. 307–317, Dec. 1995.

[24] J. A. Hoffer, G. E. Loeb, and C. A. Pratt, "Single unit conduction velocities from averaged nerve cuff electrode records in freely moving cats," *J. Neurosci. Methods,* vol. 4, pp. 211–225, 1981.

[25] J. A. Hoffer, "Techniques to study spinal-cord, peripheral nerve, and muscle activity in freely moving animals," *Neuromethods, Neurophysiologic. Techniques: Applications to Neural Syst.,* vol. 15, pp. 65–145, 1990.

[26] T. E. Milner, C. Dugas, N. Picard, and A. M. Smith, "Cutaneous afferent activity in the median nerve during grasping in the primate," *Brain Res.,* vol. 548, pp. 228–241, 1991.

[27] R. B. Stein, D. Charles, L. Davis, J. Jhamandas, A. Mannard, and T. R. Nichols, "Principles underlying new methods for chronic neural recording," *Le Journal Canadien des Sciences Neurologiques,* pp. 235–244, Aug. 1975.

[28] B. Upshaw and T. Sinkjær, "Digital signal processing algorithms for the detection of afferent nerve activity recorded from cuff electrodes," *IEEE Trans. Rehab. Eng.,* vol. 6, pp. 172–181, June 1998.

[29] M. K. Haugland, A. Lickel, R. Riso, M. M. Adamczyk, M. Keith, I. L. Jensen, J. Haase, and T. Sinkjær, "Restoration of lateral hand grasp using natural sensors," *Artificial Organs,* vol. 21, no. 3, Mar. 1997.

[30] M. K. Haugland and T. Sinkjær, "Control with natural sensors," in *Synthesis of Posture and Movement in Neural Prostheses,* J. Winters and P. Crago, Eds., to be published.

[31] A. Lickel, M. K. Haugland, J. Haase, and T. Sinkjær, "Restoration of lateral hand grasp using natural sensors," in *Neuroprosthetics: From Basic Research to Clinical Applications.* New York: Springer, 1996, pp. 327–343.

[32] K. Yoshida and K. Horch, "Closed-loop control of ankle position using muscle afferent feedback with functional neuromuscular stimulation," *IEEE Trans. Biomed. Eng.,* vol. 43, pp. 167–176, Feb. 1996.

[33] D. Popovic, R. B. Stein, K. Jovanovic, R. Dai, A. Kostov, and W. W. Armstrong, "Sensory nerve recording for closed-loop control to restore motor functions," *IEEE Trans. Biomed. Eng.,* vol. 40, pp. 1024–1031, Oct. 1993.

[34] N. Hoshimiya, M. Takahashi, Y. Handa, and G. Sato, "Basic studies on electrophrenic respiration—Part 1: Electrophrenic respirator synchronised with phrenic nerve impulses," *Med. Biologic. Eng.,* pp. 387–393, July 1976.

[35] R. R. Riso, P. J. Slot, M. K. Haugland, and T. Sinkjær, "Characterization of cutaneous nerve responses for control of neuromotor prostheses," in *Proc. 5th Vienna Int. Workshop Functional Electrostimulation,* Aug. 1995, pp. 335–338.

[36] A. Lickel, "Restoration of lateral hand grasp in a tetraplegic patient applying natural sensory feedback," Ph.D. dissertation, Aalborg Univ., 1998, pp. 1–98.

[37] P. J. Slot, P. Selmar, A. Rasmussen, and T. Sinkjær, "Effect of chronically implanted nerve cuff electrodes on the electrophysiological properties of human sensory nerves," *Artificial Organs,* vol. 21, no. 3, Mar. 1997.

[38] Y. Handa, N. Hoshimiya, Y. Iguchi, and T. Oda, "Development of percutaneous intramuscular electrode for multichannel FES system," *IEEE Trans. Biomed. Eng.,* vol. 36, pp. 705–710, July 1989.

[39] A. Scheiner, G. Polando, and E. B. Marsolais, "Design and clinical application of a double helix electrode for functional electrical stimulation," *IEEE Trans. Biomed. Eng.,* vol. 41, pp. 425–431, May 1994.

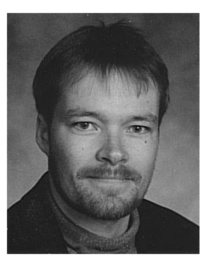
[40] R. B. Stein, T. Gordon, J. Jefferson, A. Sharfenberger, J. J. Yang, J. Totosy, D. Zepetnek, and M. Belanger, "Optimal stimulation of paralyzed muscle after human spinal cord injury," *Amer. Physiologic. Soc.,* pp. 1393–140, 1992.

[41] L. L. Baker, D. R. McNeal, L. A. Benton, B. R. Bowman, and R. L. Waters, *Neuromuscular Electrical Stimulation—A Practical Guide,* 3rd ed. Los Amigos Research & Education Institute Inc., CA, 1993.

[42] K. Kilgore, P. H. Peckham, G. B. Thrope, M. W. Keith, and K. A. Gallaher-Stone, "Synthesis of hand grasp using functional neuromuscular stimulation," *IEEE Trans. Biomed. Eng.,* vol. 36,pp. 761–770, July 1989.

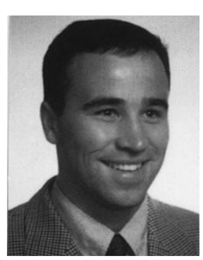
[43] F. Buchthal and A. Rosenfalck, "Evoked action potentials and conduction velocity in human sensory nerves," *Brain Res.,* vol. 3, no. 1, pp. 1–122, 1966.

[44] J. C. Burelbach and P. E. Crago, "Instrumented assessment of FNS hand control during specific manipulation tasks," *IEEE Trans. Rehab. Eng.,* vol. 2, pp. 165–176, Sept. 1994.



Morten Haugland was born on June 1, 1964, in Holstebro, Denmark. He received the M.Sc.E.E. degree in 1989 and the Ph.D. degree in biomedical engineering in 1994, both from Aalborg University, Denmark.

From 1989 to 1990, he was a Research Assistant in J. A. Hoffer's laboratory at the University of Calgary, Canada. He is presently employed as an Associate Professor at the Center for Sensory-Motor Interaction at Aalborg University. His main interests are within design and application of implantable electrodes and electronics for restoration of motor and sensory function of paralyzed people.



Andreas Lickel was born in Voitsberg, Austria, in 1967. He received the M.Sc. degree in telematics and biomedical engineering from the Technical University, Graz, Austria, in 1993. In 1998, he received the Ph.D. degree in electrical and electronic engineering at Aalborg University, Denmark.

Currently, he is a Postdoctoral fellow affiliated with the University College London, London, U.K., on a research grant issued by the European Union. His research is focused on the development of electrical stimulation devices to be used for restoring motoric function in paraplegic and tetraplegic patients.

Jens Haase was born in Copenhagen, Denmark, on May 10, 1941. He received the M.D. degree from the University of Copenhagen in 1966.

Since 1974, he has been a specialist in neurosurgery. From 1976 to 1982, he was an Associate Professor and the Head of the Department of Surgery at Odense University Hospital, Denmark. Since 1982, he has been Head of the Department of Neurosurgery at Aalborg University Hospital, Denmark. He was Chairman of the Danish Neurosurgical Organization from 1978 to 1984, President of the Scandinavian Neurosurgical Society from 1989 to 1992, a founding member of the Danish Microsurgical Society and the Chairman from 1994 to 1997. He has been Chairman of the European Association of Neurological Surgeons, European Board Examination Committee, since 1995 and a member of the training committee. Since 1998, he has been a Professor at SMI, Aalborg University. He is a member of the EANS, NNF, AANS, CNS, ISPN, ESPN, SNS, and several other neurosurgical societies. His research interests include implantable systems for restoration of function of paralyzed people and computer aided surgery by means of robotics and 3-D imaging techniques.

Thomas Sinkjær received the M.Sc.E.E. from Aalborg University, Denmark, in 1983 with a specialization in biomedical engineering. In 1988, he received the Ph.D. degree as a joint-venture project between the University of Calgary, Canada, and Aalborg University. He received the Dr.Med. degree in January 1998 from the University of Copenhagen, Denmark. From 1984 to 1986, he studied at the Department of Clinical Neurosciences, University of Calgary, Alta., Canada, and from 1989 to 1990 at the Department of Physiology, Northwestern University, Chicago, IL.

Since 1986, he has been with the Department of Medical Informatics and Image Analysis, beginning as an Associate Professor, then in 1992 as Head of the Department, and most recently as a Professor and Head of the Center for Sensory-Motor Interaction, which was established in 1993 under the Department of Medical Informatics and Image Analysis. He holds a wide number of offices, including member of “The Board of the Danish Research Councils” appointed by the Danish Minister of Research; reviewer for the Italian Research Council, as an expert for the EC (DG XIII) Programme Biomed I; and a referee for various international journals. His research interests include electrophysiology, biomechanics, motor control (human sensory-motor interaction), and neural prostheses.