Slip Information Provided by Nerve Cuff Signals: Application in Closed-Loop Control of Functional Electrical Stimulation

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***Abstract*— A model of a paralyzed hand gripping and lifting an object was developed using anaesthetized cats. Functional neuromuscular stimulation (FNS) applied to the ankle plantarflexor muscles caused the footpad to press against and grip an object. Electroneurographic activity (ENG) activity generated by skin mechanoreceptors in the footpad was recorded with a cuff electrode implanted on the tibial nerve. Sharp bursts evident in the ENG signaled any slips between the object and the skin. This information was used in an event-driven controller that allowed the FNS system to compensate for slips. In this way an "artificial gripping reflex" was implemented that compensated automatically for internal changes (fatigue) and the external perturbations (increased load, changed frictional coefficient). This control scheme proved to be robust and is proposed to be applicable for restoration of precision grip in paralyzed humans.**

I. Introduction

THE human fingertips are highly capable of detecting whether a gripped object is slipping. Surface features protruding a few micrometers from the surface of an object can be discriminated when stroked along the surface of the skin [20]. "Microslips" can be detected that involve only a small part of the skin area in contact with an object held in a precision grip [16]. This capability originates from the large number of low-threshold mechanoreceptors in the skin of the fingertips (as many as 241 units/cm^2 [14]. The information from these receptors is essential for the control of precision grip; if the skin is anaesthetized, the subject becomes incapable of adequately adjusting grip force to the weight and surface structure of an object [15].

During precision tasks, normal subjects usually produce grip forces barely greater than the minimum force required to hold an object [16], which depends on both the weight of the object and the frictional properties of the surface in contact with the fingers. If the grip force becomes insufficient and the object starts to slip, low-threshold mechanoreceptors typically respond with sharp bursts of activity. Slips may happen if the grip force decreases as a consequence of changing joint angles or fatigue; if the weight of the object increases (e.g., as a drinking cup gets filled) or if the frictional coefficient decreases (e.g., caused by excessive perspiration). It has been shown that a short-latency spinal reflex of cutaneous origin is usually elicited by such a slip, so that within 80 ms of the start of a slip, the grip force increases and the object is held securely again [16]. This rapid corrective response is automatic, and does not involve conscious participation by the subjects. The reflex action is quite powerful, to the extent that subjects often cannot voluntarily release their grip slowly to let an object fall, because the reflex tends to interfere [16].

The cumulative activity of sensory receptors can be recorded with a chronically implanted nerve cuff electrode placed around a peripheral nerve (reviewed by Hoffer [8]). This approach makes it possible to use cutaneous sensors as a source for feedback information to a system for functional neuromuscular stimulation (FNS). The relation between the electroneurographic (ENG) signal obtained from the cat tibial nerve and the force applied perpendicularly on the footpad innervated by the nerve has been investigated, but its properties make it difficult to use the ENG signal for estimation of the perpendicular skin contact force [6]. However, the sensitivity of cutaneous mechanoreceptors to slips suggests that the ENG signal may contain information about slips across the skin that could be very useful for the control of precision grip in humans. If reliable slip information were available, an FNS system implementing an "artificial gripping reflex" could be able to make a paralyzed hand hold an object with only the necessary force and maintain a good grip even if the muscles fatigue, if skin-to-object friction decreases, or if the weight of the object increases. An "artificial gripping reflex" has previously been implemented for the control of prosthetic hands, where the incipient slip of an object being held by an amputee in a prosthetic hand was measured and used to control the force of prehension [2].

In this paper we present an approach to extracting slip-related information from the neural signal. An algorithm is presented that implements an artificial gripping reflex initiated by slips during precision grip, and this approach is tested in an animal model.

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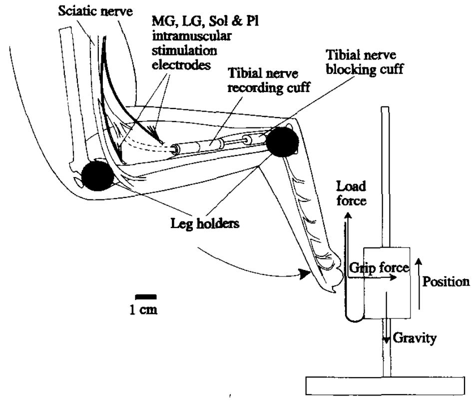

Fig. 1. Diagram of electrodes implanted in cat hindlimb and apparatus used to measure neural responses during grip. The cat was under anaesthesia. The limb was fixed at the ankle malleoli and knee with atraumatic, cupped clamps. Forces were produced by FNS via electrodes implanted in the four ankle plantarflexor muscles (MG, LG, Sol, and Pl; not all are shown). ENG activity in the tibial nerve was recorded with a tripolar cuff electrode. When the plantarflexor muscles were stimulated the paw moved (curved arrow) until the footpad pressed against an object. The surface contacted by the footpad was covered with fine sandpaper. The object could slide vertically along a low-friction bearing and contained two force transducers to measure vertical (load) force and horizontal (grip) force, as well as a linear potentiometer to measure vertical position.

Preliminary results of this work have been published by Hoffer *et al.* [9]–[10], Sinkjær *et al.* [18], and Hoffer and Haugland [11].

II. Methods

Our experimental model of a paralyzed human hand gripping an object was the foot of an anaesthetized cat pressing the central footpad against an object that was constrained to move vertically with low friction (Fig. 1). The central footpad of the cat is a reasonable model of human glabrous skin, as was discussed by Haugland and Hoffer [6].

Three adult cats (4–6 kg) were chronically implanted under gas anaesthesia (all surgical details are described in Haugland *et al.* [6]; cats 7, 9, and 10 from that series were also used for the present study), received Atravet and Morphine subcutaneously for the first 24 h after surgery to reduce postoperative pain, and were maintained in accordance to the guidelines of the National Institutes of Health (USA) and the Canadian Council for Animal Care.

Bipolar intramuscular stimulation electrodes were implanted in each of four ankle extensor muscles: medial and lateral gastrocnemius (MG, LG), soleus (Sol), and plantaris (Pl). The electrodes consisted of two Teflon coated 40-strand stainless steel wires (Cooner 634), with the ends deinsulated for 15 mm, and were inserted in the muscle diagonally to the fibers about 2 cm apart in the proximal part of the muscle (see Fig. 1). Cross stimulation among the different muscles in the intact leg was reduced by using bipolar stimulation electrodes, which was a more selective method than using monopolar electrodes with respect to a common ground.

A tripolar nerve recording cuff (30 mm long, 2.2 mm inside diameter) was implanted on the tibial nerve, distal to the muscular branches, and 4–5 cm above the ankle. At this level, the tibial nerve contains mainly afferent fibers, mostly from the plantar surface of the foot and the footpads [3], and the cuff could be implanted without obstructing the blood supply to the nerve or causing mechanical damage to the nerve.

Recording sessions were performed approximately once every 10 days. At the beginning of each recording session, the cat was anaesthetized with an intravenous injection of Thiopentothal (8–10 mg/kg) through an implanted jugular vein catheter, intubated, and maintained under anaesthesia with Halothane in a mixture of nitrous oxide and oxygen.

To remove sensory afferent contributions from hair receptors in the skin surrounding the central footpad, the foot was shaved, treated with depilatory cream, thoroughly washed, and moisturized with cream applied to the skin, before each experiment.

During the slip experiments the cat was supported by a heated sling and the implanted hindlimb was fixed by two pairs of cupped clamps pressed around the ankle and knee joints (Fig. 1). The leg temperature was kept at about 34°C with a heat lamp. The holder allowed the ankle joint to move and did not damage the skin. The foot hung vertically when not stimulated. When the ankle extensor muscles were stimulated, the footpad pushed horizontally against a test object that could slide vertically along two bars with low friction and that would fall if not gripped by the foot. In analogy to precision grip studies by Westling and Johansson [21], the object was equipped with force sensors that measured horizontal grip and vertical load force by means of strain gauges (Revere, FT50). The vertical position of the object was measured with a linear potentiometer (Waters, LRT-S-100B) (Fig. 1). As the object weighed 140 g, gravity caused a constant downward pull of about 1.4 N. The surface of the object was 600-grit sandpaper.

The stimulator we used for FNS [7] produced rectangular monophasic constant current pulses with a fixed amplitude for each channel. For each of the four muscles, pulsewidths were independently controlled between 0 and 255 μs in steps of 1 μs by an IBM PC/AT compatible 80386 microcomputer via a parallel port. Each muscle was stimulated at a fixed frequency of 25 Hz, i.e., with interpulse intervals of 40 ms. To reduce force ripple caused by unfused tetani, the four muscles were stimulated sequentially (or interleaved), so that one of the muscles was stimulated every 10 ms, giving an aggregate stimulation frequency of 100 Hz. The stimulation current was adjusted for each muscle to the lowest value that produced maximum force at maximum pulsewidth. This insured that the widest possible range of pulsewidths was available for modulation of muscle force. No attempt was made to linearize the intrinsic current-to-force recruitment curves for each of the muscles.

The ENG signal recorded from the tibial nerve was bandpass filtered (1–10 kHz), rectified, and bin-integrated (into a single 3 ms bin every 10 ms) before sampling. This procedure was used to reduce the pickup of artifacts [7] and also to sample at a lower frequency (100 Hz) than would be necessary to sample the raw ENG (20 kHz). Bin integration and sampling times were synchronized to the delivery of stimulation pulses by the computer. Thus, the ENG envelope was sampled at the same frequency as the aggregate stimulation rate (100 Hz). In the following the term "ENG" refers to the envelope of the ENG rather than to the raw ENG, since for feedback purposes only the envelope had to be sampled.

III. Results

The slip-related information contained in the ENG signal was assessed in pilot experiments in which the four plantarflexor muscles were stimulated with trains of pulses of constant width to generate an approximately constant force. The pulsewidth was chosen so that the force was not quite sufficient for the foot to grip the object securely. The experimenter had to partially support the object and could thus allow the object to fall transiently by releasing its support in several small steps. Each step caused a slip between the paw and the object.

Data from one of these experiments are shown in Fig. 2. Initially the muscles were not stimulated (thus the grip force was approximately zero) and the object was held by the experimenter about 55 mm above the table, with the cat footpad just touching the bottom edge of the object. One second after start of recording, the stimulation was started and the grip force rose to between 4 and 5 N. At this time the ENG showed a sharp peak that resembled the peaks recorded when an external force was applied to the footpad [6]. About 1 s later, the experimenter released the object, which dropped a few mm and caused the footpad to deform by the action of the increased tangential force between skin and object (the load force). When the load force increased enough to overcome the static friction, the object slipped across the skin. The slip continued until the experimenter stopped the fall of the object. The change in skin deformation and the slip caused the ENG to increase sharply, to a value even higher than at force onset; thereafter it adapted again to the new steady state situation. The action of letting the object drop a few mm at a time was repeated several times, each time causing paw deformation and slip, accompanied by a sharp burst of ENG activity, until the stimulation was turned off (at 11 s). The footpad was considerably less deformed in the second and following drops than in the first one, since by then the footpad was already partially deformed.

Sharp bursts of ENG activity signaled slip occurrences in every experiment and for all cats used in this study. Slip-related bursts were distinct enough from the background ENG to be detected with great accuracy and very early in the slip phase. The vertical lines in Fig. 2 mark the moments when slips were detected by the computer from the sampled ENG data as described below. As can be seen, all slips were detected, and always soon after the object started to move.

At the start of each new FNS run, the increase in ENG activity that normally accompanied the force increase could falsely trigger the slip detection algorithm. However, since this event happened as a result of an increase in stimulation intensity instead of unexpected movement, the controller anticipated it and simply ignored its occurrence.

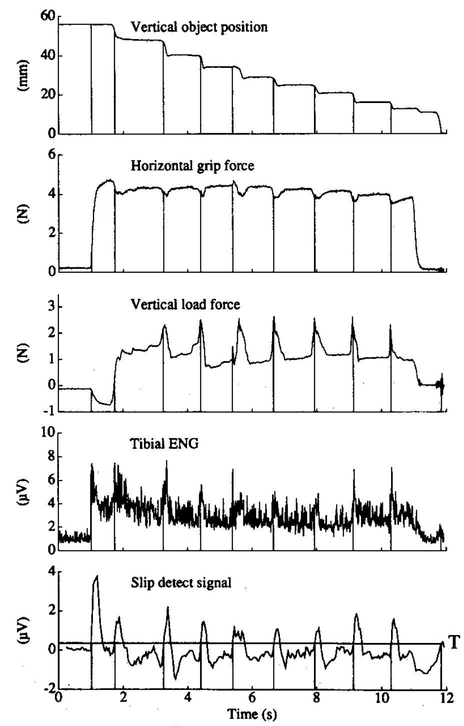

Fig. 2. Method of slip detection. The ankle plantarflexor muscles were stimulated at a constant intensity starting at 1 s, causing the paw to press against the object surface with a nearly constant force that in this case was intentionally lower than necessary to hold the object. The experimenter had to partially support the object and let it drop in steps of about 5 mm by removing support transiently. Each step caused a slip between the object and the skin on the central footpad. When a slip occurred, the ENG signal showed a phasic burst of activity that was detected by the computer from comparison of the differential and filtered ENG signal (lower trace) to a threshold value (T). In this and following figures, vertical lines indicate the times when the computer detected slips in this manner. Data from cat 7.

A. Detection of Slip

Since slips were accompanied by ENG bursts, they could be detected by differentiating the ENG signal and comparing it to a threshold value. However, simple differentiation, calculated as the difference between two consecutive ENG samples, proved too noisy. Instead, a "slip detection" signal was calculated by subtracting a low-pass filtered (time constant = 0.2 s) version of the ENG signal delayed by 20 samples (200 ms) from the "unfiltered" ENG, thereby removing the background activity from the ENG. The unfiltered ENG also needed some filtering to reduce noise, but this was done with a shorter time constant (0.07 s). Implementation of the detection algorithm was done in C (Turbo-C 2.0, Borland), and the filters were implemented as first-order autoregressive (AR) filters, which are computationally very simple. The set of time constants, delay and threshold value that gave the most sensitive and robust slip detection was found by trial and error, using data as shown in Fig. 2. Except for the threshold value, all the other parameters could be held invariant not only during one experimental day, but also from day to day and for different cats. The threshold parameter needed adjustment at the start of each experimental day, but could be increased by more than 50% of its optimal value without seriously affecting the detection of slips. Once set, the threshold value needed no further adjustment during the rest of that day.

The following algorithm was used to detect slip:

1) *repeat (this loop runs at 100 Hz)*
   a) *Update 20 samples of old ENG values*
   b) *Sample new ENG value*
   c) $BackgroundENG = BackgroundENG * a + OldENG * (1 - a)$
   d) $SlipENG = SlipENG * b + NewENG * (1 - b)$
   e) $SlipDetect = SlipENG - BackgroundENG$
   f) *if SlipDetect > Threshold then signal that a slip occurred*

2) *end*

The constants a and b were determined from the time constants discussed above (for $\tau = 0.2\,\mathrm{s} \rightarrow a = 0.951$ and $\tau = 0.07\,\mathrm{s} \rightarrow b = 0.867$).

B. Increase of Force After Slip

As soon as a slip was detected, the force was increased as fast as possible to reestablish a secure grip of the object before it moved out of reach. This was done by one of the following methods: 1) by increasing the pulsewidths markedly for a short period in order to recruit more motor units; or 2) by immediately stimulating each muscle with a "doublet:" a closely spaced pair of stimulation pulses (interpulse interval = 5 ms). This mechanism is used by the central nervous system in normal conditions [12]; after a doublet, the force not only increases rapidly, but can also remain elevated for a prolonged period afterward (the "catch" property [1], [22], [23]).

Once the grip was reestablished, it could be maintained securely by a constant-frequency pulse train if the pulsewidth was moderately longer than before the slip (typically 20% to 30% longer).

Because the ENG was sampled at 100 Hz, no more than 10 ms elapsed between the detection of a slip and an updated response from the controller. Because the ENG was sampled for 3 ms prior to each stimulus pulse, the speed of the response to a detected slip was determined only by the time it took to do the calculation (less than 1 ms). The main delay in slip detection was caused by the low-pass filtering of the ENG, which was necessary to remove false detections caused by normal variations in the ENG. The time constant for this filter was 70 ms, as described above. The delay from the moment the object first started to move until the slip was detected was typically between 50 to 100 ms, but could vary considerably (see below).

A doublet stimulation pulse caused additional stimulus artifact in the next ENG sample [7], and the sudden increase in force gave rise to an ENG burst [6]. Both these events could be misinterpreted as further slips. To avoid erroneous detections of slip during increased stimulation, the controller ignored all samples for 300 ms after a slip detection. In principle, if the force increase had not been sufficient during this "deaf period," the object could have dropped a considerable distance before

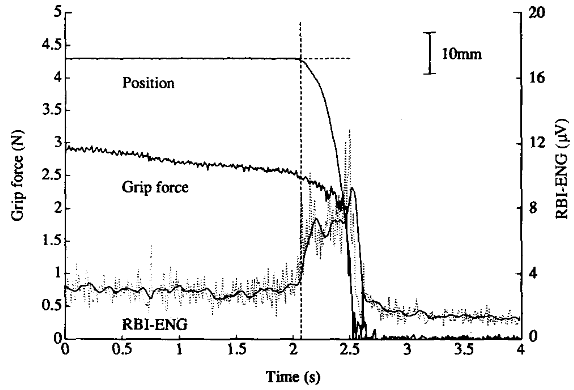

Fig. 3. Slip detection during declining grip force (simulated muscle fatigue), without subsequent compensation. Initial part of experiment as described in Fig. 2, except here the grip force was initially sufficient to hold the object. The stimulation intensity was gradually decreased. The decline in grip force caused the object to eventually slip. The slip was detected by the computer (dotted line), but in this example was ignored (the muscle stimulation intensity was not modified). The RBI-ENG (dotted trace) is shown as it was sampled, and the solid trace is the RBI-ENG low-pass filtered at 5 Hz. Data from cat 10.

a new compensatory increase in force could be implemented. In practice, if the pulsewidths were increased by 30% to 40%, no further slips were observed during the correction phase.

The stimulation algorithm included an automatic shut-off of stimulation if the object dropped out of range, triggered by the vertical position transducer signal.

C. Test of Closed-Loop Slip Compensation Controller

To test the slip compensation algorithm, experiments were made that included slips caused by both external (e.g., "increased load") and internal (e.g., "fatigue") changes. Examples of slip caused by fatigue-like declines in muscle force are shown in Figs. 3 and 4.

The initial part of the experiment was the same as in Fig. 2. The object was lifted to about 75 mm above the table, and stimulation started. The stimulation intensity was initially set to 40% of maximum, which, after the experimenter released the object, was high enough for the foot to hold the object about 68 mm above the table. Thereafter, the stimulation intensity was decreased at a slow constant rate. This caused the grip force to decrease in a nonlinear way that depended on the collective recruitment state for the four muscles, their lengths, and the extent of fatigue. Fig. 3 shows on an expanded time scale that when the grip force became too low and the object started to slip, the ENG responded with an increase in activity. This ENG burst was detected by the computer (vertical dotted line), but for the purpose of this figure no compensation in stimulation was implemented, and the object consequently continued to slide out of reach of the paw.

In Fig. 4 a slip occurred as in Fig. 3, but in this case the slip detection information available from the ENG was used to increase the stimulation intensity and the paw gripped the object again; i.e., the "artificial gripping reflex" described in the preceding section was activated. By the time the grip was

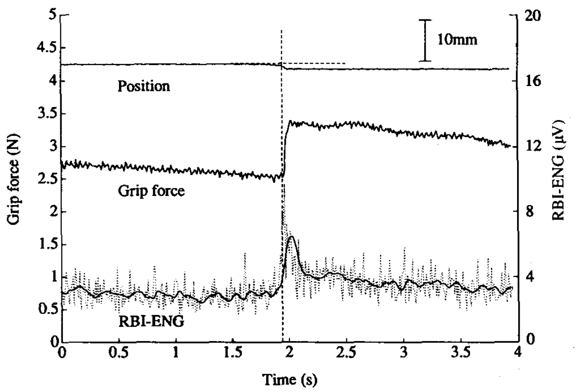

Fig. 4. Slip detection during declining grip force with subsequent compensation. Initial part as in Fig. 3, but when the object started to slip (dotted line), the grip force was rapidly increased by eliciting a doublet (see text) and subsequently increasing the pulsewidth by 30%, which was sufficient to grip the object securely again. The stimulation intensity was gradually decreased afterward to provoke further slips (not shown). The RBI-ENG (dotted trace) is shown as it was sampled, and the solid trace is the RBI-ENG low-pass filtered at 5 Hz. Data from cat 10.

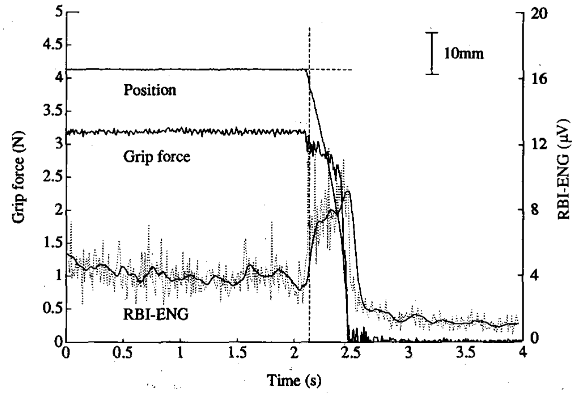

Fig. 5. Slip detection during an increase in external load, without subsequent compensation. Initial part as Figs. 3 and 4, except the stimulation intensity was constant. A 50-g load was suddenly dropped on the 140-g object, causing it to slip. The slip was detected by the computer (dotted line), but in this case was ignored (the stimulation intensity was not increased) and thus the object continued to fall. The RBI-ENG (dotted trace) is shown as it was sampled, and the solid trace is the RBI-ENG low pass filtered at 5 Hz. Data from cat 10.

reestablished the object had moved less than 2 mm. Once the object was held securely again, the intensity of stimulation was allowed to decrease at the same rate as before the slip.

For slips caused by fatigue-like decreases in grip force (e.g., Figs. 3 and 4), the time delay from the start of a slip to its detection was variable. The slip detector "reaction time" was typically 70 ms, but in some cases it was less than 40 ms and in other cases more than 100 ms. If the object started to move very slowly (e.g., Fig. 4) it was difficult to determine when the slip started; this was attributed in part to deformation of the footpad when the grip force decreased slowly.

Once a slip was detected, the time required to regrip the object securely was typically 50 ms, but appeared to depend on several factors such as the velocity of the object (which depended on the time since start of slip), its weight, the surface structure (static versus dynamic friction coefficients), the stimulation pattern (e.g., how much the pulsewidth was increased after slip detection) as well as the strength and contraction speed of the stimulated muscles. It was evident that the longer it took to detect a slip, the longer it took to stop it afterwards.

Examples of slip caused by a sudden increase in weight of the object are shown in Figs. 5 and 6. In these experiments, the object (weight = 140 g) was gripped by stimulating the muscles at a constant intensity. Two seconds into the run, a 50-g load was suddenly dropped onto the object. Fig. 5 shows a run without compensation for slips, and Fig. 6 shows a run with compensation. In Fig. 5, a slip was detected but the controller did not change the FNS pattern. In Fig. 6, the controller compensated for the slip in the same manner as described previously.

Dropping the 50-g load caused the slip to happen very suddenly, as can be seen by the abrupt change in the position trace in Figs. 5 and 6. The nerve signal reacted strongly; these slips were detected faster and more regularly than slips caused by declining grip force. Typically the delay from slip to detection was 30 to 40 ms. However, since the object was now heavier than before the slip, and moved faster than for slips caused by decreasing grip force, it took longer to catch it after the slip was detected. Typically this took 150–200 ms, but was very variable.

The robustness of this method for slip-compensation in the presence of external perturbations was investigated in multiple trials in which the slip information was alternatively used or not used. In the example of Fig. 7, 20 consecutive trials are shown, 10 with compensation and 10 without compensation. In all the trials where the slip information was ignored, the grip was lost when the extra 50-g load was added, whereas in every one of the trials where slip information was used to increase the stimulation intensity, the object and its added weight were stopped in their fall and held again in a secure grip until the end of the run. The typical distance the object moved before it was caught was less than 4 mm. It should be noted that the compensation for slips in the "fatigue" series of experiments was similarly robust; in those experiments the object typically moved less than 3 mm before it was caught (data not shown).

In two of the trials in Fig. 7, an additional slip was detected after the object and its added weight were gripped, causing the stimulation intensity to be further increased even though the object did not move detectably. These ENG bursts may have been the result of localized slips on the part of a footpad, as was described by Johansson and Westling [16] for the human fingertip.

The robustness of this approach to slip compensation was further challenged in experiments that included both external and internal perturbations. In the example of Fig. 8, the object was lifted 85 mm above the table. Stimulation was started at a, with the intensity initially set to 40% of maximum. This was sufficient to grip the object securely with the paw when the object was released at b. Thereafter, the intensity of stimulation decreased at a constant slow rate. When at c the grip force became too low, the object started to slip. Detection of the slip triggered the controller to increase the intensity (as detailed in legend). The grip was reestablished after the object

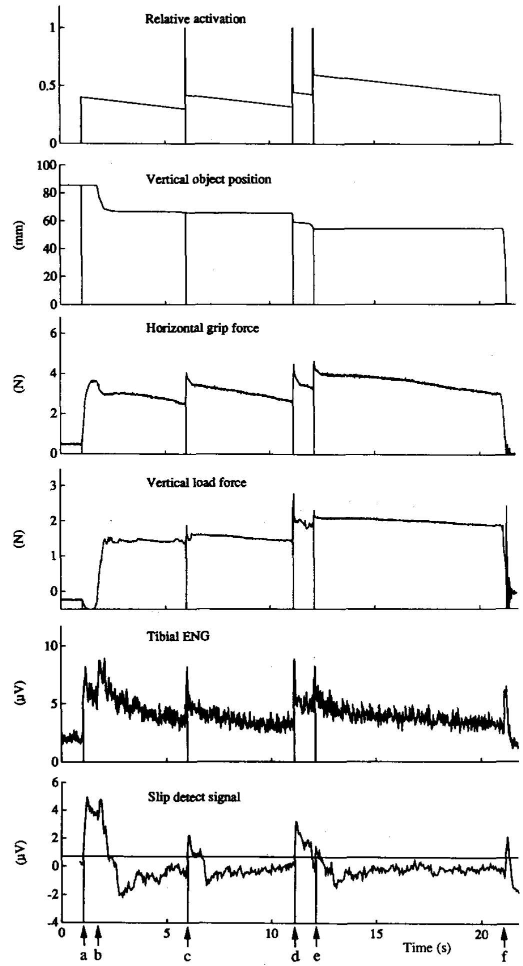

Fig. 6. Slip detection during an increase in external load, with subsequent compensation. Initial part of the run was as in Fig. 5, but as soon as a slip was detected by the controller (dotted line) a doublet pulse was elicited and the pulse width of subsequent stimuli was increased by 40%, which stopped the fall and held the object securely afterward. The RBI-ENG (dotted trace) is shown as it was sampled, and the solid trace is the RBI-ENG low pass filtered at 5 Hz. Data from cat 10.

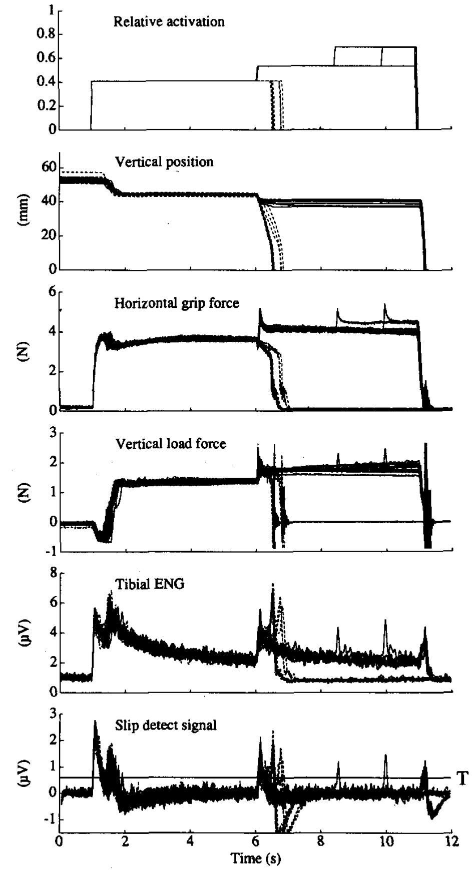

Fig. 7. Comparison of stimulator performance with and without slip compensation. The foot held the object with constant stimulation intensity, until a 50-g extra load was dropped on the object. In alternating runs, the controller was in open-loop and closed-loop modes. In 10 trials with open-loop FNS (i.e., no slip compensation; dashed lines), the extra weight caused the object to fall out of reach of the foot. In each of 10 interleaved trials with closed-loop control (i.e., with slip compensation; solid lines) the object fell less than 4 mm before the grip was secured again. Data from cat 7.

moved about 1 mm. Thereafter, the intensity of stimulation again decreased, at the same rate as before. At *d*, when the weight of the object was suddenly increased by 50 g, a new slip occurred. The controller detected the activity burst in the ENG signal, responded accordingly, and the slip was stopped. However, the new grip force was barely sufficient to hold the object because of its increased weight, and a new slip occurred shortly afterward *e*. This was also detected and compensated for. The grip force was now sufficient to hold the object securely, even though the stimulation intensity continued to decrease during the next 9 s, until the experiment ended at *f*.

The "artificial grip reflex" algorithm was not only designed to compensate for slips caused by either internal or external changes; it also had the capability to measure the minimum stimulation intensity that was required to grip the object in place. As can be seen in Fig. 8, the 140-g object could not be gripped securely if the intensity was below about 30% of maximal. After the extra 50-g weight was added, the 190-g object could not be gripped securely if the intensity was below about 43% of maximal. This means that muscle fatigue could be minimized by fixing the stimulation intensity to a level only a little higher than it was before a slip occurred. If, for any reason, a higher intensity became necessary, a slip occurred and the intensity was automatically increased. On the other hand, in some circumstances a lower intensity of stimulation could suffice (e.g., if the weight of the object had decreased). By slowly and regularly decreasing the intensity of stimulation until a slip occurred, the controller was designed to probe systematically the lower limit for safe grip and to operate within a reasonable safety margin.

IV. Discussion

In this paper we have demonstrated, in an animal model of human precision grip, that signals recorded with a cuff electrode implanted on a cutaneous nerve provide reliable feedback that can be used to control a functional neuromuscu-

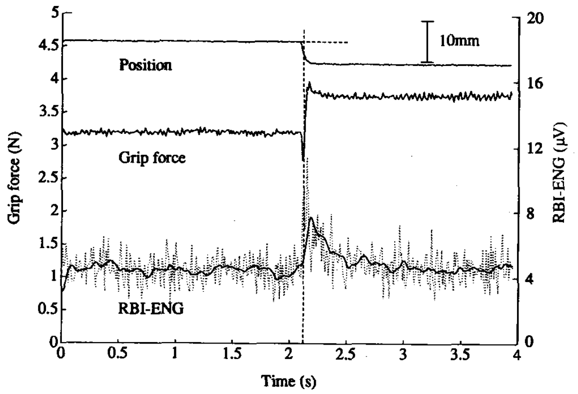

Fig. 8. Slip compensation during both simulated "fatigue" and increased load. Top trace indicates relative muscle activation with FNS. When slips were detected at times c, d, and e, a doublet (interpulse interval $= 5$ ms) of activation pulses with maximum pulsewidth (255 μs) was sent once to each of the four muscles to mediate a rapid increase in force. The stimulation intensity was subsequently increased by 40% with respect to the level just prior to the detection of each slip. Further description in text. Data from cat 10.

lar stimulation system that adjusts grip force automatically in response to slips. We believe that this approach can be applied equally well to implement an "artificial gripping reflex" in neurologically impaired humans. It could be used, for example, for the restoration of precision grip in spinal cord injured patients (e.g., C4–C6 quadriplegics [13], [19]).

An important concern when evaluating the clinical applicability of a control system is the range of circumstances for which it will operate reliably. Our experimental model system was equally reliable at detecting slips from the recorded ENG signal when the cause for the slip was a decrease in grip force, or an increase in load force. In either situation, the grip was fully regained by issuing a single doublet of stimuli to each of four muscles, combined with a 30%–40% increase in the duration of subsequent pulses. The time from when the object first started to move, to slip detection, was typically in the range of 40 to 100 ms. For slips caused by decreasing force, the object typically moved less than 3 mm before it was gripped again. For slips caused by a sudden 35% increase in load, the object moved less than 4 mm before it was gripped again. The method proved to be equally robust in all three cats used in this study.

Another concern when evaluating the clinical applicability of a control system, especially if it is to be used by severely disabled persons, is the simplicity of its operation. Only one parameter in the slip detection algorithm, the threshold value, needed adjustment, and it was only adjusted once, at the start of each day of experiment. No further changes on any parameters were required for the rest of the recording session, which usually lasted 4–6 h. In the clinical setting, it would be easy to provide an automatic self-determination of threshold mode whenever the unit is turned on.

The increment in the stimulation intensity after slip, set to 30%–40% in the present experiments, was necessary and sufficent to secure the grip in the situations we tested. In a clinical FNS system, the value of the this factor will depend not only on the instantaneous recruitment curve for the stimulated muscle(s), but also on the reason for slips, e.g., how much the weight of objects changes, how strong are the perturbations encountered, and a number of other factors such as the fragility of gripped objects. It is anticipated that this value may be easily adjustable by the user to match appropriately the properties of his/her muscles, temperament, and FNS system.

A further important concern regarding FNS systems is minimization of muscle fatigue. We have shown in this study that, in addition to enabling the FNS system to compensate for declining grip force or external perturbations, slip information can also be used for continuous determination of the optimal stimulation intensity and thus avoid causing muscle fatigue from using unnecessarily strong stimulation. The control scheme (shown in Figs. 4 and 8) of slowly decreasing stimulation intensities, combined with slip detection and compensation, automatically determines the minimum necessary intensity to grip a given object. This scheme may be used repeatedly during an otherwise "static hold phase" to make adjustments to slowly varying load changes, or it may just be used initially to determine a suitable intensity that is then held constant unless other slips occur.

An advantage of implementing an FNS system that used feedback information derived from natural sensors is that it is likely to replicate natural performance between than other approaches. Slip information from skin receptors is known to be essential for the regulation of force during precision grip in the intact human. Visual cues such as size, material, and surface structure play a role in deciding the initial forces when gripping and lifting an object, but for the steady-state holding phase, the stomatosensory information from the skin effectively defines the necessary force with no apparent relation to prelift visual cues [4]. The results reported in the present paper suggests that an FNS system for precision grip in paralyzed humans could use the same cutaneous afferent information as the intact organism and implement a similar control strategy, thereby adjusting the grip force automatically to the weight and surface structure of the object in hand in the form of an "artificial gripping reflex."

This approach appears suitable for restoration of function of paralyzed upper as well as lower limbs. In upper limb applications, tactile information arising from the thumb, index, and/or middle fingers could be recorded with nerve cuff electrodes either from the palmar cutaneous branch of the median nerve proximal to the wrist, or from individual internal digital nerve branches in the hand [6], [11]. Experiments with a nerve cuff implanted on the median nerve of a monkey [17] have produced ENG signals during perturbations of precision grip that closely resemble those presented in the present paper, suggesting that the median nerve may be a suitable source of the desired signal. An initial clinical implementation in lower limb for correction of foot-drop in a hemiplegic patient, based on aspects of the present work, was reported recently by Sinkjær *et al.* [18] and Haugland *et al.* [5].

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