Skin Contact Forces Extracted From Human Nerve Signals—A Possible Feedback Signal for FES-Aided Control of Standing

Lotte N. S. Andreasen* and Johannes J. Struijk

***Abstract*—In this paper, information about stance related skin contact forces was extracted from nerve cuff electrode recordings of human neural signals. Forces measured under the heel during standing were scaled and applied to the innervation area of the sural nerve on the side of the foot using a hand held force probe. The neural response to the stimuli was measured with a cuff chronically implanted around the sural nerve in one hemiplegic person. An artificial neural network was used for extraction of the applied force from the recorded nerve signal. The results showed that it is possible to extract information about absolute skin contact forces from the nerve signal with an average goodness of fit of 69.3% for all trials and 82.2% for the more dynamic trials. This information may be applicable as feedback signal in control of standing.**

***Index Terms*—Author, please supply your own keywords or send a blank e-mail to keywords@ieee.org to receive a list of suggested keywords..**

I. Introduction

THE success of functional electrical stimulation (FES) in the rehabilitation of spinal cord injured individuals depends on the ability of the FES system to adapt to external changes and disturbances. For example, the stimulation of the leg muscles in FES-supported standing should be adjusted to the momentaneous balance of the standing person in order to compensate for postural instability, arm motions and postural perturbations, to prevent the person from falling. This adaptation to environmental changes can be obtained by the use of feedback signals in the FES system, which update the system with the state of the controlled motion, such as the ankle angle or ankle momentum in standing [1], the slip of a held object in hand grasp [2], or the occurrence of foot lift in drop foot stimulation [3], [4].

Neural signals from cutaneous sensors in man contain information that is applicable as such feedback [2]–[5], and has been used as feedback for restoration of hand grasp and in drop foot stimulation. The cutaneous sensors provide information about pressure, skin stretch and vibration [6]. Studies of standing in intact subjects have shown that pressure information from cutaneous sensors in the foot sole is used by the central nervous system to stabilise the posture [7], [8]. These sensors, thus, provide information about the balance of the standing person. This motivates extraction of neural information from natural sensors in the foot sole to provide feed back in FES-assisted control of standing.

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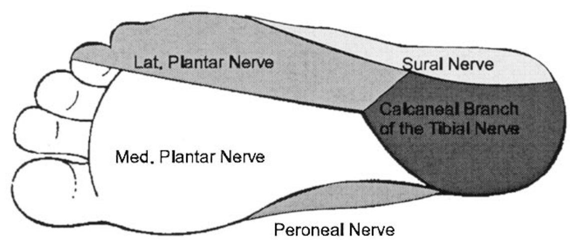

Fig. 1. The three main innervating nerves of the foot sole Adapted form [14].

Previous use of natural sensory feedback in man has been based on event detection such as in detection of the slip of a held object. For control of standing, more quantitative features such as the center of pressure (COP) under the foot sole, is desirable [9]. The low S/N ratio in the recordings of the electroneurogram (ENG) [10] and the rather complex input-output relation of the natural sensors [11] limits the obtainable accuracy of the extracted features, but recent control schemes [12] suggest that the COP does not need to be known very accurately to be useful as a feed back signal.

The innervating nerves of the human foot sole splits it into three main areas (Fig. 1). These areas are conveniently located, such that neural information from the different nerves or nerve branches would provide pressure information from the lateral, medial, anterior and posterior part of the foot, which in combination, could give information about the COP [13], [14].

The usefulness of the separate innervation areas for the purpose of feedback in FES is conditional on the feasibility of extracting pressure/force related information from the cutaneous sensors. To study this feasibility, we simulated skin contact forces as imposed on the cutaneous sensors in the foot sole during standing and applied these forces to the innervation area of the human sural nerve (Fig. 2) in order to extract the applied force from the resulting ENG, using an artificial neural network (ANN).

II. Methods

The overall methodology used in this study is shown in Fig. 2. Skin contact forces were measured from the calcaneal innervation area in an intact person during standing. These forces were

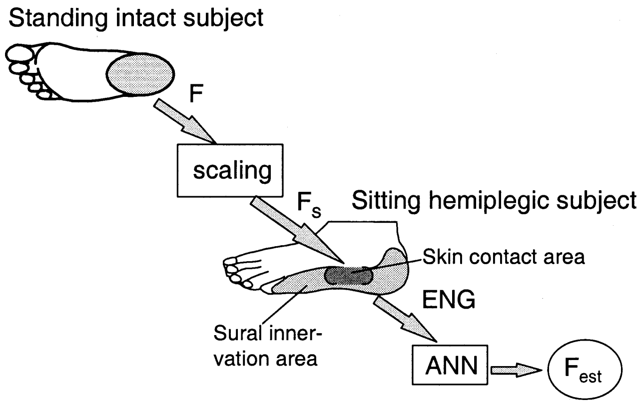

Fig. 2. Skin contact forces are measured during standing from the calcaneal innervation area, and scaled so be applied to the sural innervation area with a hand held force probe. The resulting ENG response is recorded and processed to be input to an ANN which estimates the applied force from the ENG.

scaled, and applied to the sural innervation area of a hemiplegic person with a chronically implanted nerve cuff electrode for correction of foot drop as described in [3]. The resulting ENG response was measured with the cuff electrode and processed to be used as an input to an ANN, which estimates the applied force. The reason that two different persons were subjects in the experiments is that the hemiplegic person was unable to perform the required sway around the ankle joint during standing, due to fatigue and balance problems.

A. Measurement of Skin Contact Forces During Standing

One 31-year-old intact female was standing and performing sway around the ankle joint, while she was keeping the knee and hip joints as fixed as possible in extended positions. She was performing sways of different frequencies and excursion amplitudes in the sagittal plane (Table I). The feet were placed close together and part of the right foot (resembling the calcaneal innervation area, Fig. 1) was resting on a force plate allowing skin contact forces to be measured.

At the beginning of the experiments the standing subject was asked to classify the range of the excursion amplitude into large and small sways and to classify the speed of the movement as very fast, fast, medium, slow, and very slow.

Each trial lasted for 20 s and was repeated three times. The perpendicular skin contact force measured with the force plate was sampled at 200 Hz.

B. Simulation of Skin Contact Forces During Standing

1) Scaling of Force: The skin contact area of the hand held force probe used to apply forces to the sural innervation area was smaller than the area of the foot on the force plate. Therefore, the vertical force measured with the force plate during standing was scaled according to the ratio between the area of the foot sole on the force plate (34.5 cm^2) and the contact area of the force probe (20.75 cm^2). Then the force was further reduced (as shown in the results) to avoid unpleasant stimulation when the forces later were applied to the sural innervation area with the hand held force probe.

2) Application of Forces With Force Probe: To create a visual display that the holder of the force probe could use in order to track the scaled forces in time and amplitude, and thereby simulate skin contact forces during standing, the scaled forces were calibrated to the force probe and plotted on transparencies which were later mounted on the display of an oscilloscope, as in [16], to which the force probe was connected. The time division of the scope was adjusted until the light beam became a dot traversing the display with a speed according to the time course of the force plots on the transparencies. Only 10-s sequences were used (due to the lower limit of the time/div. of the oscilloscope) and for each measurement these were tracked twice giving a total trial duration of 20 s.

While the force was applied to the sural nerve, the holder of the force probe paid attention to keep the probe perpendicular to the innervation area of the sural nerve, while supporting the subject's foot with the other hand (Fig. 3). The force probe data was sampled at 400 Hz, low-pass filtered at 6 Hz, and down-sampled to 25 samples/s.

C. Recording of ENG Signals

The ENG response to the applied force on the sural innervation area was measured with a nerve cuff electrode which was chronically implanted around the sural nerve of a 57-year-old woman, who suffered from a drop foot due to a stroke ten years earlier and had volunteered to have the cuff implanted for use in drop foot stimulation research. The cuff had been implanted for six weeks at the time of these experiments. It was a silicone cuff fabricated as described in [17] and closed with inter digitating tubes [18]. The cuff was 30 mm long and had an inner diameter of 3.4 mm. The three inner cuff electrodes were 2 mm wide and made of 25-μm platinum foil. The lead wires were coiled stainless-steel wires (AS 634 Cooner wire C0.) and were 500 mm long. The cuff itself was implanted around the sural nerve just proximal to the malleolus and the lead wires were running subcutaneously from the cuff up to a location 25 cm proximal to the malleolus, where they penetrated the skin.

During the measurements the woman was sitting in a comfortable chair and was encouraged to relax and allowed to sleep. The experiment lasted for about 4 h, including a 30-min lunch break.

The ENG signal from the cuff was recorded using a tripolar configuration and was bandpass filtered at 10 Hz–10 kHz (fourth-order) and sampled at 20 kHz. Each trial lasted for 20 s and was repeated three times. After the measurements, only trials with little electromyogram (EMG) interference was chosen to be used as input to the ANN.

1) The Artificial Neural Network:

a) Preprocessing: Before being used as input to the ANN, the recorded ENG signal was Fourier-transformed and the frequency content below 700 Hz and above 1500 Hz was removed, since very little of neural signal was located outside this frequency band. After inverse Fourier-transform, the signal was rectified and the envelope of the ENG was obtained by low-pass filtering with several cutoff frequencies. The goodness of fit (GOF) parameter (see below) between the ENG envelopes and the applied force was used to choose two cutoff frequencies for the ENG envelopes. These two envelopes were down-sampled to 25 samples/s and simultaneously used

TABLE I
THE DIFFERENT SWAYS USED IN THE TRIALS AND THE ANN RESULTS, COLUMN 2 AND 3 REPRESENT TRAINING TRIALS, AND COLUMN 4 AND 5 REPRESENT TEST TRIALS

Trial type	training data	ANN results GOF [%]	test data	ANN results GOF [%]	Frequency [Hz] and amplitude [N] of the trial types
Large sway very slow	lsvs1	68.3	lsvs2	57.5	f=0.1Hz, amp.=10-60N
Large sway fast	lsf1	76.3	lsf3	84.3	f=0.35Hz, amp.=10-55N
	lsf2	82.0			
Large sway very fast	lsvf1	85.1	lsvf2	74.8	f=0.45Hz, amp.=10-60N
Small sway slow			sss1	53.7	f=0.2Hz, amp.=20-50N
Small sway very slow	ssvs1	65.5	ssvs3	65.8	f=0.18Hz, amp.=20-50N
	ssvs2	55.5			
small sway very fast	ssvf1	88.4	ssvf2	89.3	f=0.65Hz, amp.=15-50N
			ssvf3	81.3	
small sway fast			ssf1	84.5	f=0.60Hz, amp.=10-55N
quiet standing	qs1	13.0			amp.= 30-35N
	qs2	12.0			
	qs3	21.0			
combined1	cmb1	65.5	cmb2	45.6	amp.=15-60N
combined2	cmb3	80.5	cmb4	68.4	amp.=15-60N
step	stp1	75.0	stp2	59.8	amp.=15-55N

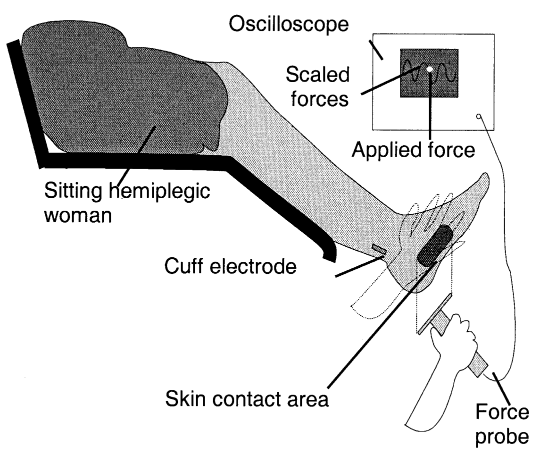

Fig. 3. The scaled forces are applied to the sural innervation area by the experimenter, who supports the foot during the application. A visual display for tracking the scaled forces is created by mounting transparencies with the calibrated force on the display of an Oscilloscope, and adjusting the time division to the trial duration.

as input for the ANN. The applied force was used as output for the ANN during the training of the ANN.

b) The ANN Structure: The ANN was implemented as a three-layer feed forward network [19] with two input variables and one output (using the Matlab neural network toolbox). There were 12 hyperbolic tangent sigmoid neurons in the first and second layer and one linear neuron in the third layer. To cope with the hysteresis in the force modulation of the ENG [11], the network was given a memory (160 ms) by using the four previous samples as inputs as well, so five input vectors were simultaneously applied for each of the two ENG-envelopes used as input variables.

The network was trained with the Levenberg–Marquardt backpropagation algorithm on a training set created by 13 trials of 20 s, and tested on data that was not part of the training set. The network was trained on 1000 Epochs.

The output of the ANN, $\hat{F}$ was compared with the applied force, F, using a GOF parameter defined by the squared correlation coefficient

$$\mathrm{GOF} = \frac{\left[\sum_i (F_i - \overline{F})(\hat{F}_i - \overline{\hat{F}})\right]^2}{\left[\sum_i (F_i - \overline{F})^2\right]\left[\sum_i (\hat{F}_i - \overline{\hat{F}})^2\right]} \tag{1}$$

which gives a direct measure of how much of the total variance can be explained by the straight line model (the optimal mapping).

III. RESULTS

1) The Applied Forces: An example of the measured force during standing and its scaled version applied to the sural innervation area with the hand held force probe is shown in Fig. 4. The forces have similar patterns and their first derivatives are also similar. The amplitude of the force measured during standing was 6.7 times higher than the force applied with the force probe, and the peak values of the derivative of the force was 5–10 times higher than for the hand held force probe.

a) The Measured ENG: The power density plots of the cuff signal recorded during the trials were studied in order to select trials for the ANN. Only trials with low EMG interference as shown in Fig. 5 were used as input to the ANN. The neural information was present in the frequency band 500 Hz–2 kHz, with a maximum around 1 kHz (Fig. 5). The trials used for extraction of force information are listed in Table I.

b) The Preprocessed Data: Preprocessing the ENG, as described in the methods, resulted in two envelope low-pass cutoff filter frequencies of 0.65 Hz (Fig. 7) and 2 Hz, which were used simultaneously as input to the ANN. The frequency of 0.65

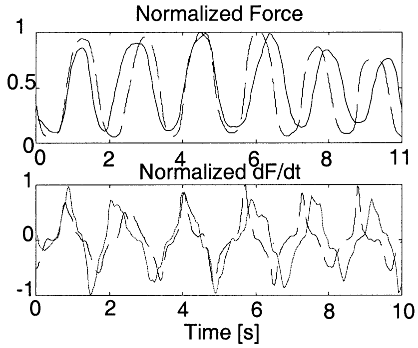

Fig. 4. Example of the force measured during standing (dashed) and the force applied to the sural innervation area with the force probe (solid), for the trial ssf1 (Table I). Top: the forces, bottom the time derivatives of the force.

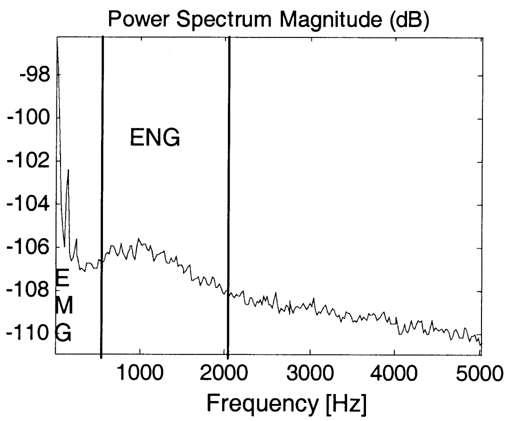

Fig. 5 Power density plot of the cuff signal recorded in the trial ssf1. The frequency band classified as ENG is shown between the vertical lines. Only trials with low power in the EMG frequency band as shown here were used as input for the ANN.

Hz was found by calculating the GOF (1) between the applied forces and the ENG envelope (Fig. 6) for frequencies of 0.25 Hz–4 Hz for all the training data (13 trials). This gave different optimal frequencies for all the trials ranging between 0.25 and 2.5 with standard deviation of 0.64. Removing the two largest and two smallest optimal frequencies reduced the standard deviation to 0.35 and the mean frequency was 0.66 Hz. Therefore, the cutoff frequency of the envelope filter was chosen to be 0.65 Hz and in addition a higher frequency envelope with a cutoff of 2 Hz was used to provide information of more sudden force changes.

The ANN performed better when these envelopes were used simultaneously as input. An example of the two ANN inputs are shown in Fig. 7 (top). The high-frequency envelope shows that the ENG peaks when the applied force [Fig. 7 (bottom)] increases and again when it decreases, indicating a correlation between the derivative of the force and the ENG.

This is further illustrated in Fig. 8 (top), where the applied force, its derivative and the high-frequency envelope of the ENG are plotted together for the same trial as shown in Fig. 7. There is a clear correlation between the ENG and the absolute value of the derivative of the force, suggesting velocity sensitivity of the natural sensors. Further, there is a tendency of the ENG to respond stronger to an increase in the force than to a decrease, which is also described by [11] for ENG from cats.

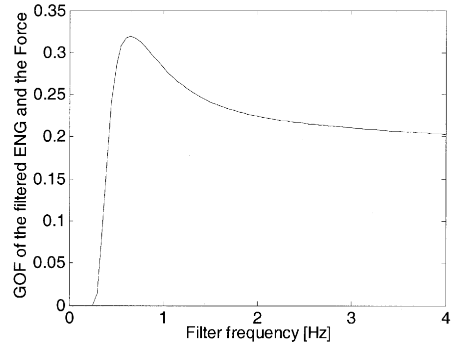

Fig. 6. Example of GOF values between the applied force and the ENG envelopes when different cutoff frequencies were used for the envelope low-pass filter. The trial used in this example is lsvf1 (Table I).

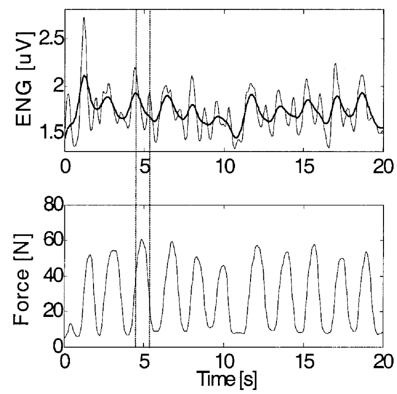

Fig. 7. Example of input-output signals for the ANN for the trial ssf1. Top: The two ENG envelopes used as input to the ANN, bold: the 0.65 Hz envelope, solid: the 2-Hz envelope. Bottom, the related applied force.

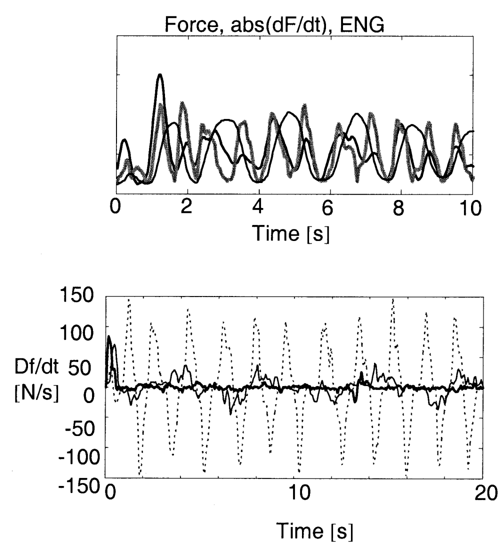

Fig. 8. Top: The applied force (solid), the absolute, scaled value of its derivative (grey) and the 1-Hz envelope of the ENG (bold). The ENG and the force derivative shows a clear correlation (trial:ssf1). Bottom: the force derivatives of the three different trials, ssf1 (dashed), ssvs (solid) and qs3 (bold). There is a big difference in the amplitude of the dF/dt for the different trials.

As shown in Fig. 8 (bottom), there is a big difference in the amplitude of the dF/dt for the different trials, and therefore, also in the neural activity. For trial qs3 (Table I) the dF/dt is close to 100 times smaller than the maximal values for trial ssf1.

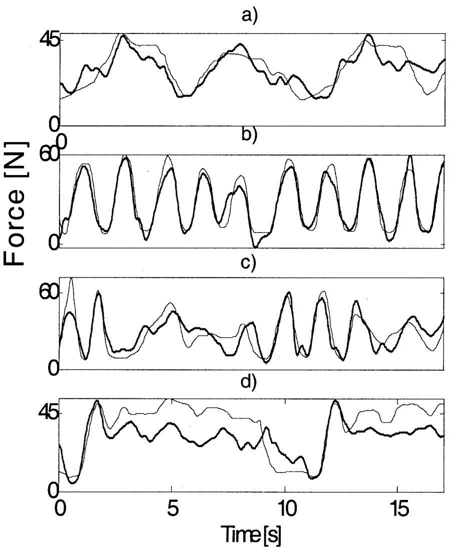

Fig. 9. ANN results on test data. Bold: the estimated force using the ANN, solid the actual applied force. (a) The results form the trial: ssvs3, with a $\mathrm{GOF} = 65.8$, (b) The results form the trial: ssf1, with a $\mathrm{GOF} = 84.5$%, (c) The results form the trial:cmb4, with a $\mathrm{GOF} = 68.4$%, and (d) The results form the trial: stp2, with a $\mathrm{GOF} = 59.8$%.

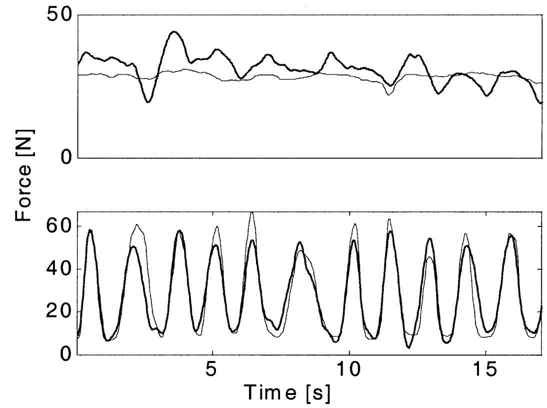

Fig. 10. ANN results on training data. Bold: the estimated force using the ANN, solid the actual applied force. Top: The results form the trial: qs3, with a $\mathrm{GOF} = 21.0$. Bottom: The results form the trial: ssvf1, with a $\mathrm{GOF} = 88.4$.

2) ANN Results: The ANN was trained with one input file containing all 13 training trials. These 13 trials are listed in column 2 in Table I, and the test trials (not used for training) are listed in column 4. In addition, Table I shows the resulting GOF for both training (column 3) and test (column 5) trials obtained from a comparison of the applied force with the estimated force.

Comparing the results on training data and test data indicates that the ANN did not overfit the training set.

The results on the test data give GOF parameters of 45.6% to 89.3% with an average of 69.5% (Table I). There is a clear tendency for the high-frequency trials to give better results. The average GOF of the sway-trials is 76.8% and if only trials with fast sway are included in the calculations, the average GOF is 82.2%.

In Fig. 9, the extracted force is shown together with the applied force for four different test trials with GOF in the range 59.8%–84.5%. The applied and extracted forces are comparable, but again the best results are obtained for the faster movements.

Even though the best results are obtained for faster movements, Fig. 10 (top) showing the ANN results on training data, illustrates that the offset level around 30–35 N for the more static trial qs3 can be approximated by the ANN, indicating that static information about the force is also available in the ENG.

IV. Discussion

The forces applied to the sural innervation area had patterns that were comparable with the skin contact forces measured from the calcaneal innervation area during standing. The amplitudes and the derivatives of the forces, applied with the hand held force probe, were in the order of 5–10 times smaller than the ones measured during standing, because of the smaller area of the force probe and a further reduction to avoid discomfort at high force levels. Since the recorded ENG showed a clear correlation with the force derivative, this difference in amplitude and derivative might also have caused a relatively reduced ENG response to the applied force as compared to the one that could be expected during standing. Another reason for the sural ENG response to be less than could be expected from the skin of the foot sole during standing is that part of the stimulated skin area was nonglabrous skin, which has lower densities of mechanoreceptors as compared with the glabrous skin of the sural innervation area, which extends 1–1.5 cm into the sole of the foot [20].

The most prominent information in the ENG had a dynamic character as described in [11] and the extraction of force from the ENG by the ANN gave the highest GOF values for the fastest movements which in average gave a GOF of 82.8% (five trials), whereas the average GOF for all test trials was 69.5%. Nevertheless, a base level of the static forces, such as in the trial quiet standing, could still be estimated. This could partly be due to the fact that even quiet standing is not completely static and partly because of static information in the ENG. The static information in the recordings can be explained by a higher number of slowly adapting pressure sensors being active, due to a larger size of the force probe contact area, than has been described in other studies, with less static information in the recordings [21]. ENG responses similar to the ones found in this work, with static properties, have previously been measured from the human sural ENG [16]. Furthermore, a study by Vallbo and Hagbarth [22] shows that the firing rate of the slowly adapting receptors increases with the applied force. This is also shown for the human foot by Vedel and Roll [23], who found that the slowly adapting units in the foot were not sensitive to light scratching and only sometimes to pinching, while the most efficient stimulus was perpendicularly applied pressure. The tonic firing rate was found by averaging the nerve response in 20 to 50s after the stimulus onset, suggesting a very slow adaptation, and thereby a quite reliable static pressure sensor [23].

The demanded resolution for foot pressure as a feed back parameter suggested by [24] is 23 N, which indicates that the inaccuracies should be well below that value. The errors in the force extracted by the ANN were seldom that high (Fig. 9), even for the trials with the lowest GOF. The question is whether the error will increase accordingly if the applied force is increased.

Such an increase will probably not be linear with the applied force because the ENG activity is expected to increase with increased force, which is expected to improve the performance of the ANN. Moreover, the ANN structure used in this work was a relatively simple feed forward network with simple preprocessing. More advanced techniques are likely to improve the performance of the ANN. Nevertheless, a study by Matjacic and Bajd indicates that pressure information for feedback in FES need not to be available very accurately [12].

The results of the present work suggest that the cutaneous ENG, resulting from skin contact forces during standing, can provide pressure information for use as feedback in FES-assisted standing.

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