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2.5.2: FES II - Control Strategies

NATURAL SENSORY FEEDBACK FOR CONTROL OF STANDING

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***Abstract*-- The use of natural sensory signals as feedback for FES systems within rehabilitation is still in its early stage. So far the signals have mostly been used for event detection, e.g. detection of slip [1]. A new application of sensory feedback signals is to apply it in control of paraplegic standing. This paper describes a preliminary study to the use of natural sensory feedback in control of FES assisted standing. It was investigated whether it seems possible to extract a qualitative feature suitable for control of standing from an ENG signal recorded by a cuff electrode. The ENG from the foot sole of one normal subject, standing with voluntary sway, was modeled and used for extraction of the center of pressure (COP) within the foot support area. An artificial neural network was used for feature extraction and the extracted COP was compared with the actual COP measured by a forceplate. The result showed that the modeled ENG signal does provide information about the position of COP.**

I. INTRODUCTION

The advantages of using natural sensory signals for feedback in FES are many and have been pointed out by many researchers. Besides eliminating the technical problems, such as mounting problems, drift, lack of reliability, energy consumption etc., they might also help to eliminate the psychological problems due to the cosmetic disadvantages of the artificial sensors. But all those advantages are not for free. Also the natural sensors have their drawbacks, e.g. fast adaptation and hysteresis in the case of the cutaneous sensors [2]. These properties of the cutaneous sensors could be expected to degrade their usefulness for feedback in static situations, such as in quiet standing. Thus when a person has been standing for a while the receptors in the foot sole might adapt in such a way that no information about the center of pressure (COP) can be extracted from the ENG recorded by a cuff electrode. Nevertheless, a standing person will continuously perform small sways and thereby introduce movements of the COP. When COP moves, the activity of the receptors in the three major innervation areas in the foot sole (Figure 1) will change and by combining the ENG signals from the three nerves, it might be possible to extract information about COP.

The main hypotheses were: 1) The changes in COP during standing affects the ENG sufficiently for extraction of the COP. 2) Combining the three ENG signals it is possible to compensate for ambiguities due to hysteresis.

The main assumptions were: 1) It is possible to make selective recordings from the three major nerves. 2) The ENG was modeled using a model from Haugland et al.[2,3]. The model was developed for cuff electrode recordings from the sural nerve in a hemiplegic subject. It was assumed that the model also gave a good estimate of the activity of the three major nerves. 3) It was chosen to consider the changes of the COP in the sagittal plane only, which is assumed to be the most important for stabilizing a standing person.

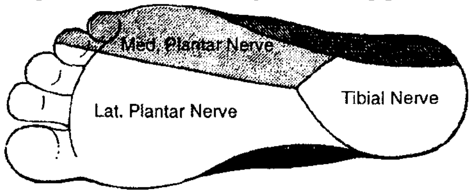

Figure 1. The three major innervating nerves of the foot sole. Adapted from Gray's Anatomy, 36th Edition.

II. METHODS

The hip and knee joints of a normal male subject were fixed by an orthosis to simulate an inverted pendulum which is assumed to be an appropriate model for FES assisted paraplegic standing [4]. Six force sensitive resistors (FSR) were placed on a forceplate with two FSRs within each of the three innervation areas of the foot sole. The subject had bare feet, one foot was placed on the resistors/forceplate and the other foot on the floor close by. The subject performed different kinds of voluntary sway: fast, slow, small, large and quiet standing and a combination. The duration of the different sway trials were 30 s. and about 60 s. for the combined sway. Each trial was performed twice. The forceplate signals and the FSR signals were sampled with 50 Hz and low pass filtered by a 3rd order Butterworth, 5 Hz cutoff frequency. The measured COP was calculated from the forceplate data using the method described in the AMTI forceplate data sheet. The six FSR signals were calibrated and scaled according to the difference between the area of the FSR and of the force probe used in the Force-ENG model [2,3]. Each of the six signals was used as input for the Force-ENG model. To obtain three ENG signals related to each of the three major innervation areas under the foot sole, the six outputs of the model were summed, two and two, and scaled again according to the difference between the area of the force probe used in the Force-ENG model and the skin contact area within each innnervation area. The three modeled ENG signals were low pass filtered with a 2nd order Butterworth, 3 Hz cut off frequency, to reduce the spikes in the signal (Figure 2). The cut off frequency at 3 Hz was chosen since a FFT analysis of the COP from the forceplates showed that the bandwidth was about 3 Hz. The three filtered signals were used as input to a three layer feedforward artificial neural network (ANN). The ANN was trained with the Levenberg-Marquardt back propagation learning rule. The output training set was the COP coordinate in the sagittal plane, measured by forceplates. The ANN performed a direct mapping of the ENG to the COP and did not include any time information.

III. RESULTS

Figure 2 shows the modeled ENG from the foot sole. The model output was a rectified and smoothed ENG [3].

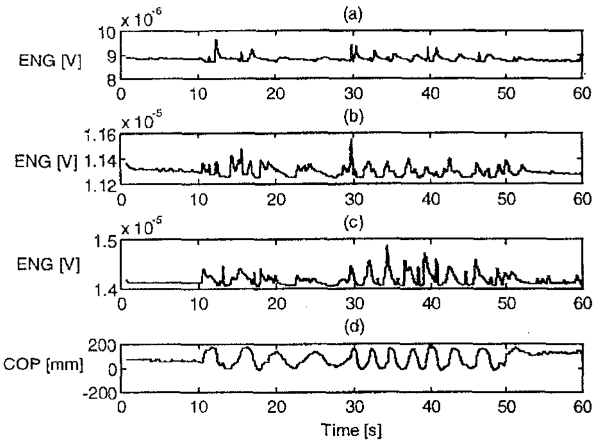

Figure 2. Example of the modeled ENG. (a): At tibialis, (b) The Lateral plantar nerve, (c) The medial plantar nerve. (d) The measured COP, combined sway.

In Figure 3 and 4, the results are shown for an ANN with 12 nodes in the first and second layer and 1 node in the output layer. The ANN was trained on the trial with combined sway. The output of the ANN was smoothed by a 2nd order Butterworth, 1 Hz cut off. Other ANNs trained on the same type of trial showed a similar performance.

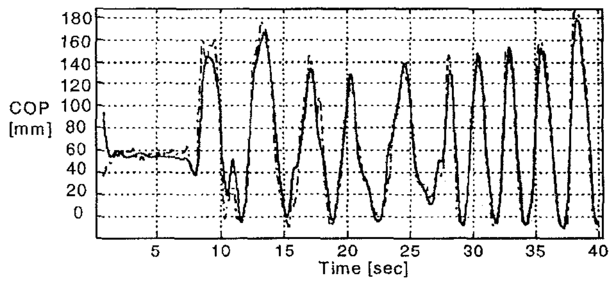

Figure 3. Solid: The output from a the ANN on training data, combined sway, Dashed: Measured COP relative to the ankle joint.

The trajectory of the COP is recognizable even though there are deviations especially in the static parts of the test data (Figure 4). There seem to be no significant signs of errors due to hysteresis; the curve seem equally well fitted in forward and backward sways.

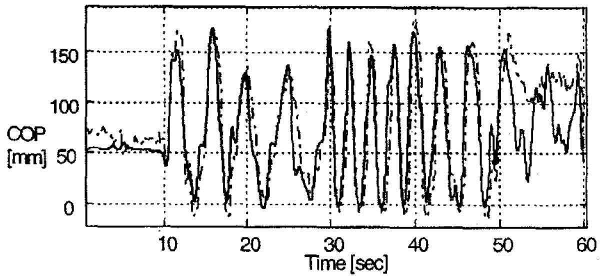

Figure 4. Solid: Output of the ANN on test data, combined trial Punctured: the measured COP relative to the ankle joint.

IV. DISCUSSION

The large errors of COP in the static parts of the trials are expected to be a result of too little static information in the training set. An ANN trained on static data gave good results on the static test trials. Further research should be put into optimizing the ANN and verifying the validity of the modeled ENG. It seems possible in future applications to extract information of the COP in both the sagittal and coronal plane. Including time information in the feature extraction might make information about velocities and accelerations of the COP achievable.

V. CONCLUSION

It was concluded that the modeled ENG does provide information about the COP in the sagittal plane, and that the combination of the three ENG signals seems to reduce the problems with hysteresis in the receptors.

ACKNOWLEDGMENT

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