SENSORY FEEDBACK FROM MOTORIZED PROSTHESES: THE INTERFERENCE BETWEEN AFFERENT ELECTRICAL NERVE STIMULATION AND PROSTHESES MYOELECTRIC CONTROL SYSTEMS

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Abstract

Afferent electrical nerve stimulation has good properties when applied as sensory feedback to motorized prostheses. This investigation deals with the interference between the electrical nerve stimulation and some different myoelectric pick-up systems. Experiments performed with different stimulation parameters show that the interference can be kept at a level low enough to permit the containment of both the stimulation system and the pick-up system within the prosthesis itself. This is considered to be important for the patient acceptance of the prosthesis.

Introduction

Reports of studies on sensory feedback systems for motorized prostheses have appeared in the literature during the last years. There is a general recognition of the importance of sensory feedback in order to improve the performance of the amputees when they are using their prostheses (e.g. Szeto et al., 1976; Szeto et al., 1977; Solomonow & Lyman, 1977; Shannon, 1976; Anani et al., 1977). Many authors have suggested that electrical stimulation possesses several advantages in relation to tactile stimulation for sensory feedback purposes (Shannon, 1976; Anani et al., 1977). As pointed out by Childress (1974), self-containment and self-suspension of a prosthetic system is a vital factor for the patients' acceptance of their prostheses. It is thus important that the addition of a feedback system to a prosthesis will not deteriorate this principle. In this context electrical stimulation has the advantage of very good miniaturization prospects and low energy consumption in contrast to mechanical tactile stimulation. On the other hand, the use of electrical stimulation close to the pick-up electrodes for myoelectric signals will inevitably cause an interference with the myoelectric control system. This interference is the only factor that might prevent the containment of the electrical feedback system within the prosthesis. The purpose of this investigation is to study the extent and the nature of the interference in order to establish whether or not electrical afferent nerve stimulation feedback in a myoelectric prosthesis can be consistent with the important concept of self-containment.

Methods

The experiments were carried out on a non-amputated male subject. The purely sensory superficial branch of the radial nerve was identified in the forearm six to eight centimeters proximal to the wrist joint by means of transcutaneous electrical stimulation. The nerve was then stimulated bipolarly by two fine Karma-needle electrodes, see Fig. 1, introduced to the nerve through the skin by means of hypodermic needles. The stimulating current was delivered from a constant current stimulator providing square pulses. The reference electrode was situated on the skin distal to the active stimulation electrodes. The stimulator allowed independent changes in amplitude and frequency of the stimulating pulse, which could be continuously monitored. The tested stimulation parameters included current intensities from 0.65 to 1.3 mA and frequencies from 10 to 80 Hz with a constant pulse duration of 100 µs. The selection of parameters was based on a study of afferent electrical nerve stimulation for sensory feedback purposes (Anani et al., 1977).

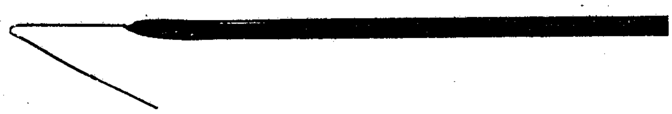

Figure 1. The Karma wire electrode in a hypodermic needle. 2.5 mm of the hook-shaped tip is deinsulated.

The interference voltage was recorded by means of active skin electrodes designed for picking up myoelectric signals in prosthetic systems. Three types of electrodes were used providing different signal processing. The first type of electrode provided pure amplification of the myoelectric signal. The second type tested was the commercially available Otto Bock electrode characterized by an on-off response mode. In the third type of electrode, the myoelectric signal was amplified, band-pass filtered, and rectified, and it provided an output DC voltage proportional to the intensity of the myoelectric signal (Almström, 1977).

The experimental set-up is illustrated in Figs. 2 and 3. The pick-up electrodes were connected to a cathode ray oscilloscope and to a digital voltmeter. The reference electrode for the pick-up system was placed on the skin at the elbow. Thus, the pick-up and stimulation systems had different, isolated references with the reference electrodes situated as far apart as possible (30 centimeters). In order to further reduce the interference, the two systems were supplied by two isolated voltage sources.

The interference of electrical stimulation with the EMG-electrode signals was studied at various stimulation parameters as a function of the distance between the stimulation electrodes and pick-up electrodes. For this purpose, the pick-up electrodes were moved along a distance scale painted on the skin of the subject, and the interference of each stimulation parameter was monitored at regular intervals.

In one experiment the interference was studied *in vitro*. The same stimulation and pick-up arrangements were used, but the stimulation was delivered to a thick sheet of porous paper soaked with a 0.9 per cent sodium chloride solution and placed on a supporting frame on the surface of a bowl with three litres of the same solution, see Fig. 4.

The active electrode providing the proportional output was used to pick up the signal, but the shape of the electrode surfaces was changed. Instead of circular surfaces, two narrow bars were used so that the distance between the stimulating electrodes and the pick-up electrodes could be more accurately measured.

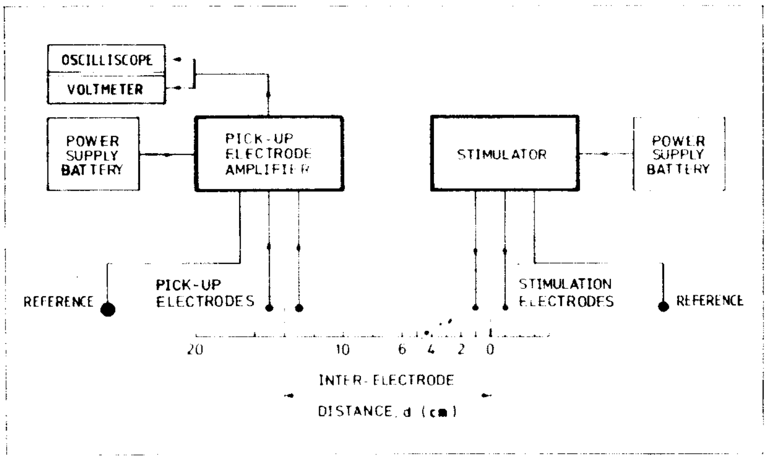

Figure 2. Experimental set-up. The block diagram is valid for both the experiments on the subject and *in vitro*.

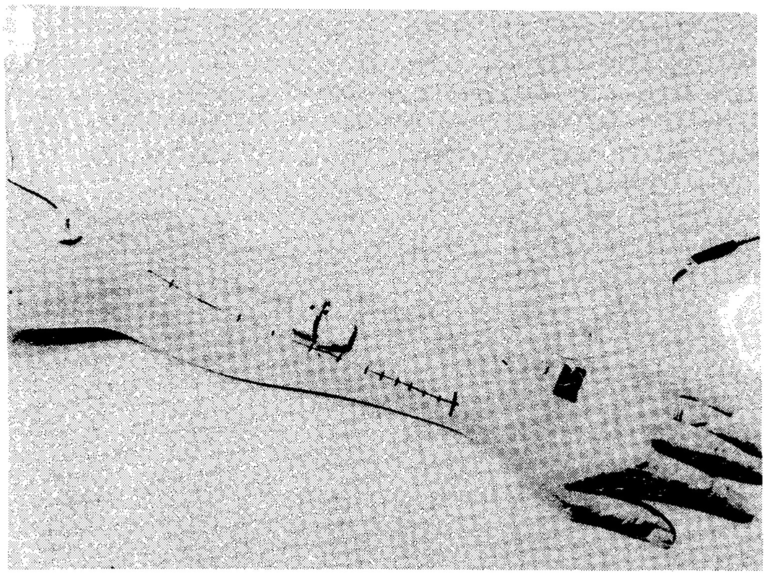

Figure 3. Arrangement of the electrodes on the subject. To the left: pick-up reference and pick-up electrodes. To the right: stimulation electrodes and stimulation reference.

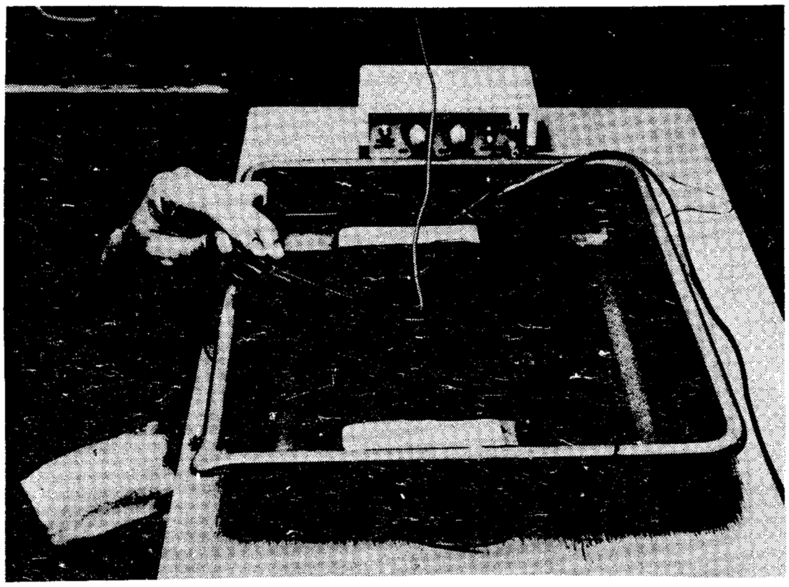

Figure 4. The arrangement of the *in vitro* experiment. Electrode locations are analogous to those of Fig. 3. The stimulator unit is seen in the background.

Results

The interference of a 0.65 mA, 30 Hz stimulation with the amplified, undetected myoelectric signal picked up 70 mm from the stimulation electrodes is shown in Fig. 5. The stimulation pulses are seen as spikes with considerable amplitudes on the EMG.

Figs. 6 and 7 illustrate the pattern of interference with the electrode providing the proportional output. Fig. 6 shows the interference voltage as a function of the distance from the stimulation electrodes to the pick-up electrodes when different stimulation parameters were applied to the experimental subject. Similar patterns were achieved when the stimulation *in vitro* was used. The interference increases rapidly with decreasing distance between the electrodes. High frequencies increase the interference more than high current intensities. Fig. 7 illustrates the behaviour of the interference voltage when the stimulation and pick-up electrodes are very close. The curve was obtained from an *in vitro* experiment but similar curves were also obtained from experiments with the subject.

Fig. 8 shows the interference of different parameters of electrical stimulation with an Otto Bock electrode signal considering the possibility of adjusting the amplification of the electrode.

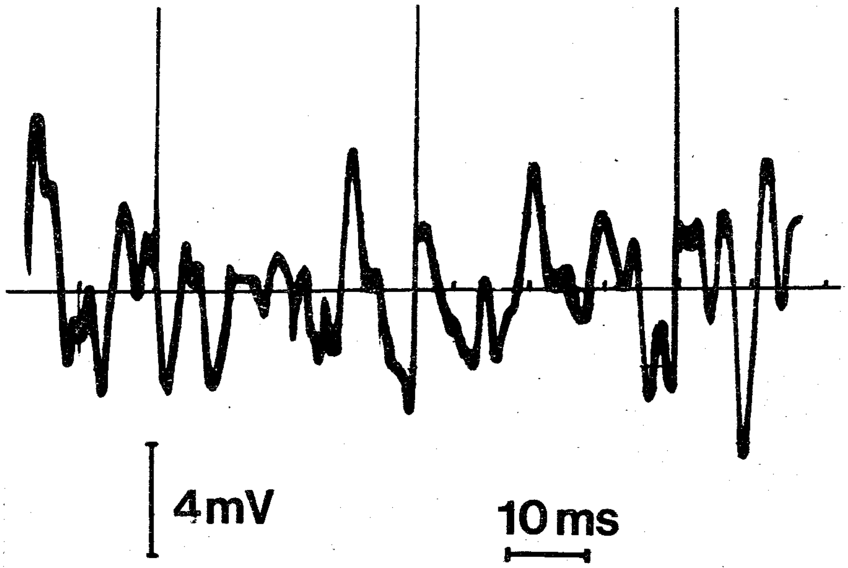

Figure 5. The undetected myoelectric signal with the interfering stimulating pulses. The stimulus was applied (70 mm from the pick-up electrodes) with an amplitude of 0.65 mA and a frequency of 30 Hz.

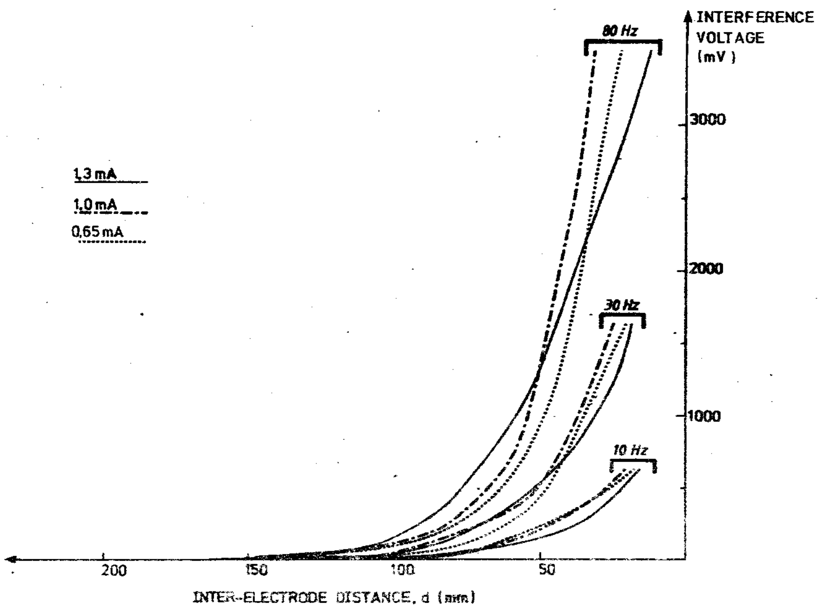

Figure 6. Interference voltage plotted as a function of inter-electrode distance at various stimulation parameters. The results were obtained from measurements on the subject with the active electrode providing the proportional output.

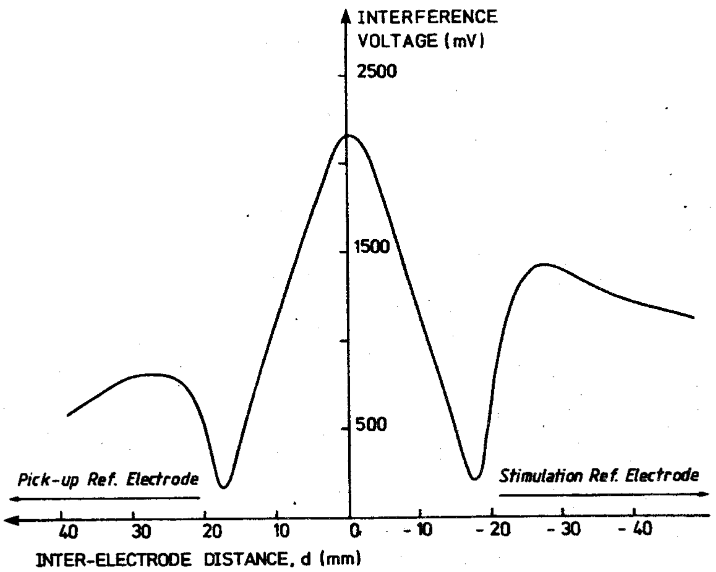

Figure 7. Interference voltage plotted as a function of inter-electrode distance at a stimulating current of 0.65 mA and 10 Hz. The results were obtained in the *in vitro* study with the active electrode providing the proportional output.

Discussion

The clinical usefulness of a sensory feedback system for motorized prostheses will probably be as dependent upon its capacity to be miniaturized and contained within the prosthesis itself as its capacity to transmit information. An electrical stimulator with its control circuits for this purpose can be miniaturized to an extent that it will easily be contained within a conventional myoelectric prosthesis. The energy consumption of such a stimulator is negligible compared to that of the prosthesis motor. The methods of chronic implantation in humans of nerve stimulation electrodes are now so well elaborated and tested that they should not cause major problems (Mooney & Roth, 1976). For these reasons, electrical stimulation for sensory feedback purposes will show to be superior to mechanical vibratory stimulation provided that the interference between the generated electrical field and the prosthesis control system can be overcome.

The experiments were carried out on a non-amputated subject in order to make it easy to place the two needle electrodes within a sensory nerve. In an amputation stump this is much more difficult because scarring has deranged the anatomy. There is, however, no reason to believe that the conditions in an amputation stump otherwise would be different. One experiment was carried out *in vitro*. This experiment yielded very similar interference patterns compared with the experiments done on the subject. The

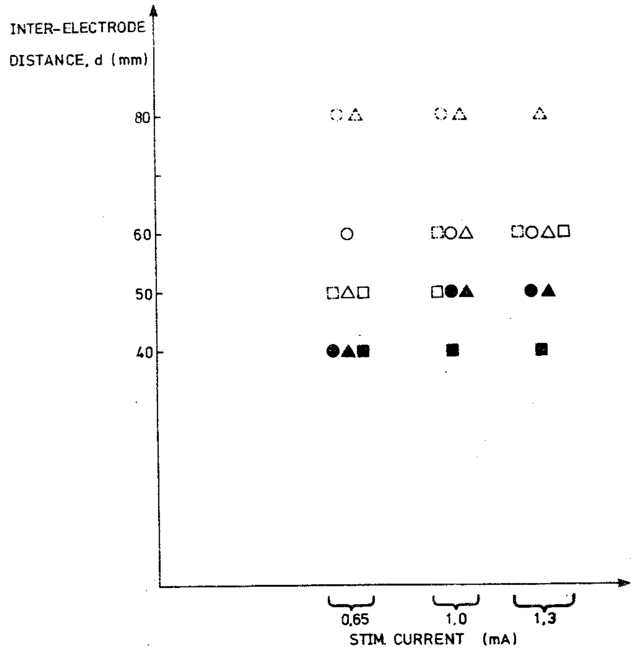

Figure 8. Interference of electrical stimulation on the Otto Bock electrode. Symbols denote the shortest inter-electrode distance where an off response of the electrode was maintained during stimulation at various stimulation parameters and electrode amplifications.
Circles indicate stimulation at 80 Hz, triangles stimulation at 30 Hz, and squares stimulation at 10 Hz.
Dotted symbols indicate electrode amplification 6, open symbols amplification 4 and filled symbols amplification 2.
Results were obtained from measurements on the subject.

reason for doing the experiments *in vitro* was to study the behaviour of the interference voltage when the pick-up and stimulation electrodes were very close. As is shown in Fig. 7, when the inter-electrode distance is very small even minute changes in this parameter will cause considerable variations in the interference voltage. It was found difficult to measure the distance with enough accuracy in the subject, mainly due to the fact that the skin tended to slide over the subcutaneous tissue together with the pick-up electrodes in a quite uncontrollable manner.

The stimulation parameters tested are relevant to intraneural stimulation. With intraneural stimulation it is usually possible to achieve optimal coding of information for sensory feedback purposes by means of pulse amplitude modulation with a current of 0.3 to 1.0 mA. Likewise it has been shown that with pulse frequency modulation optimal coding is achieved between 10 and 80 Hz (Anani et al., 1977).

The interference of electrical stimulation with the amplified, undetected myoelectric signal in a distance as large as 70 mm is considerable, see Fig. 5. However, when the signals are band-pass filtered and rectified by the pick-up electrode and the stimulation frequency is low - which preferably can be the case in sensory feedback electrical stimulation - the interference with the detected signal is considerably smaller. If this type of electrode is used for on-off operation of a prosthesis, a reasonable value for maximum tolerated interference voltage will be around 1000 mV. This will make it possible to apply an 80 Hz stimulation current of 1.0 mA about 60 mm from the pick-up electrodes while a 10 Hz stimulation current with an amplitude of 0.65 mA will not cause an interference exceeding 1000 mV until the inter-electrode distance falls below 15 mm. As shown in Fig. 6, the changes in frequency have a greater influence on the interference voltage than the current intensities tested. This is an effect of the band-pass filtering in the pick-up electrode. The filter characteristics are illustrated in Fig. 9.

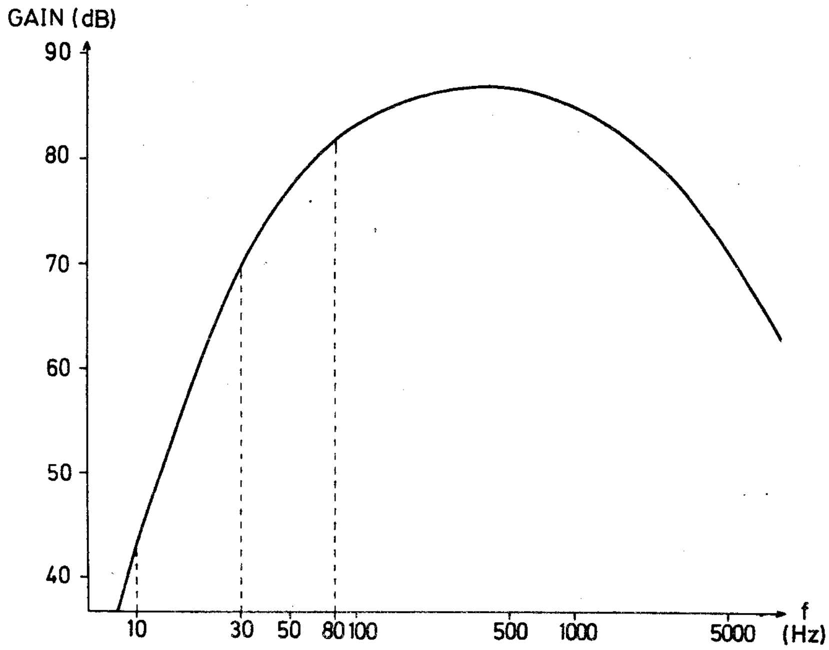

Figure 9. Frequency characteristics for the electrode providing the proportional output. Note the three frequencies 10 Hz, 30 Hz, and 80 Hz that were used in the measurements. The electrode gain is the ratio of the DC-output signal level and the RMS-level of the input signal.

Together with sophisticated prosthesis control systems, e.g. where pattern recognition methods are utilized for identification of myoelectric signal levels (Almström & Herberts, 1975; Lyman et al., 1976), this type of electrical feedback stimulation can also be used. In such systems the control algorithm can be compensated for the interference voltage caused by the stimulation, since the interference occurs simultaneously at all pick-up electrodes. The maximum interference voltage that can be accepted in the Swedish pattern recognition system is estimated to 500 mV, a level that, however, is strongly individual. This level corresponds at 10 Hz to a minimum inter-electrode distance of 40 mm, an important reduction compared to conventional systems when the amputation stump is very short.

From Fig. 8 it can be read that also with an Otto Bock electrode with a commonly clinically used amplification of position 4, the interference is eliminated if the inter-electrode distance exceeds 60 mm. For this electrode the filtering characteristics are different and changes in frequency have a smaller influence on the interference voltage.

The study of the interference voltage when the pick-up and stimulation electrodes come close shows that, as expected, the interference voltage is greatest when the two pick-up electrodes are placed just over the two stimulation electrodes. The interference voltage then rapidly falls to a minimum, the location of which changes with the current intensity. With increasing current, the minimum falls closer to the stimulation electrodes. Changes in frequency do not alter this location.

The interference voltage reaches a new maximum on both sides of the stimulation electrodes when the inter-electrode distance is further increased. The level of the interference voltage of this maximum is asymmetrical on the two sides of the stimulation electrodes, being higher on the side where the stimulation reference electrode is situated. The critical inter-electrode distances for the interference mentioned above refer to the side where the pick-up reference electrode is situated. Thus, the location of the reference electrodes is of great importance.

Conclusion

This investigation shows that it is possible to apply optimal afferent electrical nerve stimulation parameters for prostheses sensory feedback use at a distance exceeding 60 mm from the pick-up electrodes without significant interference with the myoelectric control system. If frequency limitations of the stimulation currents are accepted and only amplitude modulated stimulation at 10 Hz is applied, the inter-electrode distance can be reduced considerably, as is the case with pattern recognition methods. In most amputees it is possible to separate the implanted stimulation electrodes in the nerves of the amputation stump from the pick-up electrodes with the critical 60 mm. Thus, from this point of view, the use of an electrical nerve stimulation feedback system is in accordance with the principle of prostheses self-containment.

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