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Afferent Electrical Nerve Stimulation for Sensory Feedback in Hand Prostheses

Clinical and Physiological Aspects

BY

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AFFERENT ELECTRICAL NERVE STIMULATION FOR SENSORY FEEDBACK IN HAND PROSTHESES

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ABSTRACT

The development of externally energized hand prostheses has been very rapid during the past two decades. The prostheses' control systems have reached a high degree of sophistication. Nevertheless, motorized prostheses are difficult to control as they supply the amputees with much less feedback information about movement, position, and force than a conventional cable-operated device. This creates a demand for sensory feedback systems in externally energized hand prostheses. It is concluded from a review of current neurophysiological theories on kinesthesis that a properly designed myoelectric control system may yield significant amounts of feedback information through action of muscles at the myoelectric control sites. Therefore it seems reasonable to believe that the need for purely artificial sensory feedback systems can be reduced but not totally abolished by further development of control systems. Electrical stimulation of nerves in the amputation stump can be used as an artificial system to convey sensory feedback from hand prostheses. The method has several advantages consistent with the principle of prostheses self-containment, an important factor for patient acceptance of prostheses.

In an experimental study performed on both normal subjects and on amputees, electrical stimulation was applied to nerves in the forearm through percutaneous, intraneural wire electrodes. The subjects' capacity to discriminate changes in stimulation intensity and frequency was studied both with intermittent stimulation and in a tracking task using continuous stimulation. Optimal stimulation parameters for sensory feedback purposes were defined. The interference between the nerve stimulation current and various myoelectric control systems was measured on human subjects and *in vitro*. Nerve stimulation was found to be readily accepted by the subjects. In the amputees the sensations were located to the phantom hand. The discrimination capacity was deemed to be sufficient for sensory feedback applications. In the tracking performances the delay of the response was considerable, but was diminished by training to an acceptable level. The interference between the electrical stimulation and the myoelectric control systems was shown to be insignificant in amputation stumps of ordinary lengths for the pick-up systems studied. This was also corroborated in theoretical calculations using the electrical field theory.

From this study it is concluded that sensory feedback from externally energized hand prostheses primarily should be derived from the operation of the myoelectric control systems. Feedback information not obtainable in this way can be transmitted through electrical stimulation of the stump nerves. This method is safe, reliable, and consistent with the concept of prosthesis self-containment.

Key words: Hand prostheses – Myoelectric control – Sensory feedback – Nerve stimulation – Tracking performance – Kinesthesis – Electrical stimulation – Electric fields.

CONTENTS

This thesis is based in part on the following papers which will be referred to in the text by their Roman numerals I – V.

I. Anani, A.B., Ikeda, K. & Körner, L.: Human ability to discriminate various parameters in afferent electrical nerve stimulation with particular reference to prostheses sensory feedback. Med. & Biol. Eng. & Comput., 1977, 15: 363–373.

II. Anani, A. & Körner, L.: Afferent electrical nerve stimulation: Human tracking performance relevant to prostheses sensory feedback. Accepted for publication in Med. & Biol. Eng. & Comput., 1979.

III. Anani, A. & Körner, L.: Discrimination of phantom hand sensations elicited by afferent electrical nerve stimulation in below-elbow amputees. Accepted for publication in Medical Progress through Technology, vol 6, 1979.

IV. Anani, A., Körner, L., Almström, C. & Herberts, P.: The interference of electrical nerve stimulation with myoelectric prostheses control systems. A theoretical and applied study relevant to prostheses sensory feedback. Technical report no 2: 79. Research Laboratory of Medical Electronics, Chalmers University of Technology, Göteborg, Sweden, 1979.

V. Anani, A. & Körner, L.: An electronic instrument for investigating human response to afferent electrical nerve stimulation. Technical report no 3: 79. Research Laboratory of Medical Electronics, Chalmers University of Technology, Göteborg, Sweden, 1979.

The main responsibility for the technical parts of Papers I – V has been carried by Adi Anani. The technical considerations are summarized in "Afferent electrical nerve stimulation for sensory feedback in hand prostheses. Technical aspects", dissertation for the degree Doctor of Technology at Chalmers University of Technology, Göteborg, Sweden, 1979.

INTRODUCTION

The development of hand prosthetics

According to Sauerbruch (1916), the history of artificial hands is short and simple. None of the classical physicians of the Roman empire has written anything about prosthetics. The first prosthesis mentioned in history is considered to be the iron hand of the Roman soldier Marcus Sergius, who lost his hand in the second Punic war. Obviously, the interest for prosthetics among man is much older than this. The museum at the University of Durham, England is said to house the mummy of an Egyptian of noble birth whose missing hand was replaced by a solid gold prosthesis (Reswick, 1972). One can only assume that this prosthesis served more for cosmetics than for function.

Not until the 16th century was the use of hand prostheses more regularly documented. The functional hand of Goetz von Berlichingen is legendary. It was manufactured by an armourer in Olhausen in the year of 1504. This hand is shown in figure 1. The prosthesis had several passive functions which could be locked in various positions. Other similar prostheses were described over the centuries but not until 1835 was the first actively controlled functional prosthesis invented by Ballif, dentist and engineer in Berlin. This prosthesis applied several of the ideas behind modern cable-operated conventional prostheses. After this breakthrough for cable operation several modified, actively controlled prostheses were presented.

An important milestone in the development of hand prosthetics was the presentation of the ideas of cineplasty introduced by Vanguetty at the end of the 19th century. Cineplasties allow movement of prostheses with the use of the muscles in the amputation stump. In this way, cables and shoulder harnessess can be eliminated. Vanguety's ideas were modified and developed by Sauerbruch who in 1916, published his famous work on cineplasties.

In the 20th century the technical development of conventional prostheses has been rapid, in pace with new materials for the construction of prostheses. The basic ideas for function and movement have, however, remained the same. In recent years the cineplastic procedures have become less popular, and are performed today on special indications only.

Powered prostheses were introduced as early as in 1919 by Borchardt and Schlesinger (Herberts, 1969). These prostheses did not, however, come into widespread clinical use because of their weight and because of the difficulty in controlling them. Control by myoelectric signals was first suggested by researchers working in Germany and a system was implemented by Reiter (1948). Wiener of the Massachusets Institute of Technology was the first to suggest the application of sensory feedback to prostheses (Wiener, 1951). Practical application of myoelectric control in powered prostheses did not, however, become a clinical reality until around 1960. In Western Europe it was Battye *et al.* (1955), and in the Soviet Union it was Kobrinski *et al.* (1961) who pioneered this field. During the last two decades the development of the myoelectric prostheses has been very rapid and has led to highly sophisticated control systems.

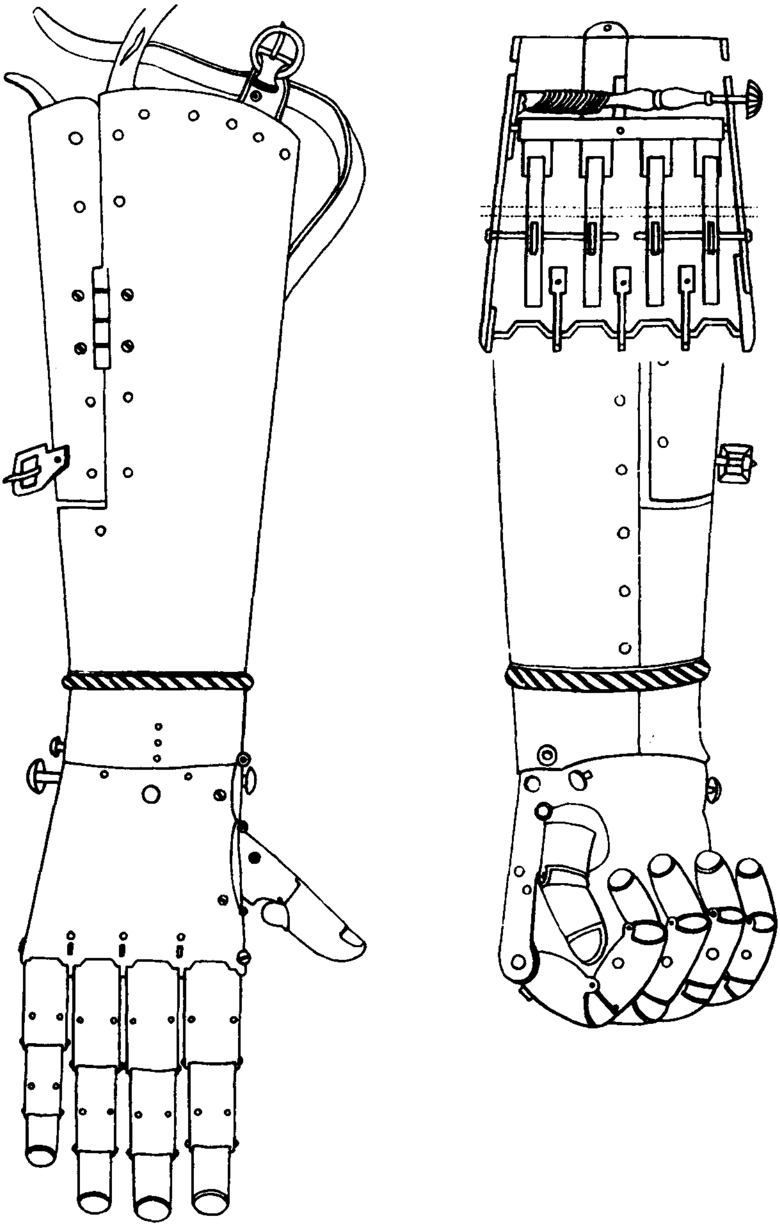

Fig. 1. The iron hand of Goetz von Berlichingen from the 16th century. With a modern glove the cosmetics of this prosthesis would have been fairly good.

The myoelectrically controlled, externally powered prostheses of today are, of course, sophisticated devices. Still, the patient acceptance of the prostheses is quite low (see Discussion). In view of this it may appear strange that over the centuries a significant number of amputees have accepted primitive iron hands without a single function. This illustrates the fact that patient acceptance is only partly dependent on the technical sophistication of the device. The real need of the prosthesis is the most important factor for acceptance. This means that social, psychological, and economic factors affecting the amputee and his handicap finally determine whether a prosthesis will be accepted or not. This should be taken into consideration while working with all kinds of aids for the handicapped.

The concept of sensory feedback

It has been known for a long time that the loss of a hand implies not only the loss of prehension but also the loss of sensory function. The sensation of a hand, as pointed out by Moberg (1964), is a very complex quality which does not lend itself easily to replacement by artificial devices. The development of powered hand prostheses, which has been rapid and successful during the last decade, has therefore mainly aimed towards improvement of the prehension and control function. During this work it has been realized that a functioning prehension is dependent upon certain feedback inputs from the prosthesis.

The term sensory feedback from prostheses has to do with the information which can be relayed from the prosthesis to the amputee about movement, force, and position of the prosthesis. The feedback is a way to compare the output of the machine with the input as illustrated in figure 2. In this context, the term feedback is used with the same meaning as in control engineering (Considine, 1971). Thus, it is an aid to increase the accuracy

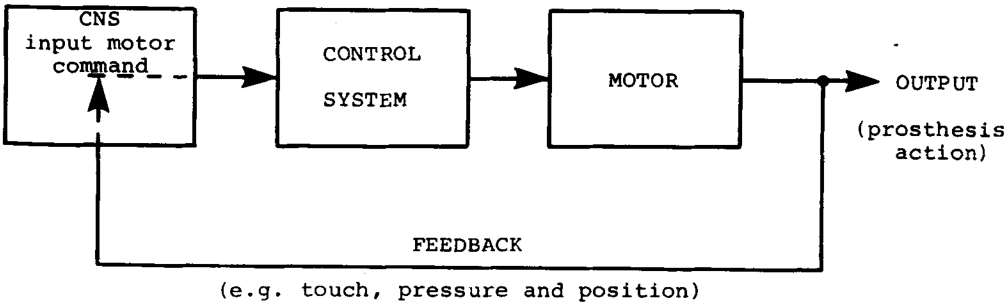

Fig. 2. Definition of feedback control as applied to a man-prosthesis system. A specific parameter of the output (prosthesis action) is fed back to the input (intention to move) and compared with it in a specific way.

of the control system of the prosthesis, and not an artificial touch designed to mimic the sensory function of a normal hand giving the subject information about the world around him. According to Moberg (1976), the concept of sensory feedback for the normal use of a hand is inadequate, since most useful motor actions are answers to various afferent impulses. Even though the sensory function of a hand which is responsible for communication with the world around us may dominate the afferent inflow, feedback functions are also required in order to control movements (Stein, 1974). In the case of prosthesis action the problem is more distinct as the capacity for active touch is missing. Successful replacement of the prehension function of a hand, which is performed with a prosthesis, requires feedback information about force, position, and movement of the function to be controlled. Thus, a prosthesis is primarily a machine for movement and not for sensation. In the normal hand the sensory function is so dominant that the term sensory feedback for hand prostheses easily leads to the misconception that this is a sensory function designed for active touch and communication with the outside world.

Several authors have shown that sensory feedback from prostheses will increase the precision by which the prosthesis is operated (Mann, 1973; Clippinger *et al.*, 1974; Prior *et al.*, 1977). Most researchers working in this field realize that it would be worthwhile to replace the sophisticated and delicate sensory function of a normal hand as well. However, this is not a realistic goal for the nearest future. The most delicate quality of hand sensation is the one responsible for contact between human beings and the world around us (Moberg, 1976). The importance of this touch was emphasized as early as four thousand years ago in rock carvings done by cave men living in Scandinavia. In all probability this natural and sophisticated sensation can never be replaced by any artificial effort of man.

Forchheimer *et al.* (1978) state that no significant improvement of the efficiency of hand prostheses is possible unless the prostheses are able to transfer information about objects in the hand and about the control of the objects. They feel that feedback concerning force and movement in prostheses is important, but that it is secondary to information about the handling of objects. There are considerable technical difficulties in constructing a tactile system which can transmit any meaningful information about qualities of specified objects. Besides, it is questionable whether such systems are ever indicated. As shown by Forchheimer *et al.* (1978) – and certainly not surprising – even the most skilled wearer of a unilateral prosthesis when analyzed in detail has a very one-handed behaviour, with his non-amputated hand as the dominant one. In the bilateral amputee, and certainly in blind bilateral amputees, the indication for cineplasties or for the Krukenberg procedure must be considered. Thus, the need for prostheses with feedback of a somatosensory kind, not directly related to the control of the prosthesis, is limited.

The concept of sensory feedback as dealt with in the present paper is thus related to the motor function of hand prostheses. Feedback conveys important information about force, position, and movement of the prostheses, which enables control of prostheses with increased precision. A somatosensory part is included in this feedback concept, for example the sense of pressure in the grip. However, this is primarily in order to enhance the performance of the control system, and not in order to relay information about the object itself.

General principles for the design of sensory feedback systems

The general principle in designing aids for the handicapped should be to analyze the needs of the patient and to satisfy those needs with solutions as simple, as reliable and as inconspicuous as possible. The importance of self-containment and self-suspension of hand prostheses for patient acceptance has been repeatedly stressed by Childress (Childress, 1973; Childress *et al.*, 1974). It is essential that prospective sensory feedback systems do not violate the principle of self-containment or seriously endanger prosthesis reliability. An equally important principle has been put forwards by Hermann (1973) and Hirsch & Klasson (1974). These authors stress the importance of applying as much as possible of basal physiological principles and knowledge when designing rehabilitation aids.

These concepts form the basis of the present analysis of prosthesis feedback. First, a review of modern neurophysiological theories concerning kinesthesis will be given and the relevance of these theories for prosthesis control and feedback will be discussed. In the second part of the paper, a system for purely artificial sensory feedback is described. One of the most important features of this system is that it satisfies the concept of self-containment in a hand prosthesis.

PURPOSE OF THE PRESENT INVESTIGATION

The purpose of the present paper is two-fold:

(1) To perform an analysis of the indications for artificial sensory feedback in externally powered prostheses.

(2) To describe afferent electrical stimulation of nerves in the amputation stump as a method for providing artificial sensory feedback from motorized prostheses with special reference to the possible incorporation of the system in a self-contained prosthesis. This section is based on an experimental study in man (Papers I–V).

GENERAL ASPECTS OF SENSORY FEEDBACK IN MOTORIZED PROSTHESES

The aim when designing systems for sensory feedback from prostheses should be to obtain as high an acceptance as possible among the patients. As mentioned above, the most important factors in obtaining acceptance probably are self-containment and simplicity of the prosthesis and its feedback system. It is probably better to apply a properly miniaturized feedback system with a low information capacity than a conspicuous system with a high information capacity. One obvious way to achieve sensory feedback is to use the physiological sensations resulting from the actions of the human body necessary for prosthesis control. This has been tried before on motorized prostheses, but with few exceptions the results have been disappointing. Modern neurophysiological theories about human sense of effort and sense of position (Matthews, 1977) make it plausible to assume that significant amounts of sensory feedback could be derived from prosthesis control systems. It is important that the control systems should be developed not only for optimal prosthesis control but also designed in such a way that the control of the prosthesis will produce as much as possible of sensory feedback information.

Sense of muscular force

The neural mechanism of the human sense of muscular force has been a matter of discussion for a long time. Evidence in the literature during recent years gives strong support to the theory that the sense of effort at least to a large extent is a central phenomenon. As early as in the 19th century Helmholtz suggested the presence of a centrally located sensation of innervation responsible for the control of eye movements (Helmholtz, 1962, new, translated edition). Sperry (1950) introduced the term corollary discharge for the responsible mechanism. The corollary discharges are activities in the neural mechanisms connecting the motor and sensory parts of the central nervous system. By a study of the subjective estimation of weight under different circumstances it has been shown that the perception of heaviness is mainly based on sensing the effort or the motor command required to lift or support an object (McCloskey & Gandevia, 1978). Obviously, cutaneous receptors and receptors in the muscles, joints and tendons may signal the force excerted. By paralysing or disturbing these receptors it is, however, possible to show that in most individuals the sensation of muscular force remains unchanged (Gandevia & McCloskey, 1977). Thus, it seems likely that most individuals choose to use the corollary discharges for judgements of muscular force whereas concomitant inputs from muscles, skin, and tendons are neglected. Nevertheless, it is possible that inputs from tendon organs, as well, are important for the sense of muscular tension (McCloskey *et al.*, 1974; Roland *et al.*, 1977). Experiments made by Taube and Berman (1968) on de-afferented monkeys show that complex trained motor behaviour patterns can remain virtually unchanged even after a very extensive de-afferentation of the forelimbs. The authors propose the presence of either some sort of central feedback mechanism or an engraved sensation of motor effort as an explanation of these phenomena. The theory of engraved motor effort is supported by the experiments of Radonjic and Long (1970), who have looked at the EMG-pattern of forearm flexor muscles three years after they have been transposed to function as extensors. In spite of their new function no change in their pattern of innervation was noted. They were still active together with the non-transposed flexors. This was taken as evidence that engraved motor patterns cannot be changed easily.

These central mechanisms are, however, not totally unaffected by peripheral inputs (McCloskey, 1974; Gandevia & McCloskey, 1976a; McCloskey & Gandevia, 1978). It appears that higher structures of the nervous system have the ability to process and integrate spatial and temporal information derived from the two sensory-motor sources, *i.e.* the corollary discharges (sense of innervation) and the peripheral afferent inflow (sense of performance) (Hermann, 1973).

This theory implicates that amputees using myoelectrically controlled prostheses should have a pronounced sense of effort in the muscles at the myoelectric control site. The sense of effort should be available for force proprioception from the prosthesis, since the level of the EMG-signal and the force of the muscle are roughly proportional (Lippold, 1952; Lindström *et al.*, 1974). This concept has been utilized with success in the force proprioception system of the Boston arm (Mann, 1973). It has, however, not been possible to use the control signal without processing. In order to augment the sense of effort, a force-sensitive transducer has been put into the motor axis giving a negative feedback signal which is subtracted from the EMG control signal when the transducer is activated. Thus, the level of the control signal must be increased when the movement is resisted if further movement is to take place. In this way the sense of effort is augmented.

The control of myoelectric prostheses by means of multiple signal pattern recognition was first suggested and described by Finley & Wirta (1967), and was applied for control of multifunctional hand prostheses by Herberts *et al.* (1973). The method is based on the perception of the phantom hand which is present in almost all non-congenital amputees (Solonen, 1962). These amputees are able to perform and feel imaginary movements of the phantom hand. When this is done the muscles in the amputation stump are activated in much the same way as if a normal hand would have performed the movement. Myoelectric signals resulting from this complex muscle activity can be picked up by means of skin electrodes. The relative intensities of the various signals acquired form a pattern specific to the movement imagined. The patterns can be classified by electronic pattern recognition circuits and the prosthesis can be controlled in such a way that the intended movement is executed. This relating of movements of the prosthesis to movements of the phantom perception enables the amputee to control a multifunctional hand prosthesis accurately with a minimal amount of training (Herberts *et al.*, 1978).

Although in most individuals the sense of effort is mainly due to the corollary discharges between the motor and sensory parts of the central nervous system, it is known to be extensively and in a very complex manner affected by peripheral inputs (Dyhre-Poulsen & Djörup, 1976). It is reasonable to assume that the sense of effort is related more to the patterns of specific movements than to the load on isolated muscles. The control of myoelectric prostheses by means of pattern recognition utilizes the same concept of integrated patterns of movement but on a peripheral level. Therefore, it seems reasonable to postulate that the construction of a proportional control signal based on pattern recognition should increase the amount of force proprioception extracted from the prosthesis control system. Investigations in this field are in progress, and preliminary data support the hypothesis (Almström *et al.*, unpublished data).

Sense of position

The view on mechanisms of the human sense of position has until recently been that it is totally dependent upon activities in skin and joint afferents, while afferent inflow from the muscles is unimportant (Rose & Mountcastle, 1959; Merton, 1964). In consequence, it has been regarded impossible to find any neurophysiological basis for a sense of position, derived from the control of a myoelectric prosthesis. However, studies on joint receptors have recently shown that they are not able to transmit information sufficiently detailed to provide a sense of position (Clark & Burgess, 1975). Furthermore, experiments with total joint replacement indicate that the sense of position is unaffected by a complete synovectomy and excision of the fibrous joint capsule (Grigg *et al.*, 1973; Cross & McCloskey, 1973). In experiments with hemiplegic and with partially curarized persons, McCloskey & Torda (1975) have shown that corollary discharges have no part in the human sense of position. Experiments made by Goodwin *et al.* (1972), and by Gandevia & McCloskey (1976 b), give substantial evidence that selective stimulation of muscle afferents produces illusions of movement which are clearly perceived. There is thus indirect evidence opposing the theory that joint and skin afferents are the only mediators of sense of position.

This new neurophysiological evidence led to a re-evaluation of the classical experiments concerning subjective sensations resulting from finger movements with blocked cutaneous and joint afferents. By repeating these experiments Gandevia & McCloskey (1978) demonstrated that the muscle sense has a role in determining joint movements, especially when these movements are actively performed by a muscle.

It is obvious that these findings may be of importance also in the field of prosthetics. As mentioned above, pattern recognition control of a multifunctional hand prosthesis is based on the imaginary movement of the phantom hand (Almström, 1977). Thus, position and movement of the phantom hand serve as references for the amputee in controlling the prosthesis. Some authors claim that the illusion of movement in the phantom is generated by muscle afferents in the stump muscles (Henderson & Smyth, 1948; Granit & Burke, 1973). It seems reasonable to assume that the muscle afferents are a source of sense of position and movement that can also be used for prosthesis position feedback purposes, provided that an appropriate control system is applied. Position feedback obtainable in this way is probably much less accurate than the force proprioception feedback obtainable, since the muscles in an amputation stump work to a large extent isometrically. Present research is aimed at determining how much sense of position can be derived from generation of a proportional control signal in a pattern recognition system.

Sense of touch

The sense of touch and other somatosensory qualities have its peripheral neural elements exclusively in the skin. Consequently, substitution of these perceptions after an amputation must be carried out using purely artificial stimulation. As mentioned above somatosensory perceptions are required for the proper control of prostheses. Therefore, the application of artificial sensory substitution systems is needed for prosthesis feedback. It is important to realize that the primary aim of this sensory substitution is to enhance prosthesis control and not to obtain artificial touch.

Brain plasticity

The theoretical background for the efficiency of purely artificial sensory substitution systems has been given by Bach-y-Rita (1967, 1973). He has defined the concept of sensory or brain plasticity. Physical stimuli, *e.g.* light, sound, pressure, temperature, can activate specific physiological receptors. At the receptors the stimulus is transduced into action potentials in afferent nerves which are fed into the central nervous system where the message is decoded. According to Bach-y-Rita this decoding procedure is a plastic process, which is adaptive and can be changed according to the specific needs of the subject. In a visual substitution system the picture of an object can be coded into vibrations applied to the abdomen of a blind man. After training the blind person no longer feels a pattern of vibrations on his abdomen but actually decodes the vibrations as if he really saw the object. Thus, it seems possible for the brain to adapt to various afferent inflows and to decode incoming messages in a fashion most useful to the individual under the circumstances. The concept of brain plasticity forms the base for applying purely sensory feedback from prostheses.

Conclusions

On the basis of the neurophysiological theories on kinesthesis discussed above, the following philosophy for the design of prosthesis feedback systems is proposed.

(1) Sense of force: Force proprioception should be available mainly through the myoelectric control system. This can either be performed by augmenting the sense of effort when conventional myoelectric control systems are used by applying a negative feedback signal from a force transducer to the control signal (Mann, 1973), or it can be accomplished by a proportional control signal derived from a system based on myoelectric signal pattern recognition (Hermann, 1973; Almström, 1977). For force proprioception the application of artificial sensory systems seems to be unnecessary.

(2) Sense of position and movement: Control systems for prostheses based on pattern recognition of myoelectric signals utilize the position and movement of the phantom for control of the prosthesis. In most amputees there is a distinct kinesthesia perception in the phantom, which might be elicited by muscle afferents from the stump muscles. If this is the case it should be possible to derive some crude form of position or movement feedback from this activity in the muscle afferents. It is not as yet known to what degree this is possible. Development of new control systems specially designed to enhance the kinesthesis originating from muscle afferents

of the amputation stump is necessary before these ideas can be implemented in a prosthesis. In the prostheses of today the sense of movement and position must be fed back through an artificial signal system.

(3) Somatosensory feedback: The sensory elements for touch, pain, pressure, and temperature are lost at the amputation. A purely artificial replacement is therefore the only alternative. It is, however, possible to elicit sensations located in specific areas of the lost extremity by electrical stimulation of the amputation stump nerves. If this is done and the force proprioception can be fed back through the action of the control system the feedback information and the prosthesis control will converge towards the phantom perception. Such a convergence has been shown to increase accuracy in control tasks (Weissenberger & Sherridan, 1962).

It is essential that as much sensory feedback as possible is extracted from the prosthesis control system. This will diminish the time during which the artificial feedback systems must operate, and thus reduce the energy consumption and perhaps also the discomfort of the subject. As shown by clinical experience with control systems based on pattern recognition of myoelectric signals, the use of the movements of the phantom hand is a natural way for an amputee to control his prosthesis (Herberts *et al.*, 1978). Thus, the application of a sensory feedback system built on movements of the phantom perception will probably give a natural feeling to the amputee and require a minimum of training before it can be used properly. In addition, the mechanisms of brain plasticity can allow these imaginary movements to be integrated with the sensations in the phantom hand produced by electrical nerve stimulation.

PREVIOUS SYSTEMS FOR SENSORY FEEDBACK IN PROSTHESES

Several previously developed systems for control of hand prostheses give rise to significant amounts of sensory feedback. Some control systems are deliberatly designed in order to elicit physiological effects producing sensory feedback. This can be made in several ways which are described in detail below. The development of myoelectrically controlled, externally powered hand prostheses has, however, greatly increased the demand for artificial sensory feedback, since these sophisticated prostheses give the amputee much less information on force and movement of the prosthesis than do conventional cable-operated devices (Hirsch & Klasson, 1974). However, also myoelectric prostheses do have some physiologic feedback derived from socket pressure, noise, and movements of the prosthesis. These feedback clues are probably very important for the amputee who uses the prosthesis.

In what follows feedback systems are classified as "semi-physiological" and "artificial". The "semi-physiological" systems to some extent use feedback that is derived from the prosthesis control without the aid of any specially designed device. "Artificial systems" make use of artificial stimulation only.

Semi-physiological sensory feedback systems

Cineplasties

At the end of the 19th century, Vanguetty developed the idea of amputation stump cineplasty (Zanoli, 1957). His ideas were adopted by Ferdinand Sauerbruch who in 1916 published the work on his well-known biceps cineplasty (Sauerbruch, 1916). Two years later the first report on the Krukenberg cineplasty appeared. A cineplasty is a method to enable the utilization of the remaining muscles of the amputation stump for prosthesis movements. By doing so a considerable amount of sensory feedback is transmitted in much the same mode as in a conventional cable-operated prosthesis. In the case of the Krukenberg cineplasty, amputees have the option of wearing a prosthesis or to use the sensitive grip without any prosthesis. Due to its unsightly appearance the Krukenberg stump is today by most authors considered to have an indication only in the case of bilateral forearm amputation. Its only absolute indication is in blind bilateral amputees (Tooms, 1971). A prosthesis driven by a cineplasty has mainly a force proprioception but in the case of a Krukenberg prosthesis probably also quite a good sense of position is present.

Extended physiological proprioception (EPP)

Simpson (1974) has developed systems for achieving position servo control of pneumatic prostheses for amelic children. He uses the proprioception resulting from movements of the clavicle as control site for pneumatic arm prostheses. Movements of the clavicle enable the children to control their prostheses with great skill. Simpson compares his system, which is called extended physiological proprioception (EPP), with a blind person's stick or the golfer's club. In these examples the extra length of the terminal segment can be assimilated into the normal spatial kinesthesia of the subjects. EPP is based on the principle that the angle of the extremity in space is determined and that the central nervous system can adapt to the length of the tool (prosthesis, stick, club).

A system for control of electrically powered arm prostheses by means of EPP was described by Carlson & Primmer (1978). Various tracking and positioning tasks performed demonstrated a significantly increased precision compared to conventional velocity and position control systems.

Force proprioception

The force proprioception developed by Mann for the Boston arm (Mann, 1973; 1974) is a further example of semiphysiological feedback. As mentioned above, the system augments the sense of effort or muscle load of the control site. A transducer reacts to force in the elbow and gives a feedback signal which subtracts from the control signal to the prosthesis. Thus, if the elbow is loaded, the amputee has to increase the level of the control signal in order to make the motors run. This results in an increased sense of effort according to the mechanism described above. In this way the sense of effort is augmented and clearly detectable even though the prosthesis is controlled with a conventional myoelectric system.

Artificial sensory feedback systems

Purely artificial systems can be subdivided into two different groups with regard to where the artificial stimulation takes place. It can either be fed directly into the prosthesis control system, referred to below as automatic systems or to some specific receptor of the amputee by means of different kinds of stimulation, such as auditory, vibratory, electrocutaneous or electrical nerve stimulation.

Automatic systems

A myoelectric hand with an automatic feedback loop has been described by Tomovic & Boni (1962). The feedback allows the fingers to adjust automatically to the shape of the object handled. In addition, the feedback eliminates the need for special locking devices, since the hand is automatically locked by the feedback into any position when the motor is switched off.

Another type of automatic feedback is described by Salisbury *et al.* (1967). They have applied a slippage detector that works by picking up vibrations resulting from the slipping of an object held in a prosthesis grip. The signal from the slippage detector controls the gripping force so that the force will increase as slippage occurs. Thus, an object will not slip until an opening command signal for the grip has arrived and the prehension is safe.

A prosthesis system with both automatic and artificial sensory feedback was constructed by Schmidl (1977). The basic feedback in the prosthesis described is conveyed by electrocutaneous stimulation controlled from a pressure-sensitive transducer in the grip. Schmidl's experience with this kind of system in a prosthesis with very rapid movements showed that it was difficult for the subject to control the pressure of the grip because of the long time needed to react to the sensory feedback stimulus. This was solved by the addition of an automatic feedback system including a switch which for a short while interrupts the movements when prehension is started at the same time as the electrical feedback stimulation is initiated. Thus, the amputee has time enough to interpret the feedback message and react to it properly.

Auditory feedback

Almost all motorized prostheses give the amputees some kind of auditory feedback, since the motor noise often can be related to speed and to force. However, modern motorized prostheses give low noise intensities and the importance of auditory feedback is reduced.

Some authors have worked with artificial auditory feedback. Pfeiffer *et al.* (1969) describe a system designed to provide a substitute tactile sensation from an anaesthetic hand. In the grip of the hand they applied a pressure sensitive transducer which controls an oscillator that admits an audible sound, the frequency of which is a function of the pressure. Similar systems have been developed by Alles and Mann for the purpose of providing information about the position of myoelectric elbows (Clippinger, 1974).

Vibrotactile feedback

The use of vibratory stimuli for the substitution of sensation is an old technique that has been applied frequently. It is possible to elicit a sensation of moving vibrations on the skin by changing the relative intensities of two stationary vibrators working together. A moving vibration produced in this way is called phantom sensation and was described by Békésy (1957). Vibratory stimuli have been used in tactile visual substitution systems described by Bach-y-Rita (1973). The experiences with vibrotactile stimulation for these purposes are favourable. The technique is quite simple and does not entail discomfort or risk to the subject. It is, however, difficult to miniaturize the vibrators making them self-contained within a prosthesis. They consume more energy and are less reliable than electrical stimulators.

Alles and Mann have studied the possibility of substituting the sense of position in a prosthesis by means of phantom sensations resulting from vibrotactile stimulation (Alles, 1970). They measured how much information about the prosthesis elbow angle that could be transmitted by variation in the phantom feeling. A normal subject could transmit 2.8 bits between contralateral elbows by matching the angles without the aid of visual feedback. The phantom sensation resulted in an information measure of 2.11 bits for the same task. A clinical application of the vibrotactile system has been described by Mann (1973; 1974).

The use of vibrotactile stimulation for prosthesis feedback has also been proposed by Bach-y-Rita and Collins (1970). A system devised in Sweden for position feedback through vibrators is described by Hierton *et al.* (1967). A comparison between the properties of vibrotactile and electrocutaneous stimulation systems is provided by Shannon (1976). He concludes that both electrical and mechanical stimulation of the skin are capable of meeting the required performance capacity for sensory feedback and should be considered as viable alternatives. Shannon especially stresses the problem of cross-talk between two separate channels with electrical stimulation and suggests that both vibrotactile and electrocutaneous stimulation should be used if more than one channel is needed.

A conspicuous trend in the literature concerning artificial sensory feedback is that mechanical stimuli has been abandoned during recent years in favour of electrical ones. The reasons for this are presented below.

Electrocutaneous feedback

The first report concerning sensory feedback in motorized hand prostheses using electrical skin stimulation was published by Beeker *et al.* (1967). The information was relayed through skin electrodes situated on the amputation stump. Stimulation intensities of one to five milliamperes were applied using a sine wave of 5 kHz. The relayed information was derived from a pressure sensitive transducer located in the prosthesis' grip and thus the feedback was related to the strength of prehension.

In Japan, Kato and co-workers have applied a system for pressure sensory feedback to a multifunctional myoelectric hand prosthesis. The successful clinical application of this system was described in 1970 (Kato *et al.*, 1970 a). In addition, Kato's group has made a basic investigation of the human cognitional ability for electric stimulation signals applied to the skin (Kato *et al.*, 1970 b). They defined, in detail, the proper stimulation parameters and measured the capacity to convey information as the rate of correct recognition and the amount of transmitted information.

Rohland and co-workers have provided an Otto Bock prosthesis with a semi-conductor strain gauge mounted on the thumb (Rohland, 1975). They state that the prosthesis was easier to control when there was feedback of pressure in the grip. In order to overcome interference with the myoelectric control system the myoelectric signal had to be low-pass filtered.

Extensive research on electrocutaneous stimulation has been carried out by members of the Lyman group in Los Angeles, USA. In a dissertation from this group, Prior (1973) thoroughly describes various qualities of electrocutaneous stimulation with reference to prosthetics. Electrocutaneous feedback applied to an above elbow amputee was described by Prior & Lyman (1975 a). They report improved prosthesis performance with feedback. A sophisticated analysis of suitable stimulation codes and displays of the stimuli has been thoroughly described by Szeto *et al.* (1976 a; 1976 b; 1977) and Solomonow *et al.*, (1977), (Solomonow & Lyman, 1977). Successful applications of electrocutaneous feedback have also been described by other groups, for example Shannon (1979) and Schmidl (1977).

Electrical nerve stimulation

Ever since the days of Helmholtz it has been a well known fact that electrical stimulation of sensory nerves is subjectively perceived. In a review article, Pfeiffer (1968) gives an interesting description of the different biological uses of electrical stimulation up until now. In recent years electrical nerve stimulation has been used increasingly for the treatment of pain (Sweet & Wepsic, 1968; Nashold & Friedman, 1972), and as functional treatment for paralysis (Waters *et al.*, 1975 a). These clinical applications have greatly hastened the technical development of stimulators and percutaneous connectors and have increased the interest in using nerve stimulation for prosthesis sensory feedback purposes as well. This method was first applied by Clippinger (1973, 1974) and was evaluated in a paper 1974 (Clippinger *et al.*, 1974). The applications were reported successful. A similar procedure was described by Reswick & Nickel (1975) and by Reswick *et al.* (1975) but the results were obviously somewhat disappointing (Mooney, 1976; Reswick & Nickel, 1977). In a paper based on Clippinger's material, Walker *et al.* (1977) have made extensive psychophysical measurements concerning the properties of electrical nerve stimulation as applied in Clippinger's system.

Clinical evaluations of artificial feedback systems

The evaluation of sensory feedback systems has mainly been performed on an experimental basis. Most systems have never reached the stage of routine clinical use. The system described by Mann (1968) for the Boston arm was evaluated in practical use by Mann & Reimers (1970). By computer analysis of a large number of trials they found that a vibrotactile feedback as described above reduced the error of the performance by 50 per cent as compared to no feedback. They concluded that the EMG prosthesis with feedback achieved virtually the same kinesthetic performance as a standard mechanical prosthesis. They also stated that the feedback system caused no discomfort to the wearer and was not affected by the environmental conditions of the limb socket.

A somewhat similar evaluation of a system for two-electrode grasp force and hand opening supplementary sensory feedback system for the Veterans Administration/Northwestern University (VA/NU) myoelectric hand was carried out by Prior *et al.* (1977). They reported results of the preliminary evaluation of the system as it was applied in the case of one patient and showed that the system enabled the patient to make accurate judgements of the thickness of six different blocks.

Clippinger *et al.* (1974) reported preliminary results of afferent nerve stimulation feedback in ten patients. Clippinger found that nerve stimulation resulted in a "physiological" sensation of fist clenching. This was considered to be an appropriate interpretation of the stimulus producing a mental image that could be used as a prosthetic replacement for the missing hand. No problems were reported concerning the procedure of implanting the electrodes.

Reswick *et al.* (1975) tested electrical nerve stimulation feedback in a cable-operated prosthesis but found no significant difference when the feedback was on or off. In a more extensive report, Reswick & Nickel (1977) describe the application of a nerve stimulation system in a below-elbow amputee. In this subject the application has been successful and the patient is happy with the device. In two other patients, the application has been less successful due to technical problems with the feedback system.

Mooney (1976) states, concerning the same experiences as Reswick, that myoelectrically controlled prostheses have no significant functional advantage over standard cable control systems. Even with sensory feedback the performance is not better than with a cable-operated prosthesis. Mooney feels that the huge expenditure of technical energy in the area of myoelectric prosthetic research is not justified by sufficient functional advantages at the present time.

In summary, clinical and experimental experience of artificial sensory feedback systems gives the impression that such systems improve the performance in myoelectrically controlled prostheses up to a level equal to that of conventional cable-operated prosthesis. On the other hand, most sensory feedback systems involve a serious increase in complexity of the prosthesis, which makes it less reliable and violates the important principle of prosthesis self-containment. Such disadvantages can make the prostheses unacceptable to the patients, and in spite of a better objective performance in the laboratory, the whole system can turn out to be useless for practical purposes. Ignorance of these facts may explain why so many efforts made in this area up to now have failed.

The feedback systems discussed in this section are summarized in the block diagram of Figure 3.

Conclusions: The case for afferent electrical nerve stimulation in prosthesis feedback

An electrical stimulation system is considered by most authors to be more reliable than mechanical vibrators (Kato, 1970 b). The electrical stimulator can be miniaturized with modern technology to an extent that allows easy containment within the prosthesis itself. The electrodes, both for cutaneous and neural stimulation, are small and present no major problem as to self-containment. The energy consumption of an electrical stimulation system is negligable in relation to that of the prosthesis motors. Thus, several factors favour electrical systems compared to mechanical ones.

In making the choice between electrical skin stimulation and electrical nerve stimulation, the following points can be made. In the control of a myoelectrically operated prosthesis the movement of the phantom hand is used as a reference for the control signal. Stimulation of the stump nerves, in contrast to skin stimulation, will result in sensations located only in the phantom image of the lost hand and not at all in the amputation stump. This convergence of the control and feedback function to one organ, *i.e.* the phantom hand, can be expected to increase the accuracy of the combined control-feedback systems (Weissenberger & Sheridan, 1962).

The problem of interference between electrical stimulation and the myoelectric control systems has been noted by most authors working in this field, *e.g.* Shannon (1979), Rohland (1975) and Prior & Lyman (1975 b). Different approaches to overcoming the interference problem have been applied, such as filters, microswitches and displacement of the stimulating electrodes away from the control system. The interference between the stimulation and the myoelectric pick-up electrodes is among other things dependent upon the current in the stimulation electrodes (Paper IV). Afferent nerve stimulation with implanted electrodes operates with a current intensity on the order of 0.1 – 0.5 mA, while the corresponding skin stimulation requires between 1 and 5 mA, that is, ten times as much. In addition, with stimulation on the skin, the stimulation electrodes are situated on the same surface as the pick-up electrode which furthermore increase the risk for unwanted interference. As shown in Paper IV, the current intensities relevant to intraneural stimulation will cause very little interference with the prosthesis control system.

In summary, it seems that afferent electrical nerve stimulation is the method of choice if low energy consumption, low interference with the control system, and high prospects of self-containment are required. The vital importance of these factors for patient acceptance is unquestionable.

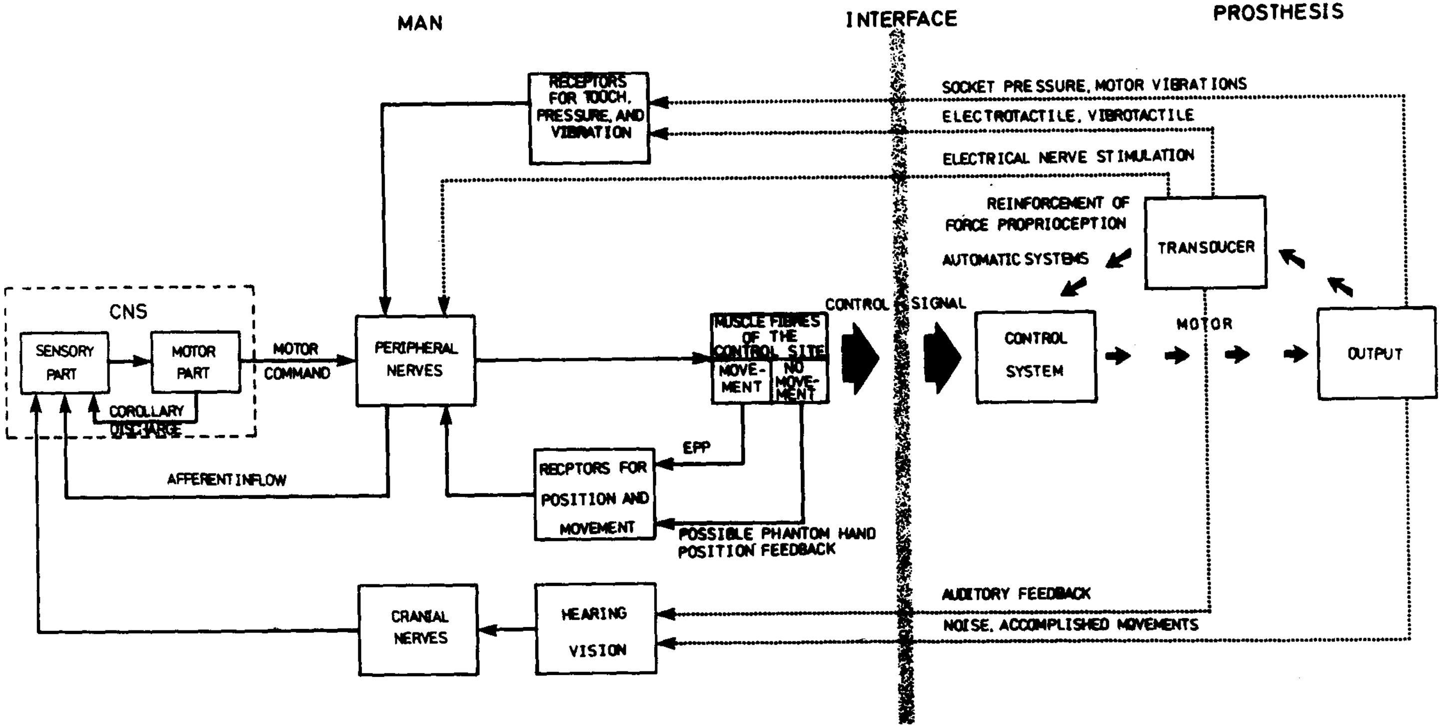

Fig. 3. Block diagram summarizing the possibilities to obtain feedback from powered hand prostheses. The different methods are described in the text.

The obvious drawback of afferent nerve stimulation is that it requires surgical implantation of electrodes. Surgery can be especially dangerous in the amputation stump, where the skin and other tissues often are damaged and thus vulnerable to surgical trauma. The dangers of implantation make intraneural stimulation contra-indicated in certain stumps. On the other hand, techniques for electrode implantation and cutaneous transmission of the stimuli are experimentally and clinically well established. Clippinger *et al.* (1974) describe a safe radiofrequency transmission of the stimuli through the skin. A technique using carbon buttons as percutaneous connectors has been developed to a high degree of sophistication at Rancho Los Amigos Hospital, as described by Mooney & Roth (1976). Several types of implantable electrodes have been developed *e.g.* Caldwell & Reswick (1975), Reswick & Nickel (1977), Friedman (1973), McNeil (1973), some of which have been thoroughly followed up after long time clinical use (Waters *et al.*, 1975 a; 1975 b).

No serious adverse effects of long-time electrical stimulation have been noticed. The stimulation thresholds remain remarkably constant even after long period of stimulation (Clippinger *et al.*, 1974; Reswick *et al.*, 1975). Thus, the documentation of afferent nerve stimulation with implanted electrodes is extensive and allows safe clinical use of the method in cases where the stump itself, because of deranged circulation or sensation, does not present a contra-indication.

ARTIFICIAL SENSORY FEEDBACK IN MOTORIZED PROSTHESES THROUGH ELECTRICAL STIMULATION OF THE AMPUTATION STUMP NERVES

Introduction

Physiological and clinical evidence presented in this paper leaves no doubt that there is a justification for the development of artificial sensory feedback systems for motorized hand prostheses. Several factors favour electrical nerve stimulation for this purpose. It is of great importance, in the author's opinion, to do as much laboratory investigation as possible on the properties of nerve stimulation, as they relate to hand prosthetics, before going into practical clinical trials. Initial difficulties in an applied test may discourage not only the researchers but also the patients and the test will not give the feedback system a fair chance. Hence, the purpose of the present investigation is to provide a theoretical basis for the clinical application of prosthesis sensory feedback systems based on nerve stimulation techniques.

Material and method with comments

The subjects

The evaluation of efficiency of a sensory feedback code must be made from experiments carried out on human subjects. We chose to use normal subjects in the initial part of the investigation in order to minimize the technical difficulties in positioning the electrodes in a nerve. In an amputee the anatomy of the stump can vary considerably due to the trauma responsible for the amputation or to the operation technique used. The nerves are deeply situated and the amputation stump is much more sensitive to pain than a normal arm when inserting an injection needle.

The superficial branch of the radial nerve is situated very close to the skin and is quite large, at a distance of 4 – 6 cm proximal to the wrist joint. This nerve was considered appropriate for the experiments because it possesses no motor function. Thus, the effects of stimulating a purely sensory nerve can be studied. The initial experiments described in Paper I and Paper II were performed entirely on non-amputated subjects. Eleven subjects were studied in the experiments of Paper I. Three of the subjects were the authors themselves. The trained group consisted of the authors only, and the subject with the reproducible sensations was the medical author. He positioned the electrodes and thus, in the experiments performed on himself, was able to adjust the electrodes to positions where he could achieve exactly the same sensations from session to session.

Once the method and the procedure were tested in the group of normal subjects, experiments were carried out on amputees (Paper III). This was done in order to assure that there would be no major difference in the reaction to electrical nerve stimulation between the normal subjects and amputees. Doubts about this may arise from the fact that the sensations of the amputees are allocated to the phantom image of the lost hand. In addition, the nerves stimulated in the amputees are not purely sensory.

As expected, there was considerable trouble when applying the method of percutaneous electrode implantation to the amputees. They felt more pain at the implantation procedure than the normal subjects did. It was somewhat difficult to evaluate what actually was paresthesias resulting from hitting the nerve with the electrode and what was reinforced phantom perception due to the pain of the needle.

The non-amputated, untrained subjects were medical and technical students who volunteered to participate in the investigation. The amputees were chosen from a group of subjects undergoing training in the clinical application of a multifunctional hand prosthesis (Herberts *et al.*, 1978).

The electrodes

It is obvious that afferent electrical nerve stimulation when applied to the amputation stump for purposes of sensory feedback together with a motorized prosthesis should employ implanted electrodes (Clippinger, 1974; Reswick *et al.*, 1975). In planning the present study we felt that surgical implantation of electrodes for experimental purposes alone could not be justified from an ethical point of view, whether it was done on normal subjects or on amputees, who volunteered for the investigation. Accordingly, we were left with the choice of using non-invasive methods, or percutaneous methods. Initially we tried non-invasive transcutaneous methods by applying the stimulation through conductive silver paint spread on the skin over a nerve in the forearm. We did succed in eliciting nerve stimulation in this manner but the stimulation current was as high as for electrocutaneous stimulation (1 – 10 mA), and skin sensations were felt simultaneously with the nerve stimulation paresthesia. Therefore we had to choose an invasive method which was derived from work with electromyography in ergonomics. We prepared fine stainless steel wire electrodes which were inserted through the skin in hypodermic needles (Stuyck & Hoogmartens, 1975). By stimulation through the needles while inserting them we obtained electrode positions close to or in the nerve. When the proper position was obtained, the hypodermic needles were removed and the stimulating electrode was left close to or in the nerve. In the beginning we did not use the hooked wires shown in figure 4. The straight electrodes, however, were very sensitive to movement of the wrist joint. This led us to use the hooked wires which allowed stable electrode conditions.

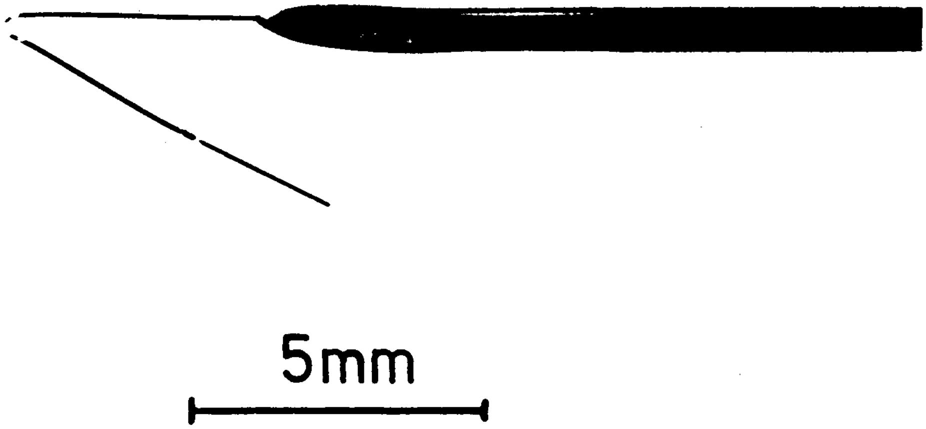

Fig. 4. The hook-shaped wire electrode in a hypodermic needle. 2,5 mm of the electrode tip is deinsulated.

Equipment for stimulation and recording

A complex of new electronic instruments was designed and constructed for the experiments. Engineering details of these instruments are given in Paper V.

Modes of stimulation

The simplest parameters that can be used for coding a message through electrical stimulation of a nerve is the strength and the frequency of the stimulating current. It was considered appropriate to first investigate these parameters before making combinations of them or modulating the message in more complicated terms. It was found that the subjective sensation of the stimulation intensity could be coded either by changing the duration or the amplitude of the stimulating current pulse (Paper I). This is not surprising as the feeling of intensity is mainly dependent upon the amount of delivered electrical charge to the nerve (Gracanin & Trnkoczy, 1975). As there was certain evidence in the literature that very short pulse duration could give painful sensations (Sweet & Wepsic, 1968) we chose to modulate the feeling of intensity by means of pulse amplitude, keeping the duration constant at 100 microseconds.

It is important to space the different intensity levels correctly in order to achieve optimum discrimination capacity. For this purpose the just noticeable difference (JND) between two current intensities was measured throughout the intensity range from threshold to maximum tolerated stimulation. According to Steven's law (Gracanin & Trnkoczy, 1975), a more or less exponential relationship between JND and strength of the current can be expected. This is confirmed by our experimental results. In the curve obtained for the range to be used, however, it seemed reasonable to make a linear approximation of the relationship. Experiments showed that there was no major difference in the discrimination capacity of the intensity levels whether they were spaced exponentially or linearly. In order to make construction of the test equipment easier, we chose to space the intensity levels along a linear scale.

In order to study the optimal conditions for the discrimination of frequency levels, identical experiments were made. These tests showed a quite different relationship which was definitively non-linear in the range to be used. The function seemed to be nearly exponential. Thus, in spacing of the frequency levels an exponential function was used as described in Paper I.

Regardless of the type of modulation used, an artificial sensory feedback code can be applied in principally two different ways, *i.e.* in discrete form or continuously. Discrete stimulation is defined as stimulus applied for a short time interrupted by a longer period of silence. Each stimulation period contains one of the possible levels of stimulation only. Because of the relatively long periods of silence interrupting the stimulation, the subject must make an absolute judgement about the stimulus every time it appears when decoding the message. Access to previous stimulation level for comparison, making relative judgements possible, is not always granted. Therefore, the discrete stimulation mode is probably most useful when information about positions or the status of a specific function is to be conveyed. In Papers II and III the discrete stimulation mode is also referred to as "interrupted stimulation".

Continuous or uninterrupted stimulation is defined as stimulation for an arbitrary period of time where the stimulus is allowed to shift in any mode between the possible levels. Changing of levels takes place while stimulation is presented to the subject. Clearly, this form of code is suitable for signalling dynamic events such as movements and changing of force. Discrimination of this kind of stimulation can be done not only by absolute judgements but also by relative judgements, since the levels are continuously shifting while the subjects are stimulated. In fact, absolute judgements can be difficult to make during continuous stimulation due to more pronounced adaptation, more so than with stimulation in discrete form.

In continuous stimulation tracking performance was studied using four different patterns of stimulation. In two of these stimulation was changing between the same five levels as in discrete stimulation. In one of the five-level-patterns stimulation changed with a speed corresponding to the movement of a fairly rapid prosthesis, while in the second pattern stimulation shifted at half this speed. In the other two stimulation patterns used to study tracking performance, one fast and one slow, stimulation moved between forty discrete levels. With this high number of discrete steps the difference between two consecutive stimulation intensity or frequency levels is smaller than the JND. Stimulation is therefore discriminated as if it could take any value of intensity or frequency within the range studied. These modes of stimulation were considered relevant to prosthetics as described in Paper II.

The properties of discrete stimulation were investigated in Paper I and that of continuous stimulation in Paper II. The amputees taking part in the investigation in Paper III received both discrete and continuous stimulation.

Evaluation of the results

The properties of electrical nerve stimulation were evaluated in both qualitative and quantitative terms. The qualitative evaluations were made by interviewing the subjects who took part in the investigation about the nature of their subjective sensations. In addition, the following parameters were described: Threshold of pain, types of changes in sensations when the stimulating parameters were varied, and extension of paresthesias at different intensities. In the qualitative description of sensations it showed to be extremely useful that the authors themselves took part in the experiments as experimental subjects. The description in Paper I and Paper II is, however, based on the quite uniform sensations of all the subjects.

The quantitative description of the properties of nerve stimulation as a method to convey sensory feedback from hand prostheses has been performed in several different ways. We found it quite difficult to choose parameters which might well be relevant to practical application of the system. It is obvious that some sort of measure concerning accuracy or correctness is required. We chose to measure this as the rate of correct recognition (RCR) which gives the ratio between correctly discriminated levels and total number of levels tested, expressed in per cent. This measure was used both for discrete and continuous stimulation. In continuous stimulation we measured correctness at only one point on each level as indicated in Figure 5. In order to compensate for the delay in tracking this point was chosen about 1.3 seconds after the start of the level. We did not feel, that RCR applied in this way was enough to describe the accuracy of tracking performance. Therefore, we also used the relative absolute error (RAE) as a measure of accuracy. This is a measure of the total amount of divergence between pattern and response curves (see paper II). The concept of RCR was also applied by Kato (1970 b) in his analysis of electrocutaneous stimulation in discrete form. Szeto *et al.* (1976 b) have used a different analysis for determining accuracy when applying continuous electrocutaneous stimulation.

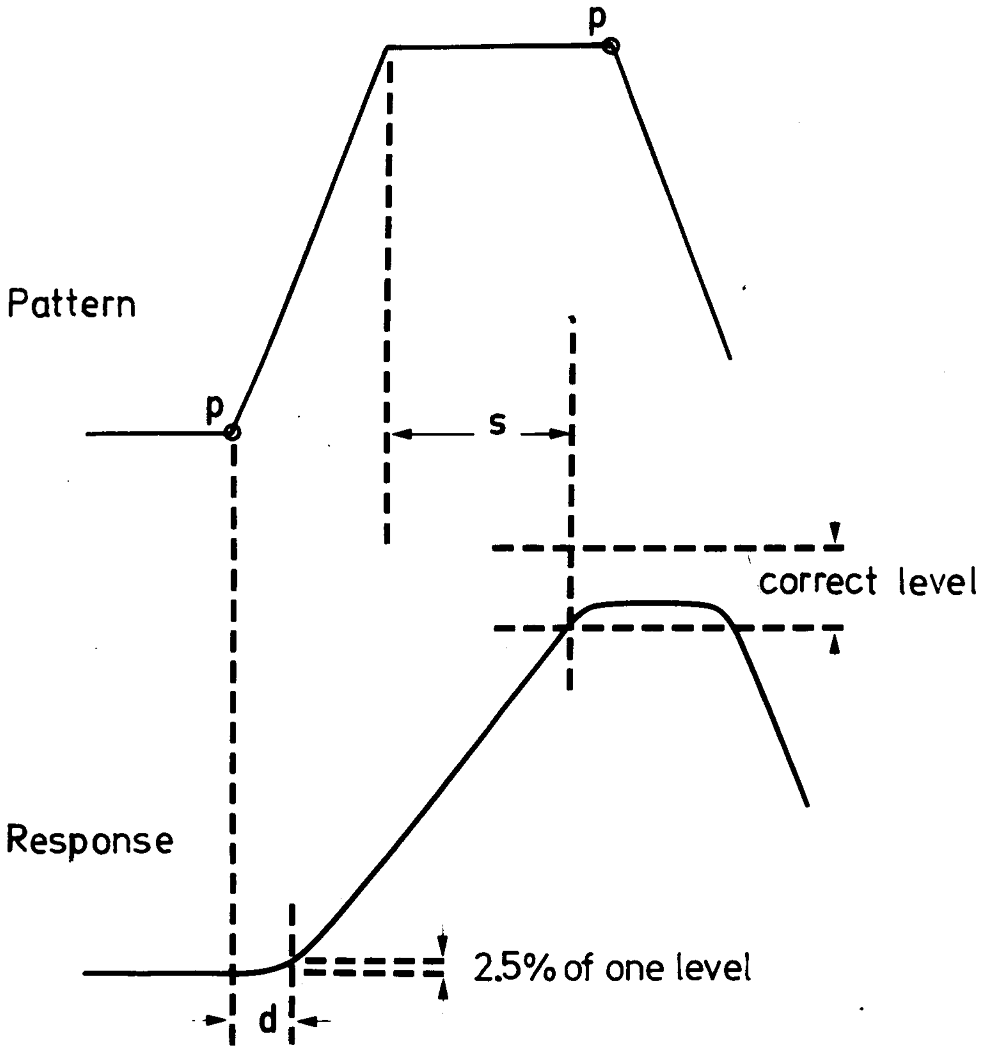

Fig. 5. Definition of simple delay time (d) and settling time (s). The points (p) were used for the calculations of the RCR and the ATI in the continuous stimulation case (see text). A response was considered to be present when the response curve had departed 2,5 per cent of one level from its original position. The response was considered settled when the response curve had reached within a 20 per cent error range from the correct level.

They chose to define an "on target" area as large as 20 per cent of the total dynamic range of stimulation and measured the time during which the response fell within this area, while the subject was under stimulation (cf Paper II).

In order to quantify the amount of information that is relayed through a sensory feedback system we applied the concepts of basic information theory on the stimulation – man situation. We considered the stimulation and the subject's response as transmitting and receiving terminals in a communication system. Calculations according to information theory were then performed as described in Paper I and Paper II. The number of bits per second or bits per symbol that were transmitted through the system could thus be expressed. We have used this measure as a base for comparison between different stimulation modes. We have not, however, tried to implicate any absolute meaning in the concept for the present application in a system for artificial sensory feedback.

In the tracking experiments, the average simple delay of response was measured. Simple delay was defined as the time from a shift in stimulation to the corresponding shift of the subject's reaction as shown in figure 5. In addition, settling time was estimated. This is the period of time starting at the point where the pattern reaches a certain level and ending when the subject has stabilized his response on the same level (see figure 5).

Experimental procedure

The experiments described in Papers I, II and III were performed for each subject at the same session. The schedule in Figure 6 shows the order in which the experiments were performed. The item "training of discrete levels" in fact included both training of discrete levels and tests for discrimination of them. In addition, experiments for determining the JND were done. Each experiment was thus quite extensive and lasted for four to five hours. This offered an opportunity to study if nearly continuous stimulation for this period of time had any detectable adverse effects on the subject. Except for the adaptation phenomena described in Papers I and II no such effects were noticed. The amputees taking part in the experiments described in Paper III underwent the same standardized experimental procedure. Because of the small group of subjects and lack of training of the amputees we have chosen to report only a small part of the data obtained from the amputees. The other data which are not reported here do not, however, differ in any substantial respect from the data obtained from the normal subjects.

The material and method in Paper IV differ from that of the first three papers described here. In the experiments described in Paper IV the interference between electrical stimulation and prosthesis control were studied in a purely experimental situation by simulating the conditions of the forearm. This was done in a bath of saline in which a porous cloth was soaked. The stimulation was applied to the interior of the cloth and pick-up was performed at its surface. Extensive experimentation was done on this set-up which is described in detail in Paper IV. When the pattern of interference through this method had been thoroughly determined, practically and theoretically, some experiments were done

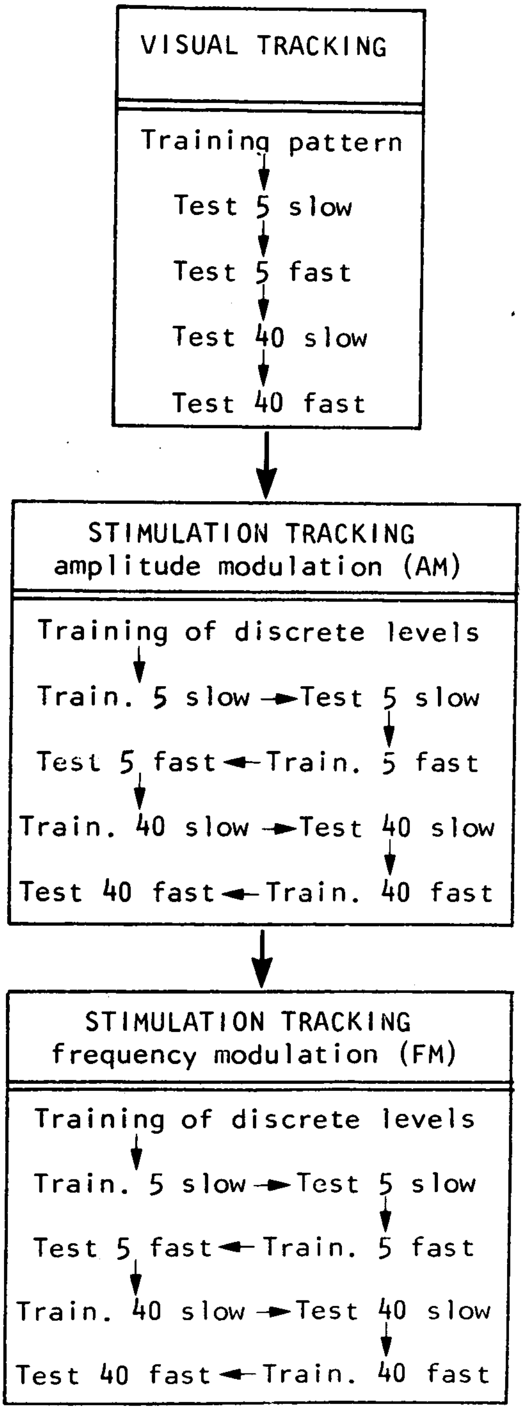

Fig. 6. Experimental procedure.

on one of the authors. These experiments had the character of verification of results obtained and were not as extensive as the experiments made with the experimental set-up.

Stimulation equipment, stimulation parameters, and stimulation electrodes in Paper IV were the same as described for the first three papers.

Summary of the results with comments

In Papers I, II and III the properties of afferent electrical nerve stimulation applied to the forearm of human subjects are studied with relevance to sensory feedback from motorized hand prostheses. The results of these experiments are summarized below.

Qualitative properties

When the sensory nerves were stimulated both in the amputees and in the normal subjects, the subjective sensations resulting from stimulation were located to the receptive fields of the nerves. One of the criteria for intraneural stimulation was in fact that no local sense of stimulation was allowed if the electrode position was to be accepted as intraneural. Thus, the discrimination of stimulation was based entirely upon sensations felt in the hand both in the amputees (the phantom hand) and in the normal subjects. All the amputees had a fairly distinct phantom image. In one of the amputees, where the amputation was performed in the early ninteen-fifties, the phantom hand was located directly on the stump as a result of the well-known telescoping phenomenon described by Henderson & Smyth (1948). This person's image of his phantom hand was somewhat incomplete. The other subjects had a clear phantom image which was shorter than the normal side. There was no difficulty for the amputees in feeling the sensation resulting from electric stimulation in defined areas of the lost hand. The amputees were able to indicate precisely on the contralateral hand where in the phantom the sensation was felt and how it changed if the electrodes were moved or the stimulation intensity shifted. In figure 7 some examples of extension of paresthesias resulting from electrical stimulation is shown for both normal subjects and for amputees.

The location of sensations resulting from stimulation depended on which nerve was stimulated. In larger nerves the position of the electrode within the nerve was also of importance, since it was possible to move the paresthesias resulting from stimulation by moving the electrode within the nerve. In addition, the extension of paresthesia depended on stimulation intensity. With high stimulation intensities the ara where paresthesias were felt was quite large but it diminished with decreased stimulation. As discussed in Paper I, this is due to a greater spread of the electrical field with intensive stimulation compared to weaker one.

None of the subjects, normal or amputees, stated that they felt any discomfort from the stimulation. Usually, the threshold for eliciting sensations, multiplied two or three times indicated the pain threshold or the threshold for eliciting sensation of local stimulation

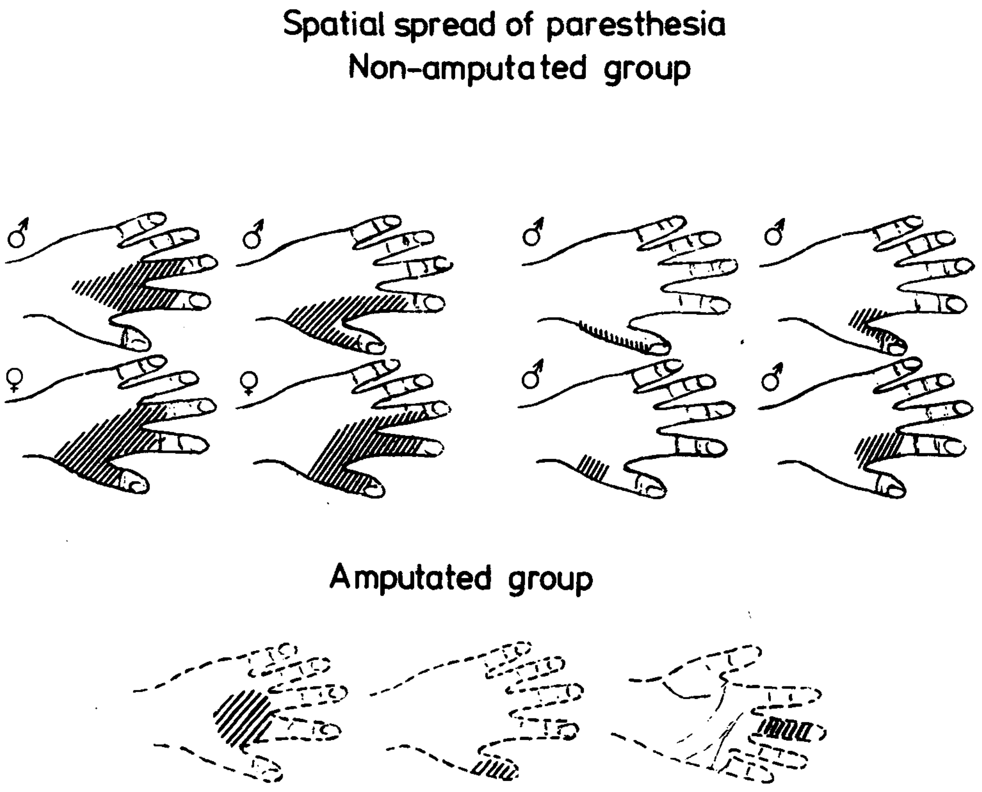

Fig. 7. Spread of paresthesia resulting from stimulation of forearm nerves in normal subjects and in amputees. Extension of paresthesia resulting from maximum tolerated stimulation is indicated. In the group of non-amputated subjects the two cathegories of paresthesia extension are shown.

in the forearm. These intensities of stimulation were never exceeded in the experiments. The intensity range tolerated, however, was quite individual.

The reaction to frequency changes was uniform. The subjects described two types of sensations. One of throbbing for frequencies below 10 Hz and one of buzzing or tingling for frequencies above 10 Hz. The same types of sensations were described by Walker *et al.* (1977). For very high frequencies, above the range used during the experiments, the sensations were changed again but were also more variable among individuals. Some subjects stated that they totally lost the feeling of stimulation above 200 Hz, while others had a weak feeling of stimulation, which increased and decreased periodically at a very low frequency (Paper I).

The adaptation phenomenon after a period of stimulation was clearly detectable. In the discrete experiments described in Paper I, an average increase in threshold of about 10 per cent was noted after 10 seconds of stimulation in half of the experimental group. With a continuous stimulation for two minutes or more all the experimental subjects had a pronounced adaptation, which caused an increase in the intensity necessary for detectable stimulation. The exact amount of this adaptation was not measured but the impression was that it was more pronounced than during the discrete experiments. During the continuous experiments the effect of adaptation had to be compensated for when determining the positions of the specific levels. After compensation the adaptation did not affect the accuracy of the subject's response.

Quantitative properties

The quantitative properties were described in terms of accuracy, transinformation, simple delay and settling time. The general appearance of a tracking performance is illustrated in figure 8. A fair resemblance between pattern and tracking recordings is obvious.

Accuracy

Accuracy was measured as mentioned above in terms of rate of correct recognition (RCR). For tracking experiments relative absolute error (RAE) was also measured. The two measures were linearly correlated to quite a large extent (r = –0.7). The lack of total linear correlation between the two measures is explained by the fact that RCR in tracking experiments measures the correctness in only one point of the curve at each level, while RAE is an intergration of the total error of the curve, thus also including simple delay and settling time.

The results of the measurements of accuracy expressed as RCR are summarized in figure 9. In this figure, the level "75 per cent RCR" is indicated.

Every communication system must have a certain degree of accuracy to make the conveyed information meaningful. Without a thorough clinical evaluation it is very difficult to know which is the lowest acceptable accuracy. It is obvious that a 100 per cent degree of accuracy is not necessary in order to feed back information of such an arbitrary nature as movement and position of prostheses. We have chosen a limit of 75 per cent as a reasonable accuracy, which should be required in order to make the feedback meaningful. Kato *et al.* (1970 b) used an 80 per cent limit in their investigations. As can be read from Figure 9 very few stimulation modes exceed a RCR of 75 per cent. For the untrained subjects it is only those who have a large spread of paresthesia who exceed this level with amplitude modulated stimulation applied intermittently. Considerable training is required to obtain the 75 per cent accuracy in all other cases. This is especially true for continuous stimulation. In these cases the effect of further training was estimated by compensating for the simple delay. As will be described below a decrease in simple delay is a major effect of training. Through this manipulation it can be shown that with amplitude modulation it is always possible to exceed the 75 per cent limit with training. With fre-

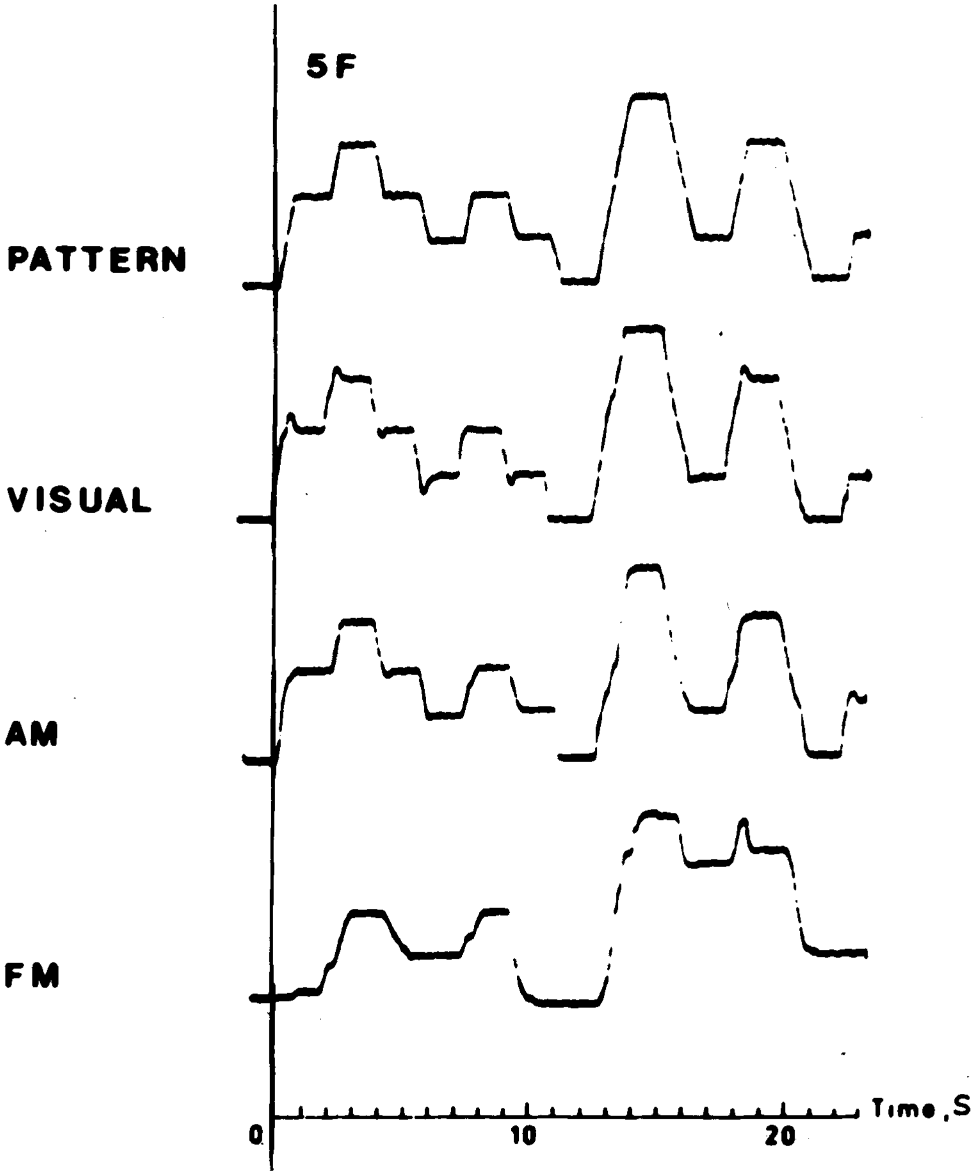

Fig. 8. The tracking performance for stimulation pattern "five-levels-fast" of the trained subject using visual feedback, amplitude modulated stimulation feedback and frequency modulated stimulation feedback.

quency modulation this is only possible in certain stimulation modes, although the accuracy in all instances came very close to the 75 per cent limit. In addition, there was no statistically significant difference between the modes of frequency modulated stimulation exceeding the 75 per cent limit and the ones that did not.

It is obvious that discrimination of amplitude modulated stimulation is easier than frequency modulated stimulation. For the individuals with large spatial spread of paresthesia the difference in discrimination can be explained by the fact that increased intensity is also accompanied by an increased area in which paresthesias are felt. With a small spread of paresthesias, however, absolute judgements about intensity of stimulation must be made. This requires an absolutely stable electrode position, since even minor movements of the electrode will result in great variations of the intensity sensations. With implanted electrodes the possibility to obtain stable stimulation conditions are well documented (Clippinger *et al.*, 1974; Waters *et al.*, 1975 a). Thus, it seems that transmission of information with high accuracy should be done by means of amplitude modulated stimulation. The amputees, however, showed very low degrees of accuracy especially for amplitude modulated stimulation. This is explained by the difficulties in achieving stable electrode conditions during the experiments as described in Paper III.

The degree of accuracy of frequency modulated stimulation was somewhat lower than that of amplitude modulated stimulation when electrode conditions were stable. With unstable electrode conditions as, for example, in the amputee group, it was possible to obtain a higher degree of accuracy with frequency modulated stimulation than with amplitude modulated stimulation. This is, of course, due to the fact that the discrimination of frequency levels is not related to the intensity of stimulation and thus not to stable position of electrodes.

There is a considerable difference in accuracy between trained and untrained individuals. The difference would be even greater if our experimental conditions had enabled us to elicit reproducible sensations of stimulation from session to session. One of the subjects. obtained reproducible sensations which resulted in a higher degree of accuracy. The training had better effect on this subject than on the other trained subjects.

In Paper I, experiments were done in order to study the effect on accuracy of applying an increasing number of levels. Five to six levels seemed to give maximum accuracy. Other psychophysical studies have shown that for physical stimuli discrimination of seven levels plus or minus two is the usual human capacity (Miller, 1967). Therefore, it is not to be expected that the number of accurately discriminated levels could be substantially increased even with training. If more levels are required, it is probably necessary to apply a multichannel approach or to modulate the amplitude and frequency simultaneously. With this procedure, the number of accurately discriminated levels can be expected to increase by about 50 per cent (Miller, 1967).

In Paper II, a comparison is made between accuracy in a system conveying information through nerve stimulation with a system using cutaneous stimulation. The nerve stimulation system shows lower accuracy than the skin stimulation system, which can be explained by problems with unstable electrode positions. A stable position is crucial for eliciting reproducible sensations from session to session and thus, for a true training effect. For the only subject who obtained adequate training in the nerve stimulation group

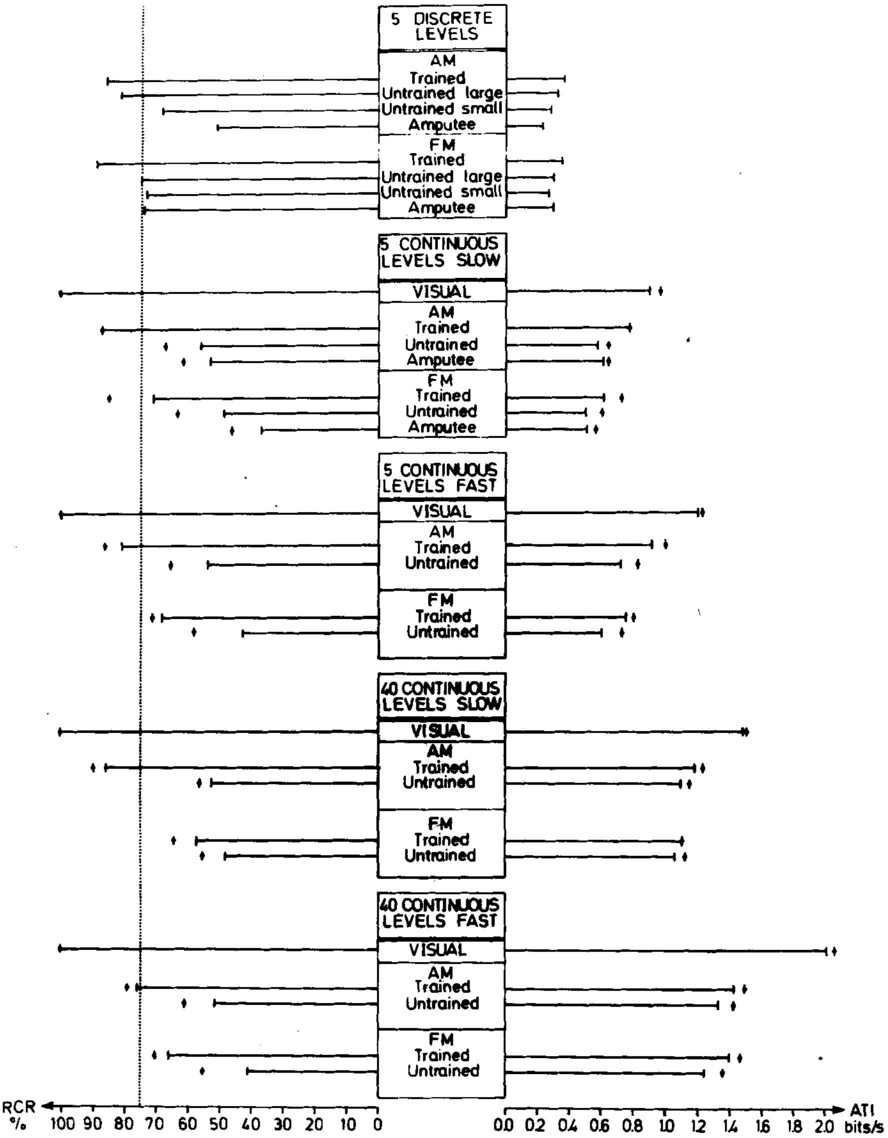

Fig. 9. Diagram summarizing results obtained in terms of rate of correct recognition (RCR) and amount of transmitted information (ATI) in Papers I, II and III. The upper section represents the discrete stimulation experiments and the four lower sections the different continuous stimulating patterns. The results of the trained and untrained subjects as well as of the amputees are indicated for visual, amplitude modulated and frequency modulated stimulation tracking. In the case of discrete stimulation the untrained non-amputated subjects are divided into two sub-groups with regard to the size of the maximum spatial spread of paresthesias (see fig. 7). The bars with circles indicate the values for RCR and ATI after compensation for the delay. The dotted line represents the 75 per cent limit for acceptable accuracy (see text).

discrimination capacity and accuracy was comparable with that of electrical skin stimulation.

Transinformation

The concept of transinformation is related to information theory, and its immediate practical application in the field of prosthetics might be a matter of discussion. However, as a means of comparing the properties of different feedback systems the concept may be of some use. Consequently, we have calculated the transinformation for different modes of stimulation both in discrete and continuous form (see figure 9). It may be of interest to consider how transinformation and accuracy vary together. High transinformation values can easily be obtained by feeding in a lot of information at the input, which results in a slowly increasing transinformation but in a rapidly decreasing degree of accuracy. This connexion is illustrated by comparing the results of tracking of five-level patterns and of forty-level patterns as shown in figure 9 and Paper II. Training, however, will to a certain extent permit an increase in transinformation without a severe decrease in correctness.

Transinformation in discrete stimulation was calculated and expressed in bits per symbol in Paper I. Since each stimulation appeared every fifth second transinformation expressed in bits per second can easily be calculated. This will, however, not give a proper image of maximum transinformation that can be obtained from discrete stimulation per time unit. The discrete stimulation experiments were not performed at a maximum speed. Usually, the subject responded quite promptly to the stimulus and did not need the full four seconds of rest in order to judge the level correctly. An accurate value for maximum transinformation in discrete stimulation will probably be two or three times the value showed in Figure 9. In continuous stimulation, the amount of transmitted information (ATI) was calculated from one point at each level as was discussed in Paper II. It is, thus, not calculated through an integration of all points in the curve.

Alles calculated that a vibrotactile stimulation system could convey an amount of information of 2.11 bits through the use of the above mentioned phantom sensation of Békésy (Alles, 1970). This can be compared to the results of the trained subject in our group which at six discrete levels could obtain a transinformation of 2.05 bits per symbol. The results obtained with our nerve stimulation were better than those from Kato's experiments with electrotactile stimulation (Kato 1970 b). Thus, it seems that the transinformation that can be obtained from electrical nerve stimulation is in the same range as that obtained by electrical or mechanical skin stimulation.

Delay and settling time

In a tracking performance such as the one described in Paper II there is always a delay between the pattern followed and the response. This delay can be considered to have two components. The first one is the simple delay which is the time needed for the person to react to the change in the stimulus. The other component of the delay is referred to in

Paper II as the settling time and is defined in figure 5. The settling time is composed of several parts, one of which is the simple delay mentioned above. Another part is the time needed to operate the response potentiometer to the correct position. This part is affected by characteristics of the response device. Finally, the time needed to analyze the new stimulus and to identify it correctly will also be included. Thus, settling time is considerably longer than the simple delay. In Paper II, the average simple delay time among the untrained subjects was 1.2 seconds while the average settling time was 2.0 seconds. The difference between these two values is statistically significant. The results of the measurements of simple delay in the untrained subjects for different kinds of stimulation is given in Table 1.

	5 LEVELS		40 LEVELS	
	Slow	Fast	Slow	Fast
Visual	0,32	0,27	0,29	0,26
AM	1,33	0,96	1,20	0,71
FM	1,74	1,09	1,38	1,04

Table I. The simple delay in seconds for the untrained subjects tracking with visual feedback, amplitude and frequency modulated stimulation feedback.

As described in Paper II, simple delay is the parameter that is most sensitive to training. Thus, it seems that training increases the ability to identify correctly specific levels but contributes even more to increase the speed by which discrimination can be performed.

The simple delay is considerably different in various kinds of stimulation. As described in Paper II, amplitude modulated stimulation generally gives a shorter simple delay than frequency modulated stimulation. This difference can be explained by the smaller just noticable difference (JND) for amplitude modulated stimulation in comparison to frequency modulated stimulation. Thus, a small change in stimulation intensity is detected earlier than a corresponding small shift in frequency. Accordingly, the subject will respond faster to the shift in amplitude modulated stimulation. A similar explanation can be used for the difference in simple delay between fast and slow moving patterns of stimulation. In fast moving patterns the JND is reached quicker, which is the reason that fast moving patterns have shorter simple delay than slow moving ones.

The settling time, on the other hand, shows no such significant difference between patterns in the material studied. Because of its more complex nature, including properties of the response potentiometer and complex identification processes, which are equal for all patterns, this is to be expected.

The average simple delay for the trained subjects was 0.7 seconds. This may seem a bit long for practical applications in a prosthesis. The corresponding delay for visual tracking is about 0.3 s. However, even the trained individuals in these experiments have little experience with electrical stimulation. As shown in Paper II, the simple delay is very sensitive to training. Thus, one can postulate that the simple delay will decrease significantly with prolonged use of the feedback system. At least for more unspecific performances like on-off responses the delay can be expected to approach the delay with visual feedback.

The settling time is relatively long and is not as easily affected by training as simple delay. For practical applications in the field of prosthetics, however, settling time probably has less relevance than simple delay. In the tracking of the forty-level-pattern described in Paper II the response almost never settles, but the subjects tend to track very arbitrarily at the same rhythm as the patterns. It seems that this type of tracking is relevant to practical application in, for example, a cyclic prosthesis movement. Thus, the subjects are able to estimate with sufficient precision the arbitrary levels at which a movement starts or stops. Such an estimation can be made much faster than discrimination of each level, accurately and separately. Nevertheless, long delay and settling time is probably a property that may seriously limit the practical use of artificial stimulation for prosthesis feedback.

Interference between systems for electrical nerve stimulation and myoelectric signal aquisition

Electric fields from currents used for nerve stimulation will be picked-up by electrodes of prosthesis myoelectric control systems. If the electric fields are strong enough they will interfere with the function of the control system. Theoretical analyses and experiments were carried out in order to illustrate this interference. The results are presented in Paper IV. The electric field as sensed by an active EMG-electrode providing a proportional output is showed in Figure 10. Similar curves can be computed for different stimulation parameters. By computing the interference voltage resulting from stimulation with maximum intensity and frequency relevant to the field of prosthetics, the shortest distance that will not cause interference between stimulation and pick-up electrodes can be determined. Such calculations show that it is possible to apply the implanted stimulation electrode as close as 60 mm from the myoelectric pick-up electrodes without any disturbing interference. This makes it possible to use the system even in very short amputation stumps.

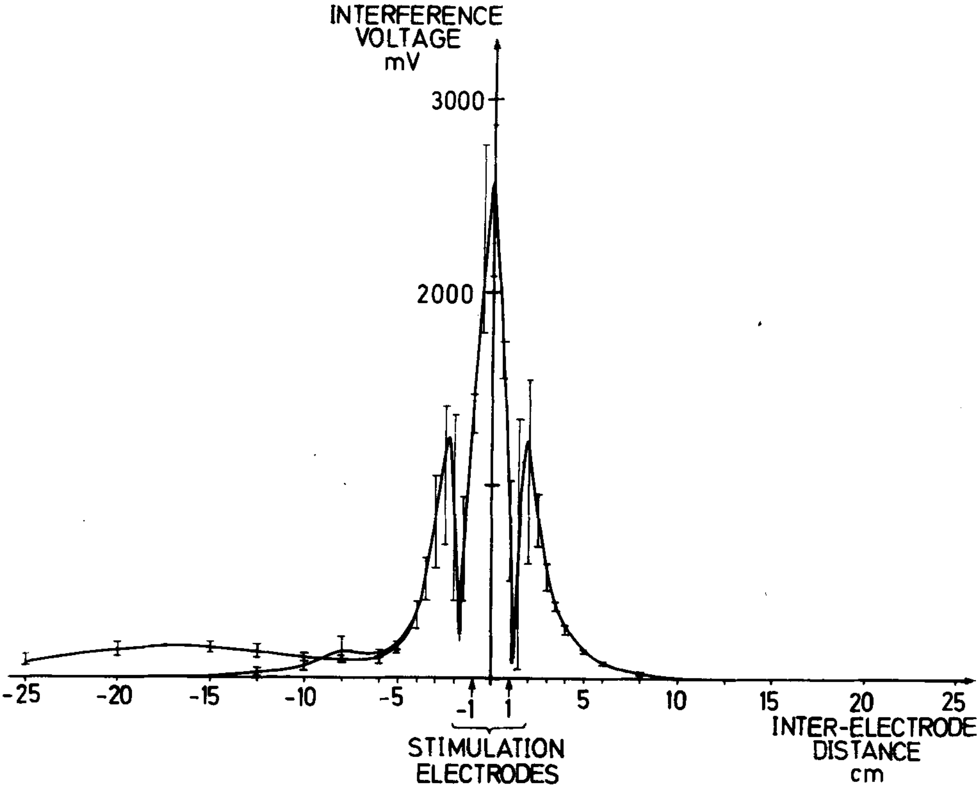

Fig. 10. The interference voltage plotted as a function of the interelectrode distance for various reference electrode positions. Stimulating reference at – 10 and – 25 cm, pick-up reference at – 25, – 10, 10 and 25 cm. As can be read from the curve the interference is high near the active stimulation electrodes and near the stimulation reference electrode. The stimulation parameters used for this curve was 30 Hz, 1,0 mA and 100 microseconds.

Experiments were done to study the interference with a commercially available Otto Bock electrode as well. The results of these measurements are shown in Figure 11. Since the gain of the Otto Bock electrode is adjustable, it is possible to have the stimulating electrode quite near to the pick-up electrode if the myoelectric signals are strong and the gain can be low. If, on the other hand, the myoelectric signals are weak, the interference will be greater and an interelectrode distance between pick-up and stimulating electrode below 60 mm is not advisable. Due to its on-off characteristics, the Otto Bock electrode behaves a little more unpredictably in the electric field than does the electrode with the proportional output. Thus, the practical applicable interelectrode distance for this type of pick-up system probably must be individually tested out.

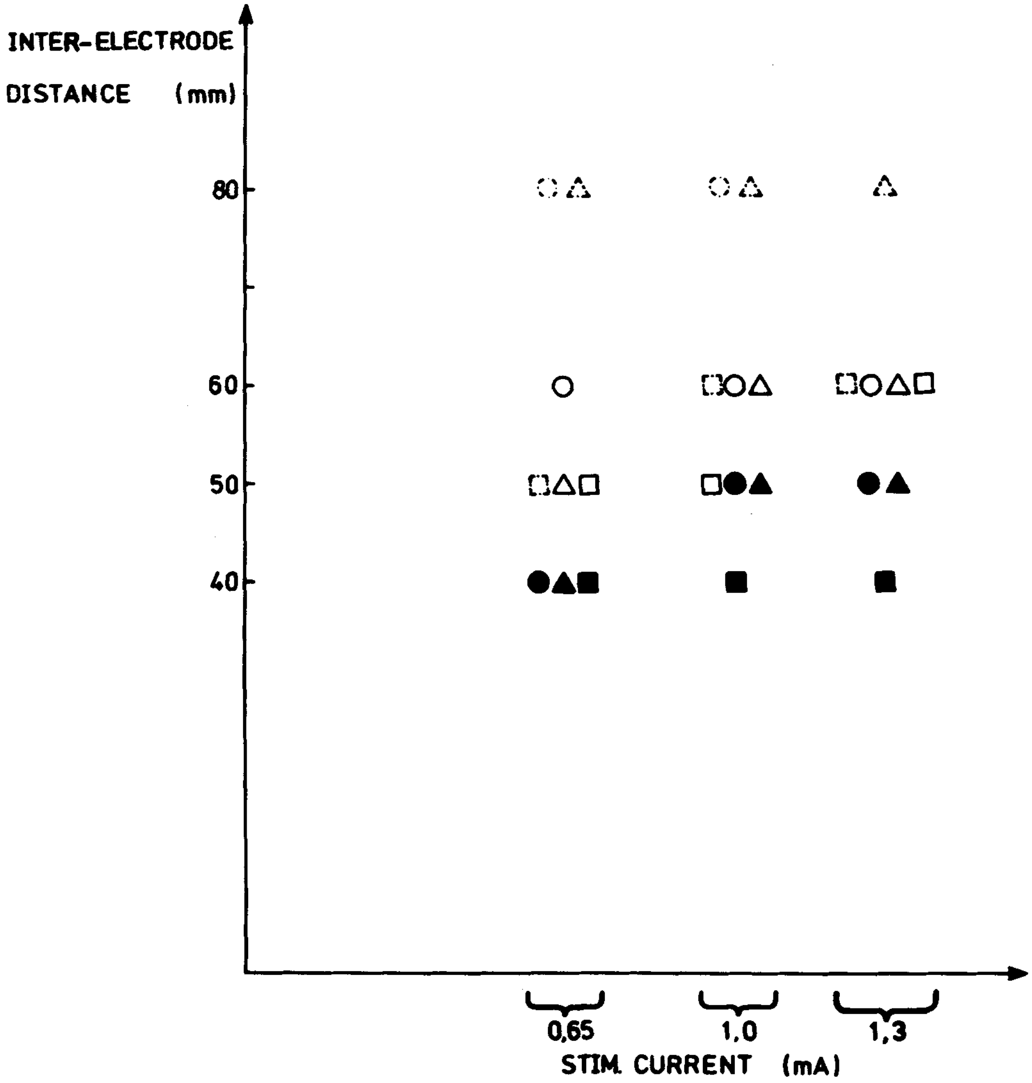

Fig. 11. Interference of electrical stimulation with the Otto Bock electrode. Symbols denote the shortest interelectrode distance where an off-response of the electrode was maintained during stimulation at various stimulation parameters and electrode amplifications. Circles indicate stimulation at 80 Hz, triangles stimulation at 30 Hz and squares stimulation at 10 Hz. Dotted symbols indicate electrode amplification position 6, open symbols position 4 and filled symbols position 2. The stimulating pulse duration was 100 microseconds.

In Paper IV it is clearly shown that electrical nerve stimulation in an amputation stump is definitively compatible with the concept of self-containment of prostheses. It is not necessary to use different reference electrodes for the pick-up and stimulation systems as described by Prior and Lyman (1975) but a common reference electrode situated at the end of the amputation stump is enough. Nor is there any need for separate power supplies. The small energy consumption of the electrical stimulator can be drained from the motor batteries. The deeper the stimulating electrodes are placed in the stump the better. Even if they are placed as close as 1 mm below the skin there will be no interference with the control system if the interelectrode distance exceeds 60 mm, a distance that is obtainable in most amputation stumps. Complex electronic arrangements as applied by other authors for electrocutaneous systems are thus not necessary with electric nerve stimulation.

GENERAL DISCUSSION

The background of the present study and the results of it have been discussed previously in the different papers and above. In this section only some general remarks will be made.

In designing aids for the handicapped it is of vital importance to start with an analysis of the needs and wishes of the patient. Experience shows that compensatory movements with healthy and uninjuried parts of the body, if only socially accepted, are preferred by the patient to the use of artificial devices. The indication for myoelectric hand prostheses in unilateral, below-elbow amputees is therefore a relative one. The same conclusion can be drawn from the results of a recently performed follow-up study of patients receiving myoelectric hand prostheses in western Sweden during the last ten years (Körner *et al.*, unpublished data). As can be seen in Table II, acceptance of the devices is low but is increased by active training and follow-up of the patients. Patients who have accepted their prostheses usually have an occupation forcing them to use two hands.

	Total number of patients	Ceased to use myoprosthesis	Regulary use of myoprosthesis
Prostheses applied without a program for training and follow-up	22	17	5
Prostheses applied according to a program for training and follow-up	16	7	9

Table 2. Acceptance of myoprostheses in a group of unilateral below-elbow amputees.

The indication for myoelectric prostheses in bilateral below-elbow amputees is, of course, stronger if the patient will not accept a cineplastic procedure such as the Krukenberg operation. In below-elbow amputees, who are also blind, the use of prostheses is probably too difficult, and here the indication for some sort of cineplasty is greater.

There are very few below-elbow amputees, who really need myoelectric prostheses. Most of these patients have a defined aim with the use of their devices. They want a prosthesis which is unconspicuous, reliable, self-contained and self-suspended. They will not accept any sensory device which disturbs the function of the prosthesis. Some of them will certainly improve their performance with a prosthesis if they have access to some feedback information on movement, position and force of the prosthesis. In the follow-up study mentioned above about half of the patients wearing myoelectrically controlled prostheses expressed a wish for some sort of sensory feedback mechanism.

With this clinical reality as a basis we have concluded that the indication for sensory feedback from prostheses is limited, and should not be applied without a thorough analysis of the patient's needs. It is clearly shown that feedback from the prosthesis' force and movement affects precision of the control in a positive direction and enables a degree of accuracy at least as high as that of a cable-operated prosthesis.

More recent physiological theories concerning kinesthetic mechanisms give substantial support to the hypothesis that important feedback information can be conveyed by application of a suitable system of prosthesis control. One such system is control of externally powered prosthesis by means of a proportional signal achieved by pattern recognition of myoelectric currents. Further development of this type of prosthesis control using modern microdator technology seems to be the most promising way to achieve sensory feedback from motorized prostheses without compromising the important principles of reliability and self-containment. Further work in the field of systems for proportional control is therefore important. The aim of this kind of development should accordingly not only be to improve the control function, but also to incorporate as much feedback as possible in the prosthesis control. If it is possible to draw feedback from the control of the prostheses there will probably remain only a very small area in which purely artificial sensory feedback systems are indicated.

Afferent electrical nerve stimulation can be applied safely as a feedback mechanism for motorized prostheses. The correct parameters of stimulation for different types of information are thoroughly defined in Papers I, II and III. The patterns of interference between electrical stimulation and control systems can be predicted to a high degree, as described in Paper IV. Electrical stimulation systems can be readily miniaturized to very small dimensions making self-containment of the prostheses possible.

A detailed knowledge about the stimulus to be used is important in order to not have any problems with the system when applied to a patient. Even minor problems will cause irritation and the need for revision of the system, during which time the patient will not have access to his prosthesis. As shown repeatedly by earlier experience discontinuous access to prostheses is a sure way of hindering acceptance. These types of initial problems will probably to a large extent account for the mixed experiences of different authors with feedback systems.

It is well-known that many patients using myoelectric prostheses complain about fatigue in the stump muscles. The fatigue can be explained by monotonous, isometric work performed by the muscles at the control site. Another explanation might be that the loss of the hand causes a decreased afferent inflow in the nerves, which can give rise to an increased sense of effort. As shown by McCloskey and Gandevia (1978), there is probably a very strong tonic inhibitory influence from the periphery on the sense of effort. Stimulation of certain receptive peripheral fields with relation to a movement will diminish the sense of effort when performing this movement. On the other hand, loss of inhibitory influence, for example, after local anaesthesia of the receptive field, will increase the sense of effort. It is plausible to assume that some of the fatigue experienced by patients controlling myoelectric prostheses is, in fact, due to an increased sense of effort. This increase can be accounted for by the loss of the inhibitory influence from the amputated hand. If this is true, application of sensory feedback through stimulation of nerves from the lost hand will restore this tonic inhibition and thus diminish the fatigue experienced by the patients. Work is in progress to test this hypothesis.

Finally, it should be repeated that hand prostheses of today are devices for substitution of prehension. Sensory feedback is mainly a method to increase the accuracy of prehension. The feedback is certainly not substitution for the complicated sensory function of a normal hand. Nevertheless, it can be concluded, that a modern motorized multifunctional hand prosthesis including an adequate feedback function will provide a very good substitute for normal prehension in a properly selected patient.

GENERAL SUMMARY AND CONCLUSION

Some modern neurophysiological theories concerning kinesthesis are reviewed with special reference to their relevance for motorized prosthesis control. The review led to the conclusion that, if appropriate control systems are applied, they will provide important feedback from the prosthesis. The need for purely artificial sensory feedback can therefore be limited to situations in which the basic information from the control system is insufficient or should be reinforced. It is also stressed that application of a system of sensory feedback to a motorized prosthesis must originate from the patient's expressed need of such a device. It should not be routinely applied together with a myoelectric prosthesis. The feedback system must never endanger the self-containment and the reliability of the prosthesis.

The properties of afferent electrical nerve stimulation as a purely artificial sensory feedback system for motorized prostheses have been thoroughly investigated. Characteristic of this system is that it will allow self-containment of the prosthesis, it is simple and technically reliable. The method requires implantation of electrodes around the nerves in the amputation stump. This is no major problem, since electrode implantation for chronic stimulation is well established.

In an experimental study nerve stimulation parameters relevant to prosthesis sensory feedback were defined. It was found that amplitude modulated stimulation gives a high degree of accuracy if the electrode position is stable. With unstable electrode position, frequency modulation is more reliable. The delay in tracking performance is shorter with amplitude modulated stimulation than with frequency modulated stimulation. In order to obtain an acceptable degree of accuracy in the discrimination of electrical stimulation a substantial amount of training is necessary.

The interference between electrical nerve stimulation and systems of prosthesis control has been investigated both from a practical and theoretical point of view. It has been found that with the charges used in nerve stimulation the interference will not pose a major problem. Nevertheless, some minimal interelectrode distances between stimulation and control system must be maintained.

Afferent electrical nerve stimulation was found to give sensations which were accepted by the subjects. The stimulation could be modulated for adequate information transfer so that no pain or discomfort occurred. In the amputee the stimulation was felt in the phantom image of the amputated hand. Thus, both the prosthesis control system and the feedback system use the phantom image as a reference. This can be of importance for the proper cooperation of the systems.

To conclude, afferent electrical nerve stimulation is appropriate for conveying sensory feedback in hand prostheses for the following reasons: An electrical stimulation system can without difficulty be included in a self-contained hand prosthesis, it can convey sufficient amounts of information with acceptable accuracy and delay, and its energy consumption is low. In addition, the risk for disturbing interference with the prosthesis control system is minimal. Sensations elicited by nerve stimulation in amputees will be felt in specific parts of the phantom hand. Such phantom hand sensations offer an opportunity for increased precision in controlling prostheses, since the feedback may be integrated with movements of the phantom hand used for prosthesis control.

The indications for artificial feedback systems are limited, especially if recent theories on kinesthetic mechanisms will lead to design of control systems giving feedback through the physiological actions of the muscles at the control site. With such control systems artificial sensory feedback is needed only for somatosensory qualities of sensation, and as supplements and reinforcements to feedback obtained from the control system. It is important that feedback for hand prostheses is applied only after a thorough analysis of the patient's needs and attitude towards the handicap. Routine application of feedback systems without consideration of these ideas are liable to fail.

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REFERENCES

ALLES, D.S. (1970) Information transmission by phantom sensations. IEEE Transactions on Man-machine Systems. MMS–11, 85–91.

ALMSTRÖM, C. (1977) Myoelectric control of multifunctional hand prostheses – contributions to the pattern recognition approach, to signal acquisition, and to clinical evaluation. Technical report 79. School of electrical engineering, Chalmers University of Technology, Göteborg, Sweden.

BACH-y-RITA, P. (1967) Sensory plasticity. Acta Neurol. Scand., 43, 417–426.

BACH-y-RITA, P. (1973) Sensory substitution. Neural mechanisms underlying prosthetic applications. In: Neural organisation and its relevance to prosthetics. Ed.: Fields, W.S. Intercontinental medical book corp., New York and London. pp 149–163.

BACH-y-RITA, P. & COLLINS, C.C. (1970) Sensory substitution and limb prostheses. In: Advances in external control of human extremities. ETAN, Yoguslavia. pp 9–21.

BATTYE, C.K., NIGHTINGALE, A. & WILLIS, J. (1955) The use of myo-electric currents in the operation of prostheses. J Bone Joint Surg. 37–B, 506–510.

BEEKER, T.W., DURING, J. & DEN HERTOG, A. (1967) Artificial touch in a handprosthesis. Med. & Biol. Engng. 5, 47–49.

von BÉKÉSY, G. (1957) Sensations on the skin similar to directional hearing, beats and harmonics of the ear. J Acoust. Soc. Am. 29, 489–501.

CALDWELL, C.W. & RESWICK, J.B. (1975) A percutaneous wire electrode for chronic research use. IEEE Transactions on Biomedical Engineering. BME–22, 429–432.

CARLSON, L.E. & PRIMMER, K.R. (1978) Extended physiological proprioception for electric prostheses. In: Proc. of the Sixth international symposium on external control of human extremities. Suppl. Yugoslav committee for electronics and automation, Belgrade. pp 201–211.

CHILDRESS, D.S. (1973) Powered limb prostheses: Their clinical significance. IEEE Transactions on Biomedical Engineering. BME–20, 200–207.

CHILDRESS, D.S., HOLMES, D.W. & BILLOCK, J.N. (1974) Ideas on myoelectric systems for upper-extremity amputees. In: The control of upper-extremity prostheses and orthoses. Eds: Herberts, P., Kadefors, R., Magnusson, R. and Petersén, I. Thomas, Springfield, Illinois. pp 86–106.

CLARK, F.J. & BURGESS, P.R. (1975) Slowly adapting receptors in cat knee joint: Can they signal joint angle? J Neurophysiol. 38, 1448–1463.

CLIPPINGER, F.W. (1973) A system to provide sensation from an upper extremity amputation prosthesis. In: Neural organization and its relevance to prosthetics. Ed: Fields, W.S. Intercontinental medical book corp., New York and London. pp 165–176.

CLIPPINGER, F.W. (1974) Experimentation at Duke University. Interclin. Information Bull. 13, 4–6.

CLIPPINGER, F.W., AVERY, R. & TITUS, B. (1974) A sensory feedback system for an upper limb amputation prosthesis. Bulletin of Prosthetics Research. BPR 10–22, 247–258.

CONSIDINE, D.M. (1971) Encyclopedica of instrumentation and control. McGraw-Hill, pp 137–138.

CROSS, M.J. & McCLOSKEY, S.I. (1973) Position sense following surgical removal of joints in man. Brain Research. 55, 443–445.

DYHRE-POULSEN, P. & DJÖRUP, A. (1976) The effect of sensory input on reflex load compensation. Acta Physiol. Scand. Suppl 440, 58.

FINLEY, F.R. & WIRTA, R.W. (1967) Myocoder computer study of electromyographic patterns. Arch. Phys. Med. & Rehab. 48, 20–24.

FORCHHEIMER, A., KLASSON, B., MARKSTRÖM, G. & RYDELL, N. (1978) Experiences from clinical evaluation of upper extremity prostheses. In: Proc. of the 6th International symposium on external control of human extremities. Suppl. Yugoslav committee for electronics and automation, Belgrade. pp 11–24.

FRIEDMAN, H. (1973) Energy considerations in the design of implantable electrodes for peripheral nerve stimulation – a preliminary report. In: Advances in external control of human extremities. ETAN, Belgrad. pp 603–614.

GANDEVIA, S.C. & McCLOSKEY, D.I. (1976) Perceived heaviness of lifted objects and effects of sensory inputs from related, non-lifting parts. Brain Research. 109, 399–401.

GANDEVIA, S.C. & McCLOSKEY, D.I. (1976) Joint sense, muscle sense and their combination as position sense measured at the distal interphalangeal joint of the middle finger. J. Physiol. (Lond.) 260, 387–407.

GANDEVIA, S.C. & McCLOSKEY, D.I. (1977) Sensations of heaviness. Brain 100, 345–354.

GOODWIN, G.M., McCLOSKEY, D.I. & MATTHEWS, P.B.C. (1972) The contribution of muscle afferents to kinaesthesia shown by vibration induced illusions of movement and by the effects of paralysing joint afferents. Brain. 95, 705–748.

GRANCANIN, F. & TRNKOCZY, A. (1975) Optimal stimulus parameters for minimum pain in the chronic stimulation of innervated muscle. Arch. Phys. Med. Rehabil. 56, 243–249.

GRANIT, R. & BURKE, R.E. (1973) Eds. Report on a conference sponsored by the Fogarty international centre for advanced study in health sciences. Brain Research. 53, 1–28.

GRIGG, P., FINERMAN, G.A. & RILEY, L.H. (1973) Joint position sense after total hip replacement. J. Bone Joint Surg. 55–A, 1016–1025.

von HELMHOLTZ, H. (1962) Physiological Optics. Translated by Southall, J.P.C. Dover, New York.

HENDERSON, W.R. & SMYTH, G.E. (1948) Phantom limbs. J. Neurol. Neurosurg. Psychiat. 11, 88–112.

HERBERTS, P. (1969) Myoelectric signals in control of prostheses. Acta Orthop. Scand. Suppl 124.

HERBERTS, P., ALMSTRÖM, C., KADEFORS, R. & LAWRENCE, P.D. (1973) Hand prosthesis control via myoelectric patterns. Acta Orthop. Scand. 44, 389–409.

HERBERTS, P., ALMSTRÖM, C. & CAINE, K. (1978) Clinical application study of multifunctional prosthetic hands. J. Bone Joint Surg. 60–B, 552–560.

HERMAN, R. (1973) Augmented sensory feedback in the control of limb movement. In: Neural organization and its relevance to prosthetics. Ed: Fields, W.S. Intercontinental medical book corporation, New York and London. pp 197–215.

HIERTONN, T., OLDBERG, B., MEYER, K.H., HEDBERG, Å., TOVE, P.A. & MÖHL, F.N. (1967) An above-elbow prosthesis with EMG controlled prehension. Digest of the 7th International conference on medical and biological engineering, Stockholm, Sweden. p 457.

HIRSCH, C. & KLASSON, B. (1974) Clinical aims and desires for today's arm prosthesis. In: The control of upper-extremity prostheses and orthoses. Eds: Herberts, P., Kadefors, R., Magnusson, R. and Petersén, I. Thomas, Springfield, Illinois, pp 58–62.

KATO, I., YAMAKAWA, S., ICHIKAWA, K. & SANO, M. (1970a) Multifunctional myoelectric hand prosthesis with pressure sensory feedback system, Waseda hand 4P. In: Advances in external control of human extremities. ETAN, Yoguslavia. pp 155–170.

KATO, I., KUMAMATO, S., TAMURA, S. & TSUNEKAWA, Y. (1970b) Human cognitional ability for electric stimulation signals. In: Advances in external control of human extremities. ETAN, Yoguslavia. pp 69–84.

KOBRINSKI, A.E., BOLKHOVITIN, S.V., VOSKOBOINIKOWA, L.M., IOFFE, D.M., POLYAN, E.P., POPOV, B.P., SLAVUTSKI, Y.L., SYSIN, A.Y. & YAKOBSON, Y.S. (1961) Problems of bioelectric control. In: Automatic and remote control. Proc. 1st Int. Congr. of the International federation of automatic control, Moscow, 1960. Butterworths, London.

LINDSTRÖM, L., MAGNUSSON, R., and PETERSÉN, I. (1974) Muscle load influence on myoelectric signal characteristics. Scand. J. Rehab. Med. Suppl. 3, 127–148.

LIPPOLD, O.C.J. (1952) The relation between integrated action potentials in a human muscle and its isometric tension. J Physiol. (Lond), 117, 492–499.

MANN, R.W. (1968) Efferent and afferent control of an electromyographic, proportional rate, force sensing artificial elbow with cutaneous display of joint angle. Proc. Instn. Mech. Engrs. 183Pt3J, 86–92.

MANN, R.W. (1973) Prostheses control and feedback via noninvasive skin and invasive peripheral nerve techniques. In: Neural organization and its relevance to prosthetics. Ed: Fields, W.S. Intercontinental medical books corp., New York and London. pp 177–195.

MANN, R.W. (1974) Force and position proprioception for prostheses. In: The control of upper-extremity prostheses and orthoses. Eds: Herberts, P., Kadefors, R., Magnusson, R., and Petersén, I. Thomas, Springfield, Ill. pp 201–219.

MANN, R.W. & REIMERS, S.D. (1970) Kinesthetic sensing for the EMG controlled Boston Arm. IEEE Transactions on Man-machine Systems, MMS–11, 110–115.

MATTHEWS, P.B.C. (1977) Muscle afferents and kinaesthesia. Br. Med. Bull. 33, 133–142.

McCLOSKEY, D.I. (1974) Muscular and cutaneous mechanisms in the estimation of the weights of grasped objects. Neuropsychologia. 12, 513–520.

McCLOSKEY, D.I., EBELING, P. & GOODWIN, G.M. (1974) Estimation of weights and apparent involvement of a "sense of effort". Experimental Neurology. 42, 220–232.

McCLOSKEY, D.I. & GANDEVIA, S.C. (1978) Role of inputs from skin, joints and muscles and of corollary discharges in human discriminatory tasks. In: Active Touch. The mechanism of recognition of objects by manipulation. Ed. Gordon, G. Pergamon Press. pp 177–187.

McCLOSKEY, D.I. & TORDA, T.A.G. (1975) Corollary motor discharges and kinesthesia. Brain Research 100, 467–470.

McNEAL, D.R. (1973) Peripheral nerve stimulation, superficial and implanted. In: Neural organization and its relevance to prosthetics. Ed. Fields, W.S. Intercontinental medical books corp. New York and London. pp 77–99.

MERTON, P.A. (1964) Human position sense and sense of effort. Symposium of the Society for Experimental Biology. 18, 387–400.

MILLER, G.A. (1967) The Psychology of Communication – Seven Essays Penguin Books, pp 21–50.

MOBERG, E. (1964) Aspects of sensation in reconstructive surgery of the upper extremity. J Bone Joint Surg. 46–A, 817–825.

MOBERG, E. (1976) Reconstructive hand surgery in tetraplegia, stroke, and in cerebral palsy: Some basic concepts in physiology and neurology. J Hand Surg. 1, 29–34.

MOONEY, V. (1976) Sensory feedback in upper-extremity amputees. Clin. Orth. Rel. Res. 119, 274–275.

MOONEY, V. & ROTH, A.M. (1976) Advances in percutaneous electrode systems. Biomaterials, Medical Devices and Artificial Organs. 4, 171–180.

NASHOLD, B. & FRIEDMAN, H. (1972) Dorsal column stimulation for control of pain. J. Neurosurg. 36, 590–597.

PFEIFFER, E.A. (1968) Electrical stimulation of sensory nerves with skin electrodes for research, diagnosis, communication and behavioral conditioning: A survey. Med. & Biol. Engng. 6, 637–651.

PFEIFFER, E.A., RHODE, C.M. & FABRIC, S.I. (1969) An experimental device to provide substitute tactile sensation from the anaesthetic hand. Med. & Biol. Engng. 7, 191–199.

PRIOR, R.E. (1973) Study of electrocutaneous parameters relevant to dynamic tactual communication systems. PhD dissertation, Univ of California, Diss Abs Int., p 167–B.

PRIOR, R.E. & LYMAN, J. (1975) Electrocutaneous sensory feedback for artificial limbs: single electrode encoding of grasp force and hand position from an externally energized hand. Read at 5th International symposium on external control of human extremities, Dubrovnik, Yoguslavia.

PRIOR, R.E. & LYMAN, J. (1975) Electrocutaneous feedback for artificial limbs. Bulletin of Prosthetics Research. BPR 10–24, 3–37.

PRIOR, R.E., MASON, C. & LYMAN, J. (1977) Supplemental sensory feedback (SSF) for the VA/NU myoelectric hand: Two electrode grasp force and hand opening system. Proceedings of 1977 ISPO World Congress New York. p 87.

RADONJIĆ, D. and LONG, C. (1970) Why myoelectric control is so difficult. Proc. 3rd Int symp on external control of human extremities. ETAN, Dubrovnik, pp 59–67.

REITER, R. (1948) Steuerung von Geräten durch Körperpotentiale. Grenzgebiete der Medizin. 1, 1.

RESWICK, J.B. (1972) Development of feedback control prosthetic and orthotic devices. Advances in Biomedical Engineering. Eds: Brown and Dickson, Academic Press, London and New York, 2, 139–217.

RESWICK, J., MOONEY, V., SCHWARTZ, A., McNEAL, D., SU, N., BEKEY, G., BOWMAN, B., SNELSON, R., IRONS, G., SCHMID, P., & SPERRY, C. (1975) Sensory feedback prosthesis using intraneural electrodes. Proc. of the 5th Int. symp. on external control of human extremities. ETAN, Yoguslavia. pp 9–24.

RESWICK, J.B. & NICKEL, V.L. (1975) Annual report of progress, Rehabilitation engineering center at Rancho Los Amigos Hospital, Los Angeles. Dec 74–Nov 75.

RESWICK, J.B. & NICKEL, V.L. (1977) Annual report of progress, Rehabilitation engineering centre at Rancho Los Amigos Hospital, Los Angeles, Dec 1975–Jan 1977.

ROHLAND, T.A. (1975) Sensory feedback for powered limb prostheses. Med. & Biol. Engng. 12, 300–301.

ROLAND, P.E. & LADEGAARD-PEDERSEN, H. (1977) A quantitative analysis of sensations of tension and of kinesthesia in man. Evidence for a peripherally originating muscular sense and for a sense of effort. Brain. 100, 671–692.

ROSE, J.E. & MOUNTCASTLE, V.B. (1959) In: Handbook of Physiology, Section I: Neurophysiology. American Physiological Society, Washington DC. Vol. I, 387–429.

SALISBURY, L.L., COLMAN, A.B. & MARCUS, L.F. (1967) An electromechanical hand with automatic proportional control of prehension. Digest of the 7th International conference on medical and biological engineering. Stockholm. p 456.

SAUERBRUCH, F. (1916) Die willkürlich bewegbare künstliche Hand. Julius Springer, Berlin.

SCHMIDL, H. (1977) The importance of information feedback in prostheses for the upper limbs. Prosthetics and Orthotics International. 1, 21–24.

SHANNON, G.F. (1976) A comparison of alternative means of providing sensory feedback on upper limb prostheses. Med. & Biol. Engng. 14, 289–294.

SHANNON, G.F. (1979) A myoelectrically-controlled prosthesis with sensory feedback. Med. & Biol. Eng. & Comput. 17, 73–80.

SIMPSON, D.C. (1974) The choice of control system for the multimovement prosthesis: extended physiological proprioception (epp). In: The control of upper-extremity prostheses and orthoses. Eds.: Herberts, P., Kadefors, R., Magnusson, R., & Petersén, I. Thomas, Springfield, Ill. pp 146–150.

SOLOMONOW, M., LYMAN, J., & FREEDY, A. (1977) Electrotactile two-point discrimination as a function of frequency, body site, laterality and stimulation codes. Annals of Biomedical Engineering, 5, 47–60.

SOLOMONOW, M. & LYMAN, J. (1977) Artificial sensory communications via the tactile sense, space and frequency optimal displays. Annals of Biomedical Engineering. 5, 273–286.

SOLONEN, K.A. (1962) The phantom phenomenon in amputated war veterans. Acta Orthop. Scand. Suppl. 54.

SPERRY, R.W. (1950) Neural basis of the spontaneous optokinetic response produced by visual neural inversion. J. Comp. Physiol. Psychol. 43, 482–489.

STEIN, R.B. (1974) Peripheral control of movements. Physiological Reviews. 54, 215–243.

STUYCK, J. & HOOGMARTENS, M. (1975) A suitable type of wire for wire electrodes. Electromyogr. Clin. Neurophysiol. 15, 291–292.

SWEET, W.H. & WEPSIC, J.G. (1968) Treatment of chronic pain by stimulation of fibres of primary afferent neuron. Trans. Am. Neurol. Ass. 93, 103–105.

SZETO, A.Y.J., PRIOR, R.E., DUNCAN, J.S. & LYMAN, J. (1976 a) Comparison of electrocutaneous codes for sensory feedback. Proc. of the San Diego Biomedical Symposium. 15, 109–114.

SZETO, A., PRIOR, R.E. & LYMAN, J. (1976 b) Evaluation of electrotactile codes. Presented at the 29th Annual Conference on Engineering in Medicine and Biology. Boston, Mass, Nov 6–10, 1976.

SZETO, A.Y.J., PRIOR, R.E. & LYMAN, J. (1977) Electrocutaneous pulse rate and pulse width psychometric functions. Proc. of 1977 ISPO World Congress, p 92.

TAUB, E. & BERMAN, A.J. (1968) Movement and learning in the absence of sensory feedback. In: The neuropsychology of spatially oriented behavior. Ed: Freedman, S.J. The Dorsey Press, Homewood, Ill. USA, pp 173–192.

TOMOVIĆ, R. & BONI, G. (1962) An adaptive artificial hand. IRE Trans. Autom. Control. AC–7: 3, 3–10.

TOOMS, R.E. (1971) In: Campbell's operative orthopaedics. 5th Ed. Ed.: Crenshaw, A.H., Mosby, St Louis. p 881.

WALKER, C.F., LOCKHEAD, G.R., MARKLE, D.R. & McELHANEY, J.H. (1977) Parameters of stimulation and perception in an artificial sensory feedback system. Journal of Bioengineering. 1, 251–260.

WATERS, R.L., McNEAL, D., & PERRY, J. (1975 a) Experimental correction of footdrop by electrical stimulation of the peroneal nerve. J Bone Joint Surg. 57–A, 1047–1054.

WATERS, R.L., McNEAL, D.R., & TASTO, J. (1975 b) Peroneal nerve conduction after chronic electrical stimulation. Arch. Phys. Med. Rehabil. 56, 240–243.

ZANOLI, R. (1957) Krukenberg-Putti amputation-plasty. Journal Bone Joint Surg. 39–B, 230–232.

WIENER, N. (1951) Problems of sensory prostheses. Bulletin of the American Mathematical Society. 56, 27–35.

WEISSENBERGER, S., & SHERRIDAN, T.B. (1962) Dynamics of human operator control systems using tactile feedback. Journal of Basic Engineering, 84, 297–301.