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MYOELECTRIC SIGNALS IN CONTROL OF PROSTHESES

Studies on arm amputees and normal individuals

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CONTENTS

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I. Power Spectra of Myoelectric Signals in Muscles of Arm Amputees and Healthy Normal Controls.
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II. Electrodes for Myoelectric Control of Prostheses.
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III. Implantation of Microcircuits for Myoelectric Control of Prostheses.
Herberts, P., Kadefors, R., Kaiser, E. & Petersén, I.: J. Bone Jt Surg. 50-B, No 4: 780, 1968.

IV. Measurement and Evaluation of Myoelectric Control in a Man-Machine System.
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The papers are referred to in the text by their Roman numerals.

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INTRODUCTION

As a result of the great number of traumatic injuries to the extremities during World War II, there was a renewed interest in various types of prostheses. During latter decades great importance has been attributed to the rehabilitation of patients with impaired motor function by means of orthoses. Prostheses and also orthoses were previously designed so that the force, required by the device, had to be mobilized by the patient. The standard method of controlling the conventional prostheses and orthoses in the upper extremities is a mechanical control by gross body movements of the trunk and the extremities, transmitted by means of cables attached to the harness. This type of mechanical control is disadvantageous mainly because the force of the movements in the prosthesis will be limited, the range of motion will be small and, at the same time, the necessary movements will cause the patient discomfort and fatigue. Furthermore, the functions of the prosthesis are limited as several movements cannot be performed simultaneously without important parts of the body being bound to the control function in a very disturbing way.

Advanced technical research has permitted the design of new types of prostheses and orthoses which are supplied with an external source of power. Already in 1919 a method of converting electric power into mechanical movements in prostheses was published (*Borchardt* & *Schlesinger*, 1919). During the interwar period this type of prosthesis and similar designs were unpractical, because of their weight, and the difficulties for the patients to control them adequately. Two different sources were later utilized to deliver power to the prostheses and the orthoses, *viz.* pneumatic and electric power. For pneumatic power liquid carbon dioxide was used, for instance, in "The Heidelberg pneumatic arm prosthesis" (*Weil*, 1959) and "The Hendon pneumatic prosthesis motor" (*Kinnier Wilson*, 1960). The great advantages of pneumatic prostheses, as compared to electric, were their smooth, comparatively noiseless running and their light weight.

During World War II and the following wars, there was a renewed interest in prostheses equipped with external power. Towards the end of World War II, *Wilms* in Berlin began designing an electric hand prosthesis which later has developed into the so-called French electric hand (*Lucaccini, Kaiser* &

Lyman, 1966). Research of many years' duration began in the USA in 1946 to design an electric arm prosthesis, which finally resulted in five different types (*Alderson*, 1954). A common phenomenon in these American models was that only a few control movements were required of the amputee to release multiple sequentially coded movements in the prosthesis. This proved to be unnatural for the patient and a considerable amount of practice was required before the amputees could use these prostheses. Thus, they were not generally accepted. The great problem has been to enable the patients to control prostheses and orthoses which are operated by external power. The difficulties become even greater with complicated devices having more degrees of freedom. For each direction of motion the action, as a rule, demands one control site, *i.e.* a part of the body from which control impulses can be picked up.

The control of the first electric prostheses introduced after World War II was carried out mechanically by wires or by utilizing the change in volume to which a muscle is subjected when contracted. Mechanical transducers were fitted in the socket of the prosthesis, causing an impulse when the amputee contracted his stump muscles. *Berger & Huppert* (1952) first suggested that in order to obtain a better control of the externally powered prostheses, the electric signals, generated in connection with muscular contraction, should be used as a control signal. Originally, this possibility was mentioned by *Wiener* (1947), who pointed out that myoelectric signals could be detected in the stump muscles of arm amputees, and that these signals would probably allow control of a prosthesis (*Bottomley*, 1965). The release of myoelectric signals—in voluntary contraction—is, via central and peripheral neurons, controlled by the brain. Advantage is taken of this fact when utilizing myoelectric signals for prosthesis control. This type of control will thus be experienced by the amputee as more natural than the conventional mechanical control (*McLaurin*, 1965). Furthermore, no extensive and disturbing movements of the body are demanded, less muscular force is required, and the risk of the patient becoming fatigued is reduced. A further important fact is that the patient will receive proprioceptive impulses from the muscles involved, giving a sensation of communication with the prosthesis (*Popov*, 1965; *Scott*, 1966).

In arm amputees and patients with impaired motion due to pareses of different origin, the myoelectric signals often differ in appearance. The frequency and degree of these changes in various muscles of arm amputees are discussed in this paper. Furthermore, the ability of arm amputees and normals to utilize their myoelectric signals as control impulses, and the electrode problems involved in picking up these signals, have been studied.

HISTORICAL REVIEW

Arm Amputees

Although the history of prostheses is closely connected with the years following the two great World Wars, the number of amputations in civilians considerably exceed that in disabled soldiers. As a result of the War 1941–1945, the number of arm and leg amputations in the USA was estimated to be 40,000, while for civilians the corresponding number a few years ago was 500,000 (*Alter*, 1966). In the same country the total number of recent amputations was approximately 35,000 per annum (*Russek*, 1961). The British Ministry of Health compiles all annual amputations, also including frequency, age, sex, and etiology. These reports are forwarded by the prostheses workshops. During a number of years this has enabled the relation between arm and leg amputees to be determined, the ratio being one to ten. At the same time an annually increased frequency of new amputations could be observed in Great Britain, in agreement with a series of leg amputations in Sweden examined by *Hansson* (1964).

In 1962, Great Britain thus registered approximately 300 fresh arm amputations. Based on the frequency of fitted prostheses, the rate of annual leg amputations in Sweden was recently estimated at 1000 cases (*Herberts*, 1968). The number of new arm amputations in that type of study is more difficult to determine, but the frequency of cases was assessed at 100 per annum, which also resulted in a relation of one to ten between arm- and leg amputees. The total number of arm amputees in Sweden was estimated at 900 patients by the Swedish Committee for Disabled in 1955, while the frequency at present could be estimated at 1500.

The difference in sex is obvious among arm amputees, inasmuch as the males are in the majority. Thus, *Watkins* & *Ford* (1962) in a follow-up of 150 arm amputees found five times more males than females. The male predominance among arm amputees was also noted by *Hansson* (1966) who in a study on 243 arm prosthesis applications found nine males to one female.

Contrary to leg amputations, the loss of an arm generally strikes young people. Several series reveal that most of the patients are younger than 50 years of age at the time of accident, the mean age ranging from 20 to 30 years (*Watkins* & *Ford*, 1962; *Hansson*, 1966).

Etiological studies of arm amputations show that most cases were caused by trauma. Thus in 60–84 per cent of the cases, trauma was the cause of the patient's handicap. Following trauma, congenital aplasia commonly resulted in amputation whereas malignant tumours amounted to less than 10 per cent as is shown in Table 1. The traumatic lesions were mostly due to industrial accidents. During 1962 in Great Britain, traumatic arm amputations were reported to be caused by industrial accidents in 60 per cent, whereas approximately 20 per cent were due to traffic accidents.

Table 1. Causes of amputation in two series.

	No. of cases	Trauma	Congenital defect	Tumour
Great Britain acc. to Brit. Ministr. of Health, 1962	296	60%	27%	6%
Gothenburg, Sweden, acc. to Hansson, 1966	74	72%	16%	8%

In principle, arms can be amputated at six functionally different levels. The level of amputation is decisive for the degree of loss of function and the resulting handicap. In exarticulation of the wrist, there generally remains a satisfactory pronation and supination ability, but there is lack of prehension. In exarticulation of the shoulder, of course no function whatever remains. The amputation levels found in patients on five different investigations are indicated in Table 2. It should be noted that *Müller* (1962) and *Nathan & Davidoff* (1965) and *Solonen* (1965) performed their investigations on disabled soldiers. In the two remaining series—representative of a country in peace—there was a significant predominance of below-elbow amputees. Extensive exarticulations of the shoulder joint are rare as these operations are mainly

Table 2. Levels of amputation in different investigations.

	Forequarter	Above elbow	Below elbow
Müller 1962	5	31	26
Watkins, Ford 1962	5	46	81
Nathan, Davidoff 1965	—	19	17
Solonen *et al.* 1965	—	24	48
Hansson 1967	—	16	52

performed on the indication of a tumour. As most of the arm amputations are the result of a trauma, it is comprehensible that the remaining muscles, nerves and other tissues may reveal changes in an amputated extremity. These changes are not merely localized to the stump area; there may also be severe pathological changes, proximally, in the injured extremity due to traction of the nerves and vessels at the time of the accident.

Myoelectric Control Systems

It was three British scientists, *Battye, Nightingale & Whillis* (1955) who again suggested that myoelectric signals should be possible to utilize for control of prostheses and who introduced a by now classic description of a myoelectric control system. They proved that below-elbow amputees could actively control prehension in a simple prosthesis by myoelectric signals produced in the stump muscles. Patients recently operated upon were considered to control the prosthesis with less difficulty than those earlier amputated. The new technique of the transistor made the work with electronic control systems more attractive than before. Disturbances in the myoelectric signals from the external surroundings, from the skin electrodes and from the amplifiers were pointed out. A few years later a myoelectrically controlled hand prosthesis was developed in the Soviet Union (*Kobrinskii et al.*, 1960). In both these prosthesis systems, skin electrodes were utilized to pick up the myoelectric signals. The Russians have applied their hand prosthesis in at least 1000 patients (*Lymark*, 1968) and it has also been sold on a licence for clinical application to Great Britain (*McKenzie*, 1965) and Canada (*Sherman*, 1964; *Gingras et al.*, 1966). This prosthesis lacks active pronation and supination and does not offer any proportional grip. It is easy to subscribe to the favourable points raised by *Scott* (1966) about the Russian electric hand having been released for clinical use and criticism which allowed valuable, practical experience to be gained. *Gingras et al.* (1966) found the technical design incomplete and modified several components on the basis of their experience.

In the beginning of the 1960's, a number of myoelectrically controlled hand prostheses were introduced (*Horn*, 1963; *Bottomley & Cowell*, 1964; *Hirsch, Kaiser & Petersén*, 1964; *Schmidl*, 1965; *Knowles, Stevens & Howe*, 1965). All these systems merely offered one function, prehension. Skin electrodes were used to pick up the myoelectric signals, and the remaining extensor and flexor muscles of the forearm stump were utilized as control sites. The brachial biceps and triceps muscles and other groups of muscles have also been suggested. No proportional control of the grip could be offered with these prosthetic systems. A few attempts have been made to obtain control of more functions than prehension (*Lyman, Groth & Weltman*, 1964). So far, nobody has managed to control three degrees of freedom (*Godden*, 1968, among others) despite many years' research activities, still going on in several laboratories.

Geddes, Moore, Spencer & Hoff (1959) introduced the use of myoelectrically controlled orthoses. Via myoelectric signals from a forearm muscle, a pneumatic orthosis, applied to a paralytic hand, could be controlled. Myoelectric control of electric orthoses has later been developed, utilizing signals from a paretic muscle (*Waring & Nickel*, 1965; *Antonelli & Waring*, 1967; *Waring & Antonelli*, 1967). The low level of myoelectric signals which can be picked up in severe paresis, demands a high sensitivity of the amplifier, in order to obtain a sufficient signal-to-noise ratio. In connection with orthoses, paretic muscles were electrically stimulated. This method was initially utilized by stimulation of the peroneal nerve in hemiplegics with a drop foot (*Liberson et al.*, 1961). Electric stimulation of paretic muscles in the upper extremities was later described which enabled a paretic hand to perform an active movement (*Long & Masciarelli*, 1963). In this case the paretic forearm extensors were electrically stimulated via skin electrodes and the hand splint, fitted to the patient, was closed by means of a spring. The patient's active control over the so-called electrophysiological hand splint was mechanical via remaining non-paretic muscles of the shoulder. Myoelectric control was later developed, where instead of mechanical control, the signals from intact shoulder muscles were utilized to control the hand splint (*Reswick & Vodovnik et al.*, 1964; *Vodovnik et al.*, 1965). The system was designed so that the amplitude of the myoelectric signals which were utilized as control signals, were proportional to the amplitude of the stimulating impulse which affected the paretic muscle. *Crochetiere, Vodovnik & Reswick* (1967) have shown that a rectified, rectangular stimulating impulse is the most favourable, and that the stimulating electrode should be large because of the muscular response to the stimulation. *Vodovnik, Crochetiere & Reswick* (1967) also reported that the position of an extremity could be determined as a function of the amplitudes of the stimulating impulses.

There are in principle two different methods of myoelectric control of prostheses, on-off or threshold control and proportional control. In the first case there will be an indication if the signal amplitude exceeds a certain threshold level which is sufficiently high to prevent, effectively, disturbances caused by noise, induction or resting potentials. Proportional control implies that the signal is satisfactorily reproducible, and that it will not be affected to a great extent by fatigue, hysteresis or artefacts. It has been pointed out that the intensity of the myoelectric signal, on the whole, varies continuously with the muscular tension (*Lippold*, 1952; *Bigland et Lippold*, 1954). Therefore, if the myoelectric signal is only utilized for on-off control of the prosthesis, a great deal of signal information will never be extracted. An important advance in prosthetic research was made when a prosthesis model was introduced which could utilize the myoelectric signals for proportional control of pneumatic as well as electric power supplies (*Bottomley*, 1962; *Bottomley, Kinnier Wilson & Nightingale*, 1963). *Bottomley & Cowell* (1964) reported a method to obtain proportional control of a myoelectrically controlled split-hook. Skin electrodes were used to receive myoelectric signals from the forearm extensors and flexors, and the difference signal (*Bottomley et al.*, 1963) between the electrodes was utilized to control both the speed of movement and the force of grip when the hook opened and closed. This was achieved by two feed-back signals, one of which depended on the speed and the other on the force of grip. The clinical observation on which this solution was based is that a hand at work either opens or closes without using any force, or grips with force without moving. The effect of the random variations on the myoelectric signals can be considerably reduced by applying a so-called "back-lash generator" (*Bottomley et al.*, 1963). This method permits a great variation in the myoelectric signals without reducing the ability to control the hand when loaded lightly. In Great Britain, the technical development of this hand has been transferred to the Atomic Weapons Research Establishment, but no clinical results of prosthetic applications on amputees have as yet been published.

In recent years, it has been stated that an improved hand function can be obtained by certain elements of the electric hand prostheses performing automatically. An adaptive grip is offered (*Tomovic*, 1961, 1962; *Rakic'*, 1964, 1968) by placing several pressure-sensitive transducers on the finger tips and palms of the hand which, when activated, transmit a feed-back signal to the prosthesis. The hand will then automatically perform a number of standard movements depending on the transducers involved. This hand, according to the latest modification, can instead of the conventional, stiff three-finger grip perform a forceful grip towards the palm of the hand, hook grip and clenching of the hand (*Lymark*, 1968). Another type of feed-back from the prosthesis to the carrier has been described where a pressure-sensitive crystal situated on the prosthesis thumb gives an electric signal when deformed (*Beeker, During & den Hertog*, 1967). The purpose of this signal was to achieve cutaneous stimulation in one part of the body via surface electrodes. At present the advantage of the transmitted sensibility is being tested on a number of amputees. An extensive investigation of different methods to offer the amputee communication with the prosthesis via the tactile senses was recently published (*Alles*, 1968).

One of the primary problems arising in the development of myoelectric control systems is, no doubt, the lack of muscle control sites which has also been emphasized by *Groth, Weltman & Lyman* (1962). The lack of sites, i.e. the fact that there are few places in the body where a control signal is available, reduces the possibility for the patient to communicate with the orthosis or prosthesis. It has been mentioned (*Bottomley*, 1965; *Scott* 1966, 1967) that to those, who are severely handicapped, unilateral or bilateral amputees, myoelectric control will be of the greatest advantage. The paradoxical situation then arises that the patients who are in the greatest need of a prosthesis with several degrees of freedom have a less number of well conducted muscle control sites available. It is obvious that the advantage of the normal hand as a gripping tool depends on its ability to perform in an adequate position (*Capener*, 1960). A prosthesis with several degrees of freedom would therefore be desirable. The myoelectrically controlled hand prostheses, available so far, offer one function only—prehension—and no proportional control. By picking up three amplitude levels in the contraction of one separate muscle, two control sites will be available in one muscle (*Dorcas* & *Scott*, 1966) and the demand of multiple control sites will be reduced. Only a little practice allows separate control of both functions, opening and closing of the hand. *Scott* (1966) mentioned that sufficient signals to control this three-state-unit can be picked up from partially denervated and atrophic muscles in amputees. Due to the lack of a description on the pathologically disturbed myoelectric signals, the magnitude of these changes cannot be assessed.

Programme-coding of prosthetic function is another form of reducing the number of control sites for the control of complicated prostheses and orthoses. The prosthesis will then have certain automatic movements. A sequential coding equals a series of movements released by one single control site, which technically can act very naturally (*Maling* & *Clarkson*, 1963). This system, however, is slow, and requires continual attention and cannot be unconsciously controlled by the patient. A better system is parallel-coding where a number of control sites are activated in various pattern combinations synchronously. By comparing control signals, various codes in a pattern detector can be combined, and the prosthesis can operate in several degrees of freedom simultaneously (*Michael* & *Crawford*, 1963; *Harrison*, 1964). This parallel-coding will increase the number of functions available from a given number of muscles. Signals from strain gauges are sometimes used to indicate a muscle contraction instead of myoelectric signals (*Lyman*, *Groth* & *Weltman*, 1964). Attempts were made to obtain control of an arm simulator with three degrees of freedom (*Lucaccini*, *Freedy*, *Rey* & *Lyman*, 1967).

Muscles which normally do not have any specific function, for instance, the auricular muscles can be trained to give voluntary contractions (*Bontrager*, 1965). It is doubtful whether this type of myoelectric control will turn out to be of any practical value, since too much effort is demanded of the patient.

The fact that the human being can control single motor units voluntarily in a muscle (*Harrison* & *Mortensen*, 1962; *Basmajian*, 1963) was believed to offer a solution to the control problem (*Bottomley*, 1965; *Scott*, 1966). Audial and visual feed-back, after training, permits normal subjects to control up to three motor units in one muscle, in a few cases even more (*Basmajian, Baeza* & *Fabrigar*, 1965). Certain subjects can maintain the control over single motor units even without artificial feed-back. The total number of units which can be activated, irrespective of each other, has not yet been determined. Using this method, it is possible that several control sites can be obtained from a single muscle to control the prosthesis. *Harrison* & *Mortensen* (1962) believe that the frequency of the single motor unit activity also can be controlled voluntarily. *Wagman* & *Pierce* (1966) pointed out that with the technique described, the remaining muscle is relaxed when the single motor unit is activated. No attention has been paid to the extent of which the single motor unit has participated in the total muscle contraction. Recently it has been pointed out, however, that subjects can be trained to maintain control over, and alter the frequency of single motor units in a muscle even when certain disturbing movements occur in the extremity which is actuated by the muscle involved (*Basmajian* & *Simard*, 1967).

Electromyography

During the middle of the 18th century the first reports were given of muscle contractions, evoked by static electricity stimulation (*Kratzenstein*, 1746). The most renowned investigation showing the connection between electricity and muscle contractions was a series of studies on frog muscle (*Galvani*, 1791). The "animal electricity" was not believed by *Galvani* to originate from muscles but from nervous tissue, particularly the brain. It was not until 1838 that it was shown that electric current really was developed in muscles (*Matteucci*, 1844). The first to pick up electric current in active muscle contraction in man, the first human electromyogram, was *DuBois-Reymond* in 1851 (*Licht*, 1961). In the beginning of the 20th century, improved methods were introduced, inter alia, by means of a string galvanometer (*Einthoven*, 1901) which permitted rapid and sure recording of the weak electric activity accompanying muscle contraction.

Electromyography (EMG), the recording of an electric phenomenon in striated muscles, was then employed mainly by physiologists. The first to apply this technique on pathological cases was *Proebster*, who in 1928 studied a number of peripheral neuron lesions. At about the same time *Adrian* & *Bronk* (1929) introduced a concentric needle electrode to be inserted into the muscle. During the following years, clinical electromyography was developed as a diagnostic aid in the studies of pareses and atrophy both of muscular (*Lindsley*, 1935; *Kugelberg*, 1947, 1949) and neurogenic origin (*Denny-Brown* & *Penny-*

backer, 1938; *Buchthal & Clemmesen*, 1941). Several compilations of the EMG changes, found by employing electromyography in different pathological conditions, have been made (*Wedell, Feinstein & Pattle*, 1944; *Buchthal & Pinelli*, 1952). With the increased information gained by electromyography this method has now become a clinical routine. Furthermore, during recent decades, electromyography has come into use for anatomical and kinesiological studies of the muscular function (*Basmajian*, 1967). Several surveys have been published (*Kugelberg*, 1953, 1959; *Isch*, 1963; *Buchthal*, 1957, 1960, 1961; *Licht*, 1961; *Basmajian*, 1967).

The Motor Unit

In the muscle the structural unit is constituted by the muscle cell or the muscle fibre. The human muscle fibre varies considerably in length in different muscles, the mean value being about 50 mm (*Lindhard*, 1931). There is also a difference in diameter of the muscle fibres ranging from 0.03–0.1 mm, with a smaller diameter in small muscles such as the external ocular muscles. A number of muscle fibres are innervated by the same motor nerve fibre or axon, originating from one anterior horn cell of the spinal cord. This nerve cell with its axon and terminal branches as well as the innervated muscle fibres form an anatomical and functional unit, the motor unit (*Sherrington*, 1925). The nerve fibres divide close to the muscle fibres and do not spread over the entire muscle, but the contact between the nerve fibre and the single muscle fibre occurs within the so-called endplate zone in the middle area of the muscle fibre (*Cöers*, 1953; *Cöers & Woolf*, 1959). This endplate zone is well demarcated and constitutes less than 10 per cent of the total length of the muscle (*Buchthal, Guld & Rosenfalck*, 1955). The number of muscle fibres per motor unit, differs considerably in each muscle. Muscles controlling precise and rapid movements, generally have the smallest number of muscle fibres per motor unit. Thus, in man there are ten fibres per unit in the tensor tympani muscle (*Wersäll*, 1958) as opposed to 1600 in the medial head of the gastrocnemius muscle (*Feinstein, Lindegård, Nyman & Wohlfart*, 1955). The remaining muscles range between these extreme values.

The muscle fibres belonging to one motor unit are dispersed over a certain area and mingled with fibres connected with other motor units, demonstrated by means of a multi-electrode technique elaborated by *Buchthal, Guld & Rosenfalck* (1957). Fibres, relating to the same motor unit, have been demonstrated in circular areas with a diameter of 5–7 mm in the upper extremities and 7–10 mm in the lower extremities. These areas allow space for muscle fibres from 15–30 different motor units (*Buchthal, Erminio & Rosenfalck*, 1959). *Edström & Kugelberg* (1968) recently reported that the fibres of an individual motor unit are dispersed within an area constituting 17 per cent of the sectional area of the rat muscle, with a considerable overlapping between different units. When a nerve impulse reaches the motor endplate, where the terminal axon fibres are close to the separate muscle fibres, the cell membrane of the muscle fibre will be gradually depolarized. At a certain level the action potential becomes released as a result of the instantaneous change of the membrane potential. This depolarization continues along the muscle fibre at a rate of 4–5.5 m/s in the human brachial biceps muscle according to *Buchthal, Guld & Rosenfalck* (1955), 3–4 m/s according to *Stålberg* (1966). The speed of dispersion along the muscle fibre becomes reduced in ischaemia and at a low intramuscular temperature (*Stålberg*, 1966). Following the depolarization of the muscle fibre membrane, the muscle fibre contracts. The problem of the relation between excitation and contraction, however, is still not solved. The brief contraction of the individual muscle fibre lasts from 1 to 2 msec and during the same time an electric potential is spread in the surrounding tissues as a result of the depolarization. The existence of different motoneurons with different functions, so-called phasic and tonic motoneurons, was demonstrated on cat (*Granit, Henatsch & Steg*, 1956; *Eccles, Eccles & Lundberg*, 1958). By means of histo-chemical methods, it has recently been proved on rat that with respect to function there are different muscle fibres and motor units (*Kugelberg & Edström*, 1968; *Edström & Kugelberg*, 1968).

Motor Unit Action Potential Parameters

The purpose of diagnostic electromyography is to determine exact, quantifiable and reproducible criteria of the myoelectric activity of the muscles in various lesions, metabolic disturbances and diseases. For this, quantitative measurements of various parameters of the individual motor unit potential have mainly been used, such as duration, shape and amplitude. The parameter which is considered as the most favourable and also offers the best information in pathological conditions of the muscles, is the duration of the motor unit potential. The duration of the motor unit potential is longer than that of the individual muscle fibre (*Buchthal, Guld & Rosenfalck*, 1954). This depends on the muscle fibres of the same motor unit having a certain territorial dispersion and an almost simultaneous activation, which results in a temporary dispersion. The electric potential which can be picked up from the motor unit, has a duration of 7–8 msec in the human extremity muscles (*Buchthal & Clemmesen*, 1941; *Petersén & Kugelberg*, 1949) and an amplitude between 0.1–0.5 mV. During latter years, duration measurements, according to stipulated criteria (*Buchthal & Rosenfalck*, 1955) resulted in higher values; thus, in the human brachial biceps, a duration of 9–10 msec was reported (*Kaiser &

Petersén, 1965). The duration of the motor unit potential varies considerably within different muscles. Muscles with fewer fibres per motor unit, for instance, the facial muscle, have a decidedly shorter duration than muscles with more fibres per motor unit (*Petersén* & *Kugelberg*, 1949). This can partly be explained by the reduced temporal dispersion which is found with a less number of fibres per unit (*Buchthal* & *Rosenfalck*, 1955). The density of the fibres is of importance for the duration of the motor unit potential. The duration of the motor unit potential depends on the type of electrodes used, which must be indicated and standardized (*Petersén* & *Kugelberg*, 1949). With increasing age, the mean duration of the potential will become prolonged (*Petersén* & *Kugelberg*, 1949; *Buchthal* & *Pinelli*, 1952). There will also be a prolongation of the mean potential duration when the intramuscular temperature drops (*Bentsen*, 1945; *Buchthal* & *Pinelli*, 1952). The same muscle reveals a deviating duration of various motor unit potentials as a result of different distances between the electrode and the endplate zone (*Buchthal, Guld* & *Rosenfalck*, 1954; *Kaiser* & *Petersén*, 1965). A shorter duration has been indicated for the long head of the biceps muscle as compared to its short head (*Kaiser* & *Petersén*, 1963, 1965). These authors stated that the mean duration is slightly longer for males than females.

The individual motor unit potential may appear in a mono-, bi- or tri-phasic form, depending on the type of electrode. Due to the scattering of individual fibres in the motor unit, poly-phasic potentials also appear and an increased spatial and temporal dispersion causes an increase in the number of poly-phasic potentials (*Buchthal*, 1960). In muscles where the motor unit potential has a long duration, there are thus more poly-phasic potentials than in muscles with a short motor unit potential (*Buchthal, Guld* & *Rosenfalck*, 1954). The number of poly-phasic potentials also increases pronouncedly at a low intramuscular temperature and also in connection with fatigue (*Buchthal, Pinelli* & *Rosenfalck*, 1954).

The third parameter, studied in the motor unit potential, is the amplitude. The amplitude is dependent on the fibre density and the number of muscle fibres of each motor unit, and will increase linearly with the territorial magnitude of the motor unit (*Buchthal, Erminio* & *Rosenfalck*, 1959). The amplitude varies noticeably with the types of electrodes and the depths of insertion into the muscles, *i.e.* the distance to the active muscle fibres. On the other hand, it has been stated that the muscle temperature only has an insignificant effect and that fatigue does not at all affect the potential amplitude (*Buchthal, Pinelli* & *Rosenfalck*, 1954).

In a relaxed, normal muscle no electric activity can be demonstrated (*Adrian* & *Bronk*, 1929; *Lindsley*, 1935). When a needle electrode is inserted into a normal, relaxed muscle, a repetitive insertion activity of a short duration is obtained. This activity, probably caused by mechanical irritation, generally appears in showers (*Buchthal* & *Clemmesen*, 1941) and sometimes can be observed during several minutes (*Petersén* & *Kugelberg*, 1949). When the muscle contraction is weak, individual motor units are activated at the lowest possible frequency of 5–10 per sec. In higher muscle contraction, more and more units are gradually activated and the discharge frequency of the individual unit increases in man up to approximately 50 per sec (*Adrian* & *Bronk*, 1929). An average maximum frequency has later been indicated at 20 per sec (*Wedell, Feinstein* & *Pattle*, 1944) and 25 per sec (*Basmajian et al.*, 1965). As gradually more and more units are added, a complex pattern is obtained and the individual motor unit potentials can no longer be identified.

Pathological Changes in EMG

Atrophy of the muscles and pareses of varying degrees may be caused by disease and damage affecting the lower motoneuron or the muscle. Both in neurogenic and myogenic pareses there are characteristic changes in the electromyogram, with respect to the individual motor unit potential and the signal pattern on higher levels of contraction. This also applies to the appearance of spontaneous electric activity of the muscle. In peripheral motoneuron lesions with muscle denervation, spontaneous, fibrillar action potentials can be recorded from the relaxed muscle about 2–3 weeks following the lesion (*Denny-Brown* & *Pennybacker*, 1938). These potentials have a short duration of 1–2 msec and are considered to reflect the activity of individual muscle fibres. They appear at a low frequency. The fibrillation potentials, however, occasionally can be observed in non-denervated muscles. Consequently, the finding of fibrillar action potentials must be carefully assessed in connection with the diagnosis. Another type of potential appearing in relaxed, denervated muscles, is repetitive with an introductory, so-called positive phase and a duration of 4–8 msec (*Kugelberg* & *Petersén*, 1949; *Jasper* & *Ballem*, 1949). In affection of the anterior horn cells of the spinal cord, large, poly-phasic potentials appear, corresponding to the fasciculation which can be seen with the naked eye (*Pinelli* & *Buchthal*, 1951). In peripheral neuron lesions the number of activated motor units can be reduced at a maximum muscle contraction (*Buchthal*, 1957). Often the activity does not fill the baseline and sometimes only single units are recorded. Furthermore, in about 50 per cent of the cases the duration becomes prolonged (*Buchthal* & *Clemmesen*, 1941) and usually there is an increase in the amplitude of the motor unit potential (*Denny-Brown* & *Pennybacker*, 1938). This can partly be explained by the regenerating nerve fibres not only contacting muscle fibres within their own motor unit but also others (*van Harreveld*, 1945), and by an increased synchronization (*Buchthal* & *Höncke*, 1944). An early sign of re-innervation may be the cessation of the fibrillation potentials or the appearance of poly-phasic potentials (*Wedell, Feinstein* & *Pattle*, 1944; *Jasper*, 1946).

In diseases of a muscle, myogenic atrophy or pareses, myosites or metabolic, toxic and degenerative lesions of the muscles, the motor unit becomes changed by a gradual destruction of individual muscle fibres, while the number of motor units remains intact (*Kugelberg*, 1947, 1949). When the level of muscle contraction is high the number of motor unit potentials picked up is large in proportion to the pareses, and fills the base-line. This is characteristic of advanced muscle dystrophy and is different from neurogenic pareses, since in paresis of a corresponding degree, a reduced number of motor units will be activated at maximum contraction. The duration of the individual motor unit potential is often greatly reduced in myogenic pareses and, moreover, there is an increased number of poly-phasic potentials (*Kugelberg*, 1947, 1949; *Buchthal* & *Pinelli*, 1952; *Pinelli* & *Buchthal*, 1953 *a*).

Frequency Analysis

Detailed investigations of the individual motor unit potential, including duration measurements, require that the level of contraction in the muscle be low. In this case a few, well isolated potentials can be picked up. As the level of contraction rises, an increasingly complex signal is recorded which unables further study of inidividual potentials. The question then arises whether measurements of isolated potentials in connection with weak contractions will offer values representative of all motor units. It is thus desirable to analyse the myoelectric signal for high levels of contraction also.

Piper (1907) was the first to apply frequency analysis to the study of myoelectric signals in voluntary contraction. He found that the signal amplitude varies regularly at a frequency of 47 to 50 Hz, the so-called Piper rhythm. It was not until 1951 that the relation of various components in the power spectrum of the myoelectric signal was studied by a simple method (*Richardson*, 1951). *Richardson* determined the relation between power spectrum components above and below 400 Hz. Increased signal activity at high frequencies was believed to indicate myopathy. A frequency analyzer was first used to describe the power spectrum in detail, both in normal and pathological cases, by *Walton* (1952). The analysis of normals revealed a predominating activity between 100 and 250 Hz, which gradually declined with increasing frequency and essentially vanished at 800 Hz. It was only in the facial muscle and the small hand muscles that frequencies up to 1250 Hz could be observed. In myopathy the predominating activity of the myoelectric signal was displaced towards higher frequencies. Peripheral neuron lesions, on the other hand, caused no changes in the power spectrum. These results have later been confirmed (*Fex & Krakau*, 1957).

Neurogenic and myogenic atrophy were believed by these authors to change the shape of the power spectrum, particularly involving frequency displacements of the spectrum peak. A certain displacement towards higher frequencies could be observed in inactivation atrophy. This is in accordance with the fact that the duration of the individual motor unit potential is shorter in these cases (*Buchthal & Pinelli*, 1952). Contrary to earlier investigations, it has later been proved (*Kopeć & Hausman-Petrusewicz*, 1966) that in peripheral neuron lesions also the power spectrum of the myoelectric signals alters. Thus, the spectrum will be more compressed, and will decline rapidly both at low, and—even more pronouncedly—at high frequencies.

The power spectrum of the myoelectric signal depends on the choice of electrode. The activity picked up with needle electrodes contains components of higher frequencies than that obtained with skin electrodes. This is partly due to the change imposed on the motor unit potential as it is transmitted through the tissues and picked up by a large electrode (*Fex & Krakau*, 1958). The power spectrum is also dependent on the positioning of the skin electrodes, the distance between the electrodes, and the angle between the direction of the electrode and the muscle fibres (*Sato*, 1964). As the electric fields in the various motor units differ in amplitude, phase, repetitive frequency, and polarity, the individual motor unit potentials will add randomly at the electrode (*Hayes*, 1960). At higher levels of contraction the myoelectric signal therefore assumes the essential characters of noise. When the levels of contraction rise, the motor unit potentials become synchronized, which results in increased potential duration and amplitude (*Krakau*, 1956). The low frequency components of the power spectrum are thus augmented. This synchronization, which has been described earlier (*Buchthal & Clemmesen*, 1941) probably is the origin of the Piper rhythm. It has been stated that the predominating frequency of the power spectrum is different for different muscles (*Sato & Tsuruma*, 1967). These authors found the predominating frequency range of the human brachial biceps muscle to be 50 to 60 Hz, as opposed to that of the gastrocnemius muscle which was stated to be 100 to 200 Hz. Other authors have found the power spectra of the myoelectric signals in various muscles to be rather similar (*Fex & Krakau*, 1958). In Paper I (*Herberts, Kaiser, Magnusson & Petersén*, 1969), however, a great number of statistically significant intermuscular differences have been demonstrated. Changes in shape of the low-frequency portion of the power spectrum due to changes in contraction level have also been observed (*Kaiser & Petersén*, 1963). *Kaiser & Petersén* (1963) were the first to find that the signal activity at frequencies exceeding 1000 Hz declined during a constant maximum contraction. They also introduced a method of describing the profile of the spectrum with points in a frequency coordinate system, *i.e.* loci. Six filters suffice to give an accurate description of the frequency profile.

Several investigations have been made dealing with the influence of fatigue on the appearance of the normal myoelectric signal. It has been pointed out that the signal pattern in isometric contraction probably deviates from that in ergometric work (*Scherrer & Bourguignon*, 1959). In forceful isometric muscle contraction, the myoelectric signal alters as a function of time during a prolonged contraction. The fact that prolonged muscle contraction reduces the amplitude of the motor unit potentials was mentioned as early as 30 years ago (*Seyffarth*, 1940). Since then, skin electrodes have been used to demonstrate that a fatiguing muscle contraction is accompanied by an increase in the myoelectric amplitude, due to synchronization of the potentials. Consequently, there will be a displacement of the power spectrum towards low frequencies (*Sherrer & Bourguignon*, 1959). Frequency analysis also revealed that the amplitude-increase in connection with muscle fatigue was concentrated to the low-frequency part of the spectrum (*Kogi & Hakamada*, 1962). These changes picked up by skin electrodes were considered to be caused by the synchronization of motor units. The existence of synchronization has been proved by means of cross-correlation analysis (*Person & Kudina*, 1968). An extensive investigation of dynamic changes in the power spectrum of the myoelectric signal during fatiguing muscle contractions was recently published by *Kadefors, Kaiser & Petersén*, (1968). During an isometric-isotonic fatiguing contraction of the brachial biceps muscle, there is a significant increase of the total myoelectric signal activity. The spectral changes are increased activity within the low frequency band, and reduced activity within the high frequency band. These effects do not occur simultaneously, and are probably of different origin. Synchronization and activation of further motor units may explain the total increase in the signal activity, and a drop-out of particular motor units may explain the reduced activity within the high-frequency range. Analyses of high-speed registrations of the myoelectric signal support this theory. These findings are of particular interest in conjunction with the previously mentioned possibilities of separating functionally different kinds of motor units.

Electrodes

Bioelectrical electrodes of various types are used for the following three main purposes: impedance measurement, bioelectric signal pick-up, and stimulation. In picking up myoelectric signals, there are three methods of placing the electrodes: on the skin, percutaneously, and subcutaneoulsy. The electrodes employed are correspondingly called skin electrodes, percutaneous and implantable electrodes. In all myoelectric control systems so far introduced for clinical use, skin electrodes are employed, and have been believed to be the only practical solution (*Alter*, 1966).

There are two classes of skin electrodes: wet and dry electrodes. The dry ones consist of a metal plate fitted to the skin, whereas the wet ones are supplied with an electrolytic layer between the skin and the metal plate. This classification is not quite adequate, since the dry electrodes also become covered with a thin electrolytic layer when they are applied to the skin. One disadvantage, which is more pronounced in dry than wet skin electrodes, is the high level of impedance, about 100 kohm. Thus, a considerably higher input impedance—about 100 Mohm—is required of the amplifiers to prevent the attenuation and distortion of the myoelectric signal from exceeding permissible values (*Geddes, Baker & McGoodwin*, 1967). These high impedances may evoke interferences from the surroundings (*Dorcas, Libbey & Scott*, 1966). Impedance differences of the two signal electrodes may cause the amplifier to be charged with disturbing signals, if its input impedance is not sufficiently high. Small movements between the skin electrode and the electrolytic layer may cause disturbances in the form of polarization potentials. This effect is particularly pronounced in the case of the formerly frequently used stainless steel electrodes (*Lykken*, 1959). Chloridized silver electrodes and a paste containing chloride ions, which permit ionic equilibrium, give a stable impedance and reduces the motion artefacts (*Day & Lippit*, 1964; *Geddes & Baker*, 1967; *Pacela*, 1967). A method of reducing motion-induced potential disturbances is to arrange a stable fixation between the electrode and the skin by means of a special paste container (*Boter, den Hertog & Kupier*, 1966). The motion artefacts can also be eliminated by a conductive polymer bridge between the skin and the electrode (*Thompson & Patterson*, 1963). One way of reducing the impedance between the tissue and the electrode—with a less disturbed signal as a result—is, besides using electrolytic paste, to rub the horny layer of the skin with sand-paper (*Nightingale*, 1959). However, this is impossible if the electrodes are used daily in a prosthesis system. The electrolytic paste used with wet electrodes may cause problems due to allergic and toxic skin reactions. It is not probable, as *Bottomley* (1965) has pointed out, that the skin of the amputation stump could stand this for any length of time. Another problem connected with skin electrodes is that they pick up signals from a rather large area, and thus activity from neighbouring muscles also. It has been stated that when two antagonistic muscles are utilized for separate control functions, signals produced by one of the muscles should also be picked up by the electrode belonging to the other muscle (*Scott*, 1967), resulting in so-called cross-talk (*Bottomley et al.*, 1963). It has been a subject of discussion whether this cross-talk actually does or does not consist of simultaneous interference activity of the antagonist.

In clinical electromyography, needle electrodes of various types are routinely used to derive the myoelectric signal intramuscularly. The use of these electrodes may cause some discomfort, and a risk of infection may arise when they are applied during long periods. For these reasons, needle electrodes cannot be used in prosthetic control. A thin wire, percutaneously run into the muscle, is a new type of electrode which has many advantages (*Scott*, 1965; *Caldwell*, 1967). Several methods of inserting these wire electrodes securely have been reported (*Scott*, 1965; *Parker*, 1968). Problems due to the segregating properties of the skin thus become eliminated but, on the other hand, activity only from a limited number of motor units in the muscle will be picked up. The range of muscle contraction available for recording will, thus, be reduced (*Bigland* & *Lippold*, 1954). Another problem is the measures which regularly must be taken to prevent infection at the site of puncture, still another is breakage of the wires. These problems were discussed by *Scott* (1966), who also suggested methods of reducing the complications. The helical percutaneous electrode, designed by *Caldwell* (1967), reduces the risk of wire breakage. This electrode offers the possibility of a more stable position, as compared to other electrodes in recording how patients can learn, by training, to control single motor units in a muscle (*Wagman* & *Pierce*, 1966; *Caldwell*, 1967). It has been stated that this type of training will provide amputees and disabled patients with greater possibilities of myoelectric control, and thus render more sophisticated prostheses and orthoses feasible (*Basmajian*, 1967).

In so-called implant biotelemetry, biologic information from a part inside a living organism is wirelessly transmitted to the outside. Since 1954, when the transistor became commercially available, and the development of microelectronics rapidly advanced, biotelemetry has become an important tool in biological studies (*Caceres*, *Cooper* & *Mackay*, 1965). Implantable electrodes for the telemetry of myoelectric signals through the skin—by means of a frequency modulated wave detected by an external receiver—have been developed (*Ko*, 1964; *Hirsch*, *Kaiser* & *Petersén*, 1966). The main difference between these units is that *Ko* uses batteries as the source of energy whereas *Hirsch et al.* utilize externally generated inductive energy. Batteries have the drawbacks of being difficult to miniaturize and having short life. They must be exchanged at regular intervals which involve the disadvantage of repeated surgery. It has been stated, however, that an implanted battery can be inductively generated. This method allows a life of up to three years in a prosthesis system (*Vodovnik* & *Greene*, 1967). Important demands on an implanted electrode are tissue compatibility, stability and capability of providing reproducible signals. It should also be possible to sterilize and miniaturize the electrode to such an extent that it will not affect the normal function of the muscle (*Ko & Neuman*, 1967). Initial experiments on animals revealed that a capsule of varying thickness was formed around the implant, and that infections and implant flating could result (*Grotz, Long, Yon & Ko*, 1965). At the first implantations on man (*Hirsch, Kaiser & Petersén*, 1966), myoelectric signals containing more high-frequency activity than those picked up with skin electrodes were obtained. The advantages of this type of signal had been stated previously (*Hirsch, Kaiser & Petersén*, 1964). The advantages of implanted electrodes as compared to other electrodes may result in: isolated muscle activity with reduced cross-talk or interference, a more favourable signal-to-noise ratio and a broader signal power spectrum. Recently, several implantable electrodes for the telemetry of myoelectric signals have been designed. Implantation experiments, primarily on animals, are planned (*Reilly*, 1968), and some results have already been published (*Scott et al.* 1968; *Tucker & Scott*, 1968).

The Information Content of Myoelectric Signals

As mentioned before, myoelectric prostheses or orthoses can be operated in two ways, by on-off or threshold control, and by proportional control. The latter method requires that the salient signal parameter is reproducible and does not become affected by fatigue or artefacts. *Scott* (1966), inter alia, discussed which property of the myoelectric signal should be selected to control the external system. The signal parameter chosen should be easy to utilize by the patient. As muscle tension can be controlled voluntarily, it was natural to seek a parameter direct correlated to the developed muscle tension.

Irrespective of the choice of electrodes, the myoelectric signal appearing at moderate levels of contraction reveals a complex pattern, in which each motor unit is responsible for certain repeated activities. The character of the myoelectric signal accordingly is quite irregular, and can only be described statistically (*Basmajian*, 1967). Consequently, several problems are involved if a single parameter of the myoelectric pattern, related to total muscle tension, is to be utilized for proportional control. Two separate parameters reveal the degree of muscle tension, *viz.* the amplitude of the myoelectric signal, and the firing frequency of the individual units. The most common description of a myoelectric signal, so far, has been to state the so-called integrated mean value, or root-mean-square (r.m.s.) value, of the amplitude. This value is obtained by amplifying, rectifying and filtering the myoelectric signal. This signal processing is considered to yield a linear relation between signal amplitude and muscle tension in isometric contraction up to the maximum level of contraction (*Lippold*, 1952; *Lenman*, 1959). It has been discussed whether the relation is linear at very small and high loads (*Nightingale*, 1960). A large time constant of the filter will give a more even relation between the signal amplitude and the muscle tension. On the other hand, in a myoelectric prosthesis system with feed-back, a large time constant will cause undesirable oscillations in the prosthesis (*Bottomley, Kinnier Wilson* & *Nightingale*, 1963). The time constant must be selected so that the delay introduced into the system will not seriously affect the patient's ability to control the prosthesis (*Bottomley* & *Cowell*, 1964). The relation between integrated signal amplitude and muscle tension in the upper extremities is not quite linear according to these authors. However, they consider this to be of no importance when the signal amplitude is utilized as a control parameter.

In isotonic contractions, the constant ratio between r.m.s. signal activity and tension is dependent on both the length of the muscle and the speed at which the muscle length changes, as well as the direction of the change (*Bigland* & *Lippold*, 1954). At constant speed the muscle tension increases linearly with the signal activity, which in turn is more pronounced in a shortening than a lengthening of the muscle. In constant tension there is a linear relation between the signal activity and the speed in the shortening of the muscle, but not in its lengthening. Muscle tension, speed and signal activity are thus related to each other. A difficulty arising when stump muscles in amputees are to be used as control sites, is that these muscles are not distally anchored. The muscles will thus be unloaded during contraction, and the speed cannot be assessed (*Alter*, 1966).

When muscle contraction increases, more motor units become activated, and the single unit fires at a gradually higher frequency (*Basmajian, Baeza* & *Fabrigar*, 1965). The myoelectric signal can thus be described by measuring the frequency of the motor unit potential spikes in the signal pattern. There is also a relation between the frequency of spikes and the muscle tension (*Close, Nickel* & *Todd*, 1960). The r.m.s. signal amplitude is proportional to the square root of the frequency of spikes (*Basmajian*, 1967) as long as fatigue or synchronization of motor unit potentials do not appear (*Bergström*, 1959). The calculation of spikes has successfully been tried on normal subjects, and *Scott* (1966) considered spike frequency to be a suitable control parameter in a myoelectric system.

In the rectified and filtered myoelectric signal there remain, due to the statistical character of the signal, variations in the mean value of the amplitude. These variations are suitably described as a disturbing noise. At low levels of contraction the noise is low, but it increases gradually with the mean value of the signal amplitude. However, the amplitude increases more rapidly with the result that the signal-to-noise ratio increases with an increased level of contraction. Not only the myoelectric signal but also back-ground noise, which depends on several factors, is picked up. Tissues and electrodes cause noise contributions, the magnitudes of which depend on the respective impedances. There are also polarization potentials and motion-induced potentials between the electrode and the skin (*Nightingale*, 1959). Frequency analysis of this complex noise has proved it to be mainly confined to low frequencies (*Nightingale*, 1959; *Hayes*, 1960).

When the amplitude of the myoelectric signal is to be used as a proportional control parameter, it is necessary to determine the number of clearly reproducible signal levels, which can be activated irrespectively of each other (*Vodovnik* & *Kreifeldt*, 1967). The number of levels depends on the level of contraction, as discussed above, and on the filtering of the myoelectric signals. At a very early stage it was pointed out that myoelectric signals for prosthesis control had to be filtered in order to reduce the disturbances caused by noise from electrodes and amplifiers (*Battye*, *Nightingale* & *Whillis*, 1955). These authors suggested 100–1000 Hz as a suitable range of frequency for the control signal. This range was later also recommended by *Horn* (1963). Below 100 Hz the background noise is too loud to be accepted. As the signal power spectrum decreases at higher frequencies, deteriorating the signal-to-noise ratio, the upper frequency limit in filtering the myoelectric signal was later reduced to 800 Hz by *Bottomley*, *Kinnier Wilson* & *Nightingale* (1963). The frequency range 300–1000 Hz has also been recommended, with the object of excluding the fluctuations of the myoelectric signal believed to exist within the lower frequency ranges (*Hirsch*, *Kaiser* & *Petersén*, 1964). Further knowledge is required to arrive at final recommendations on how myoelectric signals should be filtered. Thus, it is necessary to learn the relation between developed mechanical force and various parts of the myoelectric signal spectrum.

Essentially our knowledge about the information content of the myoelectric signal is based upon normal subjects. In amputees and in patients with neurological disturbances there are deviations from the normal physiology which may be very important, as has been pointed out by *Wagman* (1966). Thus, there may be deviations in the nervous conduction rate, in the appearance of motor unit potentials, in the power spectra—static and dynamic—and in the effects of temperature deviations also.

SCOPE OF THE INVESTIGATIONS

The purpose of the present investigation has been to characterize myoelectric signals in order to create a basis for their utilization in prosthesis control. Myoelectric signals were analysed by employing previously introduced methods, and methods especially developed for this purpose. These studies included clinical examination, conventional EMG, analysis of myoelectric power spectra and analysis of the muscle control ability. Furthermore, various types of electrodes were studied. The attempt was to obtain a detailed characterization of myoelectric signals in normal and arm amputated males. The characteristics of the myoelectric signal thus obtained, would increase the possibilities of optimizing the signal by suitable filtration.

1. In the stump muscles and the proximally situated muscles of arm and leg amputees there are frequently pathological changes in the appearance of the myoelectric signal as determined by conventional EMG (*Blom & Hagbarth*, 1964; *Petersén*, 1966). It was therefore necessary to carry out further studies on the appearance of the signals in a series of amputees.
2. The myoelectric signal must be filtered to obtain an undisturbed myoelectric prosthesis control, a fact pointed out earlier in the review of literature. Information of the shape of the myoelectric power spectra must be available for all the different muscles in normals and arm amputees.
3. Dynamic changes of the spectrum in fatiguing isometric muscle contractions were studied previously in normals (*Kadefors, Kaiser & Petersén*, 1968) but must also be determined in arm amputees.
4. The relative properties of three different types of skin electrodes, needle electrodes and a refined type of implantable electrode were studied.
5. Physiological deviations, for instance, in speed and precision, must be borne in mind when a muscle is intended to be used as a control site in prosthesis control. Furthermore, the question arises whether an arm amputee will be able to use his stump muscle in performing rapid and well quantified muscle contractions. This ability may depend on whether the remaining muscles are normal or pathological according to the EMG findings. On the basis of these series of questions, an analysis of the muscle control ability was performed both on normal and arm amputated males, by estimating their skill in manoeuvering an electro-mechanical apparatus by myoelectric signals.

METHODS

Electromyographic Methods

The electromyographic investigations were carried out by means of a 3-channelled DISA Electromyograph (Type 13 A 69). For the tests coaxial needle electrodes (DISA Elektronik 13 K 03) and skin electrodes (DISA Elektronik 13 K 62) were employed. The external diameter of the needle electrode is 0.65 mm and the surface of the internal platinum conductor at the tip is 0.07 mm^2. The external cannula of the needle electrode is made of stainless steel and well insulated from the internal thin platinum conductor. The skin electrodes are round, metal plates (ϕ 7.5 mm) situated at a distance of 23 mm between the centre of each metal plate. The plates are covered with felt and at the examination the felt is moistened with electrolytical paste (Elema Schönander Mingograph Electrode Liquid).

The muscles were first examined in a totally relaxed resting position, then at slight voluntary contraction and finally at the maximum degree of contraction. The needle electrodes were inserted perpendicularly to the direction of the muscle fibres, and in view of the existing anatomical variations, they were placed at random in four different positions of each muscle at a depth of 0.5–2 cm. The individual motor unit potentials were studied on the screen of the oscilloscope, using a sweeping speed of 1 mm/ms. In studying the motor unit potentials suitable amplifications were used; the deflection sensitivity was adjusted to values of 10 μV/mm and in extreme cases it was 1000 μV/mm. Film recording of myoelectric signals was performed at a film speed of 5 cm/s.

It would have been desirable to measure the duration of the single motor unit potential, as this parameter is the most valuable in electromyographic diagnostics. However, in several cases it was impossible to collect a sufficiently large number of single and well-defined motor unit potentials which covered the requirements of duration measurements (*Buchthal & Clemmesen*, 1941; *Petersén & Kugelberg*, 1949; *Buchthal, Guld & Rosenfalck*, 1954). The reasons were that, the remaining muscles in arm amputees often are very small and the patients sensitive to pricks, and also that the skin in this area may be very thin and atrophic. According to the conventional clinical electromyographic method of examination, both duration and amplitude as well as the number of phases of the individual motor unit potentials were assessed with the naked eye on the screen of the oscilloscope.

The duration and number of phases of the individual potentials are affected by the intramuscular temperature, which was measured in a stump muscle of 30 arm amputees. In below-elbow amputees, the temperature was measured in the remaining extensor muscle of the forearm, in above-elbow amputees in the remaining brachial biceps muscle. These measurements were taken with a thermo-couple needle from ELLAB, Copenhagen (Type TE_3). The needle has a diameter of 0.7 mm and is inserted into the centre of the muscle. The temperature was checked at the end of the examinations.

Power Spectra of Myoelectric Signals, Methods (Paper I)

Measurements of the myoelectric signal power spectra were performed on 50 healthy males, aged from 20 to 50 years, and on 30 amputated males of corresponding ages. To be included in the series, it was necessary that the arm amputees were able to activate myoelectric signals up to a certain level. The myoelectric signals were picked up by means of coaxial needle electrodes. In cases of uninjured controls, signals from eight positions of the needle in nine muscles of the right-hand side extremities were analysed.

The following muscles were examined in succession: mm. biceps brachii, brachioradialis, extensor digitorum communis, deltoideus, trapezius, vastus lateralis, tibialis anterior, gastrocnemius and soleus.

The positions of the needle in the brachial biceps muscle were carefully defined in accordance with a procedure, developed by *Kaiser & Petersén* (1965). The needle positions in the other muscles were not defined with this great accuracy. Still, the needles were placed according to a consistent plan. In addition to the nine muscles mentioned above, five other muscles were investigated. In 25 controls, mm. biceps brachii, brachioradialis and extensor digitorum communis of the left side were examined. In all 50 controls, four measurements were taken from the left side and four measurements from the right side of mm. interosseus dorsalis I manus and extensor digitorum brevis. Needle positions in the latter two muscles were not selected to form any particular pattern.

In the case of arm amputees, myoelectric signals from each muscle investigated were analysed for four needle positions only, as the stump region frequently is sensitive to pain and trauma. Due to anatomical variations between the arm amputees, the needle positions could not be standardized.

In the cases of uninjured controls, the temperature of the right brachial biceps muscle was measured in the endplate zone of the long head. For 20 of the controls, the muscle temperature was measured in two definite points on the right side and in an endplate zone of the left side. The muscle temperature of the below-elbow amputees was measured in the remaining extensor muscles of the forearm. The temperature measurements were performed by means of the same thermo-couple as mentioned before (ELLAB, Type TE_3).

The power spectra were measured by means of a method and an instrument described by *Kaiser* & *Petersén* (1963, 1965). The method is based on transmission of the myoelectric signal through four octave bandpass filters centered at 50 Hz, 200 Hz, 800 Hz and 1600 Hz, respectively. The four filter outputs are rectified, and the resulting DC voltages are presented using the voltage from the 200 Hz filter as a reference. The three ratios obtained—power in the 200 Hz octave over power in the 50 Hz octave, power in the 200 Hz octave over power in the 800 Hz octave, and power in the 200 Hz octave over power in the 1600 Hz octave—provide a measure of the shape of the power spectrum.

The muscle action potentials are first amplified in a DISA Electromyograph, the output of which is fed to a specially designed spectrum analyzer. The output voltages are presented as the positions of two bright spots, the octave loci, on the screen of an oscilloscope. The horizontal deflections of the two spots are proportional to the 200 Hz/800 Hz and 200 Hz/1600 Hz activity ratios, respectively. The vertical deflections of both spots are proportional to the 200 Hz/50 Hz activity ratio (Fig. 1). All three deflection sensitivities were 1 mm/dB.

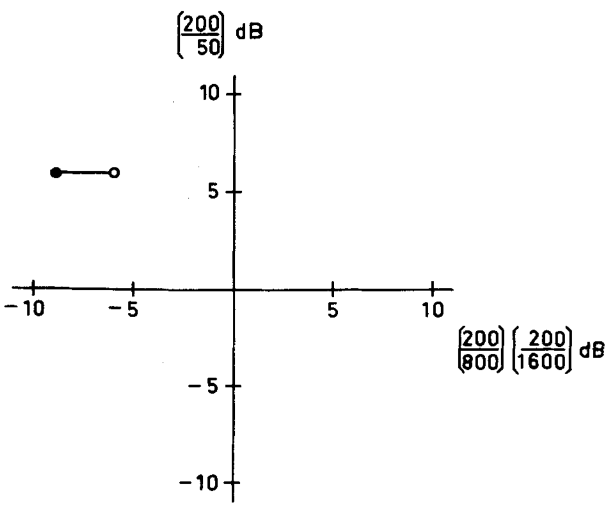

Fig. 1. Locus diagram. The vertical deflections of both spots are proportional to the ratio, in dB, of the activity (r.m.s. value) in the 200 Hz band to the activity in the 50 Hz band. The horizontal deflections are proportional to the ratio of the activities in the 200 Hz band and the 800 Hz band (circle), and to the ratio of the activities in the 200 Hz band and the 1600 Hz band (filled dot) respectively. The loci points in the figure pertain to the case of white noise.

The degree of muscle contraction was standardized with the aid of the contraction level indicator. The subjects were instructed to maintain for a few seconds, the (moderate) contraction intensity yielding a specified voltmeter deflection. During this period, the loci coordinates were read—by the same person throughout the entire test series—from the screen of the oscilloscope.

In order to give a simple description of power spectrum shape, in the standard terms of filter and circuit theory, the three power ratios were transformed into three new variables. These approximate the spectra by a constant level and a straight line, the high-frequency asymptote, which intersects the constant level at the so-called corner frequency. The three transformed spectrum parameters are: low-frequency level (*vs* 200 Hz level), high-frequency asymptote slope, and corner frequency. The statistical analyses were performed for both pairs of three parameters.

A detailed analysis of the measurement procedure indicates that the overall systematic error of the power spectrum values obtained is less than 1 dB. For further details of the procedure employed in describing the shape of the power spectrum and the statistical analysis involved, see Paper I.

Dynamic Spectrum Analysis, Method and Instrumentation

Dynamic changes in the power spectrum of the myoelectric signal, in a fatiguing isometric muscle contraction, were analysed in different octave bandpass filters by means of a broad-band frequency analyzer (*Brüel & Kjaer*, Type 2112) according to the method, used by *Kadefors, Kaiser & Petersén* (1968). Following the notation of these authors, the symbol S_f was used for the signal power, expressed in dB, within an octave band with a centre frequency of f Hz. The myoelectric signals were amplified in a DISA Electromyograph and then recorded on magnetic tape (Precision Instrument PI-200). The analysis of the spectrum was performed from the recorder. The different octave bandpass filters were switched in one by one and the signal amplitude (r.m.s.)—expressed in dB—was written by means of a level recorder (*Brüel & Kjaer*, Type 2305). The error of the entire analysis was within ± 1.5 dB.

Needle electrodes of the same type as earlier were used for picking up the myoelectric signals, and signals from several muscles were recorded for all but two of the 30 arm amputees. Among the below-elbow amputees, muscles on the extensor side of the forearm (closely corresponding to m. extensor digitorum communis) could be studied in 19 cases; this also applied to the brachioradial muscle. The remaining stump flexor muscles were studied in 12 cases. In 27 of the 30 arm amputees, the brachial biceps muscle was examined. Five of these patients had been above-elbow amputated, and thus the remaining part of the muscle only could be studied. In two above-elbow amputees and one below-elbow amputee, adequate myoelectric signals could not be picked up from the brachial biceps muscle. Where no signals were picked up the reason was either a lacking muscle or insufficient signal-to-noise ratio.

In accordance with the investigation on normal subjects, performed by *Kadefors, Kaiser & Petersén* (1968) the purpose of this study on arm amputees was to measure quantitatively spectral changes within high frequency ranges, during strong isometric muscle contraction, and to measure the recovery rate of these changes.

A standard procedure was used for the examinations. Initially, the patient performed two short, forceful muscle contractions, followed by an isometric muscle contraction—to the best of his ability—during 30 seconds, and immediately afterwards two short contractions. After 30 seconds, two short contractions were repeated and after a further 30 seconds, two short contractions and so on. The recovery of the reduced signal amplitude within high frequency ranges, which occurred during the persisting contraction, was studied during at least 90 seconds. The standard schedule is illustrated by Figure 2, showing the signal amplitudes of two octave bandpass filters. In the following, the designation $D_k = (S_{63} - S_{1000})_k$ expresses the difference in dB of the signal amplitudes for the octave band centered at 63 Hz, and the octave band centered at 1000 Hz, at a time indicated by k. The magnitude $S_{63} - S_{1000}$ was calculated for the moments indicated with arrows in Figure 2. Obviously, the reduced magnitude of the signal activity, within high frequency ranges can be calculated as $D_5 - D_{1,2,3}$ where $D_{1,2,3}$ is the mean of D_1 and D_2 and D_3. The recovery of the reduced magnitude of activity, which occurred within high frequency ranges, at different moments (k) is expressed by $D_5 - D_{k,k+1}$

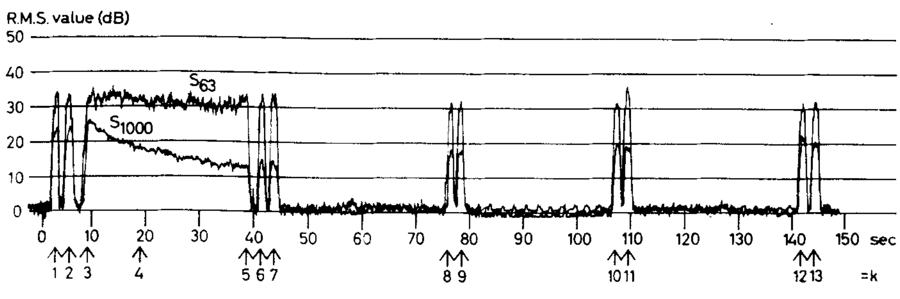

Fig. 2. The standard programme showing activity (r.m.s. value) in the 63 Hz band and the 1000 Hz band during a maximum contraction (k=3–5), followed by a relaxation period. The recovery is studied from the analysis of brisk contractions (k=6,7 . . .) and compared to the initial state (k=1, 2, 3). From *Kadefors, Kaiser & Petersén* (1968) by permission.

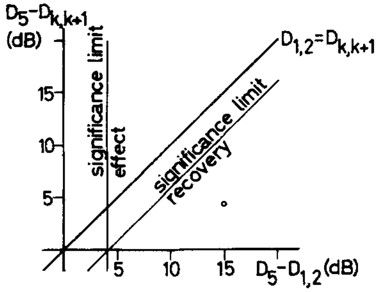

Fig. 3. Diagram showing the effect and recovery phenomena in a maximum muscle contraction. The abscissa shows increase in difference between the 63 Hz and the 1000 Hz components in dB. The ordinate shows the recovery state at a particular time instant. From *Kadefors, Kaiser & Petersén* (1968) by permission.

where $D_{k,k+1}$ is the mean of D_k and D_{k+1}. It is easier to explain this condition by a diagram (Fig. 3). In this figure a value on an abscissa of 4 dB or less, means that there is no significant reduction of the high frequency components. This value was obtained by calculating the statistical properties of the signal and by considering the error of method (*Kadefors, Kaiser & Petersén,* 1968). The figure also illustrates that if the distance between a measured value and the line $D_5-D_{1,2,3}=D_5-D_{k,k+1}$ exceeds 4 dB, the recovery is not considered to be significant. The encircled point in the figure indicates a muscle contraction, where the effect regarding the decay of high frequency components is significant, but the recovery is not complete.

The values obtained from the amputees have been compared with the normal values stated by *Kadefors, Kaiser & Petersén* (1968). No mention is made as to which side was tested at the investigation, but according to personal communication with the authors, all the signals were picked up on the right-hand side. The purpose of the present investigation was to study the dynamic changes in the spectrum of myoelectric signals obtained from amputees in connection with fatiguing muscle contractions. Due to the limited number of observations of each muscle, the series was not divided into left- and right-hand sides nor into levels of amputation. A comparison with the normal series is despite this of importance as, for practical reasons, it is valuable to know any deviating dynamic changes of the spectrum in amputees before performing signal processing. Differences in muscle effect and recovery between normals and amputees were analysed by the ordinary T-test.

Methods of Comparing Electrodes (Papers II, III)

Several methods have been used for comparison of the properties between different types of electrodes. In three commercially available surface electrodes for prostheses (Austrian-Viennatone, Italian-INAIL and the Russian) the impedance was measured as a function of frequency. The impedances of the electrodes were calculated by means of the voltage drop at a constant current of 10 μA, measured with the aid of a phase-sensitive voltmeter (Solartron VP 250). All the recordings were performed with the electrode in a standardized position on the extensor side of the forearm. The ground electrode was placed on the flexor side and consisted of a large, wet silver plate.

By means of a voltmeter (HP 425 A) differences between the half-cell potentials of the electrodes were measured. These measurements were also performed in a standardized position on the extensor side of the forearm. Before the recordings, the skin was carefully cleaned with alcohol and each type of electrode was tested ten times.

A comparison of the signal-to-noise ratio in dB between the different electrodes could be obtained by measurements performed on similarly placed electrodes on the forearm, and at accurately controlled muscle contractions. By loading the finger extensors of the forearm with a constant, rather small weight (1 kg), a comparable degree of contraction was obtained. As an amplifier a DISA Electromyograph was used and the 250 Hz octave band was selected for the analysis, giving the best signal-to-noise ratio. The noise level was measured at complete relaxation and was reproducible to within 2–3 dB.

Some spectral characteristics obtained in the signal-to-noise ratio investigation were compared between the surface electrodes and an implanted electrode. The analysis was performed on an octave band analyzer (*Brüel* & *Kjaer*, 2112).

Six electrodes for telemetry of the myoelectric signals were implanted on four voluntary subjects, two uninjured controls and two below-elbow amputees. The operations were performed under local anaesthesia and the electrodes were placed subcutaneously above the fascia. A technical description of the electrode is given in Paper III. The subjects were observed clinically, and the tissue reaction after removal of the electrodes was studied macro-and microscopically. The following methods for analysis of the recorded signals were used; study via rectification and integration of the total signal, spectral analysis and study via graphic recording. In the present study analysis has been undertaken by using a tape recording velocity of eight times the playback velocity. All frequencies were thus divided by eight. A conventional ink-jet recorder (ELEMA Schönander Mingograph 42) capable of writing frequencies up to some hundreds of Hz is adequate. Rectification and integration of the total signal and spectral analysis were performed with the same analyzer as used before (*Brüel* & *Kjaer*, 2112). Furthermore, spectral analysis was performed with the apparatus described by *Kaiser* & *Petersén* (1963, 1965).

Control Ability Test: Apparatus and Procedure (Paper IV)

When a muscle is utilized as a control site for prosthesis control the fact that anatomical and physiological conditions differ between various muscles is of importance. The distal extremity muscles often have a larger degree of innervation (*Feinstein, Lindegård, Nyman & Wohlfart*, 1955) and thus their control is more appropriate as compared to proximal muscles. This is believed to cause differences in speed and fatiguability (*Tergast*, 1873). The aim of this investigation was not to study the speed and dynamics of separate muscles in order to characterize a control site. The purpose was to get an idea of the ability to manoeuver a specially designed electro-mechanical test instrument using myoelectric signals. This yielded knowledge of the ability of uninjured and arm amputated males to perform rapid and voluntary quantified muscle contractions.

In a preliminary investigation three muscles in 30 uninjured males, aged between 20 and 50 years, were studied, *viz.*: mm. extensor carpi radialis longus, deltoideus and vastus lateralis (Paper IV). Based on the experience of this investigation, the instruction of the test subjects was modified at the definite investigation. Thus, a further series of 30 uninjured males—in the same age groups—was tested. The nine muscles included in the definite investigation were: mm. extensor carpi radialis, flexor digitorum sublimis, interosseus dorsalis I manus, biceps brachii, triceps, deltoideus, trapezius, vastus lateralis, tibialis anterior. All these tests were performed on the right-hand side.

In 30 arm amputees with varying levels of amputation, a number of muscles were studied: the stump muscle on the extensor and flexor sides of the forearm, mm. biceps brachii and deltoideus. The amputations were performed on the right-hand side in 13 cases. The method does not permit any differentiation of the manoeuverability between the right- and left-hand sides. Consequently, there was no distinction made between the right- and left-hand sides in the amputation series.

Surface electrodes (DISA Electronik 13 K 62) picked up the myoelectric signals which were amplified in a DISA Electromyograph before being connected to one of the inputs of the test instrument. A signal from a test programme—recorded on tape—was connected to the other input. The actual test instrument is electro-mechanical and the technical description of this and of the test programme is described in Paper IV.

Apparatus and Test Programme

The two input signals, the myoelectric signal and the programme signal, are rectified and converted to pulses of a frequency, proportional to the level of the input signals. The output pulses are connected to a stepping motor, which runs forwards and backwards, depending on which input signal has the highest amplitude. In the mechanical system the stepped motor activates a tiltable

Fig. 4. The electro-mechanical apparatus with its frame and fly-wheel as the test subject will experience the system during the course of examination. Below a tape recorder and the standard programme.

frame which moves freely around a horizontal axis, having moving increments of 1.7 milliradians per pulse, constituting the so-called primary movement. A cylindrical shaft is placed on the frame and when the frame tilts the shaft rotates, thus causing the secondary movement. On the front end of the cylindrical shaft, which protrudes through an aperture in the face plate of the apparatus, there is a fly-wheel with an outer diameter of 14 cm. The rate of angular acceleration of this wheel is 8 milliradians/sec^2 per milliradian tilt of the frame. A mark on the wheel indicates the centre of the system (Fig. 4).

At a certain adjustable angular deflection of the wheel, a micro-switch is activated and its position is recorded continuously (ELEMA Schönander Mingograph 42). The error is chosen so that the error-indicating switch is activated if the wheel deviates more than $\pm 90°$ from the centre. The accumulated time, the so-called off-time, during which the position of the wheel lies outside the error limits, is indicated on the recorder.

The test programme was tape-recorded from a laboratory oscillator. For each test on each muscle the total time required was 260 seconds. The programme signal is connected to a loud-speaker, one input of the apparatus and the recorder. To neutralize the great importance of the factor of adaptation, the subjects were both at the preliminary and at the definite tests initially examined twice, utilizing the same muscle. The difficulty factors of the test programme and the amplification of the DISA—apparatus were selected in a way which obviously permitted all the subjects to grasp quickly the principle of the programme and fulfil it with minor errors. In all the subjects the muscles were tested in sequences.

During the tests the subjects were comfortably seated in an adjustable chair with the possibility of relaxing completely (Fig. 5). A carefully elaborated tandard procedure was followed at each examination. To begin with the sub-

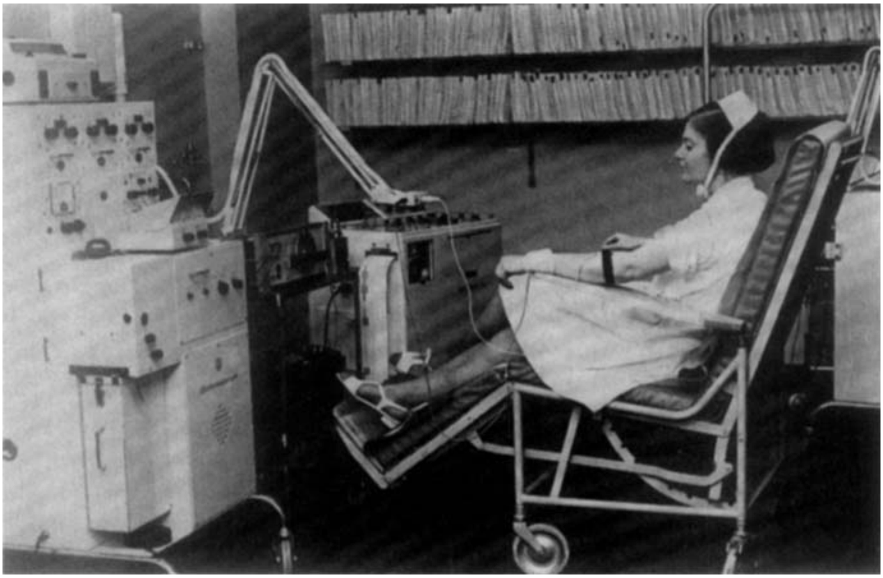

Fig. 5. A test subject sitting comfortably in front of the apparatus during the course of the manoeuvering test. To the left, the DISA Electromyograph and to the right of the apparatus the recorder. Surface electrodes are used to pick up the myoelectric signals.

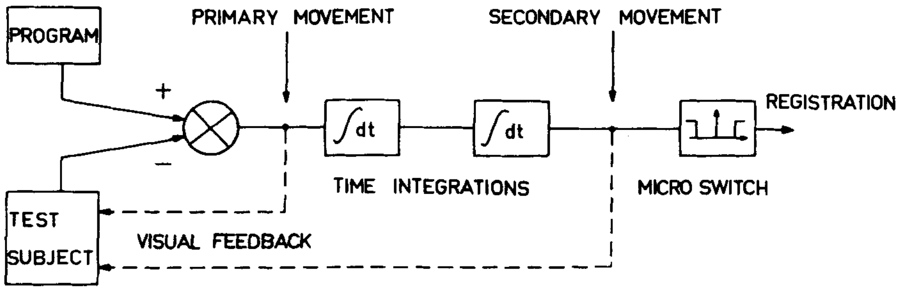

Fig. 6. Block diagram showing the function of the electro-mechanical apparatus in principle. The interaction between the deviation of the frame and the angular deviation from the centre position of the fly-wheel can be expressed as two successive time integrations. There are two visual feed-backs, the deviations of the frame and the fly-wheel, respectively.

jects were instructed about the purpose of the examination, the function of the instrument and the test procedure. They were then requested to try to maintain the frame in a horizontal position and the fly-wheel in the centre by alternating activation and relaxation of the muscle to be tested. However, there are two differences in the methods of the preliminary (Paper IV) and the definite investigations. In the first place it was especially emphasized that the subjects should concentrate on the tilt of the frame and only in extreme positions should they pay attention to the fly-wheel. Secondly, the subjects rested comfortably with their arms on a support at the definite investigation of the deltoid muscle. As a matter of fact, it proved difficult to relax the deltoid muscle with the arm hanging, as at the previous test.

The function measured by this apparatus is, of course, extremely complex. The relation between man and machine, according to the above test, is approximately illustrated in a block diagram (Fig. 6). The comparatively uncomplicated test programme resulted in the subjects satisfactory control of the apparatus during the greater part of the test. The fly-wheel deviated considerably from the centre during short periods only. The secondary movement, *i.e.* the position of the wheel is characterized by an approximately normal distribution, and the instability of the position of the wheel is best described by the standard deviation. To designate this complex function of the apparatus, the term manoeuverability was introduced.

A study of the manoeuverability of various muscles in both uninjured subjects and amputees was considered justified in view of the future prosthesis applications. The following questions arose at the definite investigation:

1. Does the ability to manoeuver the apparatus, myoelectrically, differ between various arm muscles in healthy males?

2. If such a difference exists, is it correlated to the proximal muscles as compared to the distal muscles?
3. Does the ability to manoeuver the apparatus differ with respect to the remaining muscles in arm amputees being normal or showing a peripheral neuron lesion according to conventional electromyography?
4. Does the ability to manoeuver the apparatus differ between healthy and arm amputated males with corresponding muscles?

Statistical Analysis

The statistical methods used to evaluate the results from this investigation were designed separately for each of the questions at issue.

The off-times for the nine muscles of normal males were analysed by means of a two-way analysis of variance, mixed model, where the muscles were considered as a fixed effect and individuals as a random effect. No replicates were made which means that the interaction cannot be tested. For physiological reasons, however, it can be assumed that the interaction effect is negligible. In order to fulfil the conditions of the analysis of variance it was performed on the natural logarithms of the off-times. When the analysis of variance showed significant main effects, Scheffe's S-method was employed to compare individual contrasts.

The difference in off-times between muscles in arm amputated males showing normal or pathological EMG was analysed statistically using the Wilcoxon Two-Sample Rank Test on the original values. This method was chosen because of the rather limited number of observations which did not satisfy the conditions of the normal approximation.

The difference in off-times between muscles in normal and arm amputated males was analysed by the ordinarily normal test, on the natural logarithms of the off-times.

A 5% level of significance was used throughout. The statistical analyses employed are described in full detail in *Brownlee* (1960).

MATERIAL

The series included 30 males who had been arm amputated during the period 1934–1966. With the exception of five patients, resident in Stockholm, they all belonged to the Gothenburg area and received their prostheses at the Orthopaedic Clinic in Gothenburg. Males of ages ranging from 20 to 50 years were selected as the males in the control groups were of corresponding ages. No patients with congenital defects were included. As the purpose was to study changes in stump muscles following amputation, only above- and below-elbow amputees were examined. No patients with fore-quarter resections were included. In the series available, no patients had exarticulations either in their wrists or their elbow-joints. One bilateral arm amputee was included. As the deficiency in this patient was congenital on one side, only the operated arm was examined.

Table 3. Level of amputation related to age.

Age at examination	Above elbow	Below elbow	Total
20–25		4	4
26–30	1	5	6
31–35			
36–40	1	1	2
41–45	3	5	8
46–50	2	8	10
Total	7	23	30

The series consisted of seven above-elbow amputees and 23 below-elbow amputees and the distribution of the levels of amputation in the various age groups is tabulated (Table 3). One third of the patients were aged between 20 and 30 years, and one third between 45 and 50 years, and the remaining patients were distributed between the two first age-groups. As most of the patients were operated upon at hospitals in different parts of Sweden and abroad, many years ago, information about the technique used at the operation was in few cases only available.

The patient's age at the time of amputation and the number of years elapsed since can be studied in Table 4. Most of the patients were operated on between

Table 4. Age at amputation related to number of years of handicap.

Period of amputation, years	Age at amputation 2–10	11–20	21–30	31–40	41–50	Total
2– 5		2	1		1	4
6–10					1	1
11–15	1	2	3	2		8
16–20	3	1	1	2		7
21–25		2	1			3
26–30		4	1			5
31–35	1		1			2
Total	5	11	8	4	2	30

Table 5. Causes of amputation in relation to age at amputation.

Age at amputation	Etiologi: Industrial accidents	Traffic accidents	Different trauma	Tumour	Infection	Total
2–10		1	3	1		5
11–15	1		3			4
16–20	2	1	2	1	1	7
21–25	2	1	2			5
26–30	2	1				3
31–35	2					2
36–40	2					2
41–45		1				1
46–50	1					1
Total	12	5	10	2	1	30

the ages of 11 and 30 years. This is in accordance with several earlier publications in which it was stated that arm amputations mainly strike young individuals. The number of years, during which the patients in this series had been amputated, are evenly divided into intervals of five years, from one year up to 33 years. In half of the series, the operation had been performed 10–20 years earlier.

Recent investigations show that the most common cause of arm amputations are traumatic injuries. As the congenital defects, constituting the next largest etiological group of arm amputees, were excluded from this series, the traumatic causes predominate. This is seen in Table 5, in which the causes of amputation and the age of the patients are stated. Traumatic injuries occurred in 27 cases, tumours in two cases and infection in one case. Industrial and traffic accidents strike different age groups more or less at the same rate.

In the remaining group of trauma it is mostly children and adolescents who are the victims, for instance, in blasting accidents when playing with explosive material. In 13 cases the right arm and hand were amputated and in 17 cases the left.

Despite long-standing handicap, most of the patients as a rule have a remaining phantom limb perception of the lost hand (*Cronholm*, 1951). In 19 cases this perception did not cause any discomfort, but in seven cases pain was experienced which the patient localized to the lost hand (Table 6). Besides this phantom pain, four patients experienced a varying degree of pain in the stump. Of the patients claiming phantom pain, four stated that it was gradually disappearing. Three patients mentioned that the actual phantom perception of the lost hand had become reduced. In 18 of the 26 patients who had a phantom perception, there was a general tendency to perceive the entire hand, while the remaining 8 only perceived part of the hand. Furthermore, five patients claimed that the stump was extremely sensitive to cold.

Among the types of prosthesis, the passive cosmetic hand predominates and in this series 18 patients use it regularly all day. The prostheses have been classified in active, functional hand prostheses and hooks as opposed to passive prostheses of different types. Table 7 depicts the distribution of the

Table 6. Number of patients with a phantom-hand perception in relation to amputation level.

Level of amputation	Phantom-hand perception		
	Without pain	With pain	Total
Above elbow	4	2	6
Below elbow	15	5	20
Total	19	7	26

Table 7. Number of different types of prostheses used in relation to amputation level.

Level of amputation	Type of prosthesis used					
	Active hand	Passive hand	Active hook	Passive hook	Passive other type	Total
Above elbow	1	5		2	1	9
Below elbow	3	13	4	5	3	28
Total	4	18	4	7	4	37

prostheses with regard to the level of amputation. A combination of two types of prosthesis was used by nine patients. A passive hook was exchanged for the cosmetic hand in certain activities. One above-elbow and one below-elbow amputee never used any prosthesis. Furthermore, there were two patients who only used their prosthesis at work. A third of the patients had tried but never learnt to utilize and appreciate an active prosthesis, fitted by means of wires. Among the seven above-elbow amputees, five patients use an actively controlled elbow-joint and in addition to this a cosmetic hand. These patients could not accept two cables, one for elbow flexion and one for prehension. The fact that the patients do not utilize their active prostheses may be due to the lack of specialized rehabilitation centres in Sweden for arm amputees. Seven patients stated that they are dependent on their prosthesis.

In general few arm amputees regularly use their prosthesis. However, there are great variations in earlier published findings. *Müller* (1962) in his follow-up of 64 arm amputees found that 29 per cent used their prosthesis regularly. *Nathan & Davidoff* (1965) stated that of 36 arm amputees, 14 never used their prosthesis. In the present series of 30 amputees, 25 use their prosthesis every day at work and at home. The fact that the series is a selected group in which the patients are of ages below 50 years, may explain their using the prosthesis. Due to the fact that every patient can receive a prosthesis free of charge in this country, a very detailed and individual technical fitting can be made. This may also explain the high frequency of prosthesis carriers.

When asked, 21 patients stated that they managed independently at work and at home. Eight patients required assistance in tying their shoe-laces, neck-ties and in fastening buttons. Only one patient mentioned that he required help every day with his personal care. This patient was above-elbow amputated and slightly mentally retarded.

Mental disturbances, mainly depression and a tendency to isolation are common complaints, and two thirds of the patients had similar trouble during the first years after the amputation. On the other hand, there are only four patients who have remaining mental complications due to their handicap. However, all the patients have adapted themselves socially, except one, who is a criminal.

As the patients are well taken care of and offered the possibility of learning a new job—at public expense often during several years—their social adaptation becomes facilitated.

RESULTS

Clinical Investigation

All the patients were subjected to the following clinical examination: inspection, palpation, assessment of range of motion and force, and testing of sensibility of the stump. The range of motion in the shoulder and the elbow joints and the pronation and supination were measured with a goniometer. The physical strength of the arm was compared with that of the uninjured side and classified as normal or considerably reduced. The sensibility of the stump was tested for pricks and touch and subjected to Weber's two-point discrimination test, according to the procedure emphasized by *Moberg* (1962, 1963, 1964). The sense of vibration was examined on all the stumps. The length of the amputation stump was measured. On the below-elbow amputees the length on the dorsal side of the forearm from the tip of the olecranon was measured, and likewise from the acromion on the lateral side of the stump, in above-elbow amputees. The lengths of the amputated stumps have varied considerably (Table 8).

Table 8. Stump length in relation to amputation level.

Length of stump, cm	Above elbow	Below elbow	Total
0– 5		2	2
6–10		5	5
11–15	2	2	4
16–20	2	10	12
21–25	1	3	4
26–30	2	1	3
Total	7	23	30

Inspection

Among the below-elbow amputees, the stump generally was of a good shape, but in six patients, the stumps which were short revealed different degrees of deformation, *i.e.* lengths below 10 cm. The patients with long below-elbow stumps nearly all had normal skin and thin, short, movable scars across the tip of the stump. However, two of these patients with long below-elbow stumps had larger scars, in one case consisting of two skin grafts. In seven of the below-elbow amputees with short stumps, more than half of the cases had small areas of adherent, thin skin. Two of these had excess skin which deformed the stumps and made them soft. These patients have had difficulties in getting a satisfactory prosthesis application. Only one patient with a short below-elbow stump had a symmetrical stump with acceptable skin.

At the inspection five of the seven patients subjected to above-elbow amputation were found to have considerably deformed stumps. The remaining muscles were small and retracted in these five patients and, furthermore, there were large scars, to a great extent adherent to the subjacent bone. One of the above-elbow amputees had a soft stump with an excess of skin.

Already at the inspection, the remaining part of the extremity revealed considerable atrophy in several cases. In the below-elbow amputees, atrophy of the upper arm was found in six patients. However, the shoulder muscles were not affected in these patients. Atrophy was present in all the above-elbow amputees. Among these, there were three cases with reduced shoulder muscles, mainly including the deltoid muscle and the supraspinate muscle. As all the above-elbow amputees in this series, except for one patient, only utilize cosmetic hand prostheses, atrophy will often set in due to inactivity.

Palpation

Besides the skin, also the muscles and the other tissues were examined when palpating the stump. As has been pointed out earlier, the above-elbow amputees had considerably reduced stump muscles and the remaining part of the biceps muscle was retracted in all the cases but one. In the below-elbow amputees, the stump muscles were seldom reduced. Among the patients with long below-elbow stumps, the muscle was reduced only in three cases. Retraction of the stump muscle was found in two below-elbow amputees.

In the below-elbow amputees, there were three cases of neuroma, all originating from the median nerve. In the above-elbow amputees, two neuroma, originating from the radial and axillary nerves, were found. At the palpation, all the neuroma were diagnosed due to pain or their considerable size. In that connection, attempts were made to assess the patient's ability to contract the remaining stump muscles on the extensor and the flexor sides, irrespectively of each other. Only in one below-elbow amputee with an extremely long stump was this ability noticed without earlier training. As a whole, the patients have clearly experienced their inability to contract the remaining muscles independently of each other, and had a sensation of activating the entire stump as a unit. The patients who perceive distinctly the opening and closing of their phantom-hand, no doubt, found it easier to try to contract the remaining stump muscles independently.

Range of Motion

Three below-elbow amputees and five above-elbow amputees had impaired mobility of their shoulder joint on the amputated side. The rate of below-elbow amputees with this impairment is smaller as compared to earlier investigations. Thus, *Solonen et al.* (1965) found impaired mobility of the shoulder joint in 27 per cent of 48 above-elbow amputees and in 29 per cent of 24 below-elbow amputees. This discrepancy is, no doubt, due to the low mean age of the below-elbow amputees in this series, while the mean age of Solonen's series of disabled soldiers was 45 years. Painful impaired motion of the shoulder joint was found in one patient only. The degree of limited motion was slight and in all the patients the range of abduction was at least 160° and the degree of rotation was a minimum of 120°.

Of the below-elbow amputees, 15 patients had a normal range of motion of the elbow joint on the amputated side. In a third of these patients, impaired flexion was observed. This was pronounced only in three cases, *viz.* between 25° and 75°. Hyperextension between 20° and 30° was found in three patients. None of the below-elbow amputees experienced pain when forcing the movements of the elbow-joints, which were all stable in an extended position. No radiographic examination was performed.

The remaining ability to pronate and supinate, depended on the length of the stump (Table 9). In the cases where the length of the stump exceeded 15 cm, all except two patients could rotate their forearm more than 90°.

Table 9. Range of movement, pronation, supination at different stump lengths.

Length of stump, cm	Pro-supination –45°	45–90°	> 90°	Total
0– 5	2			2
6–10	4	1		5
11–15	1	1		2
16–20		4	6	10
> 20			4	4
Total	7	6	10	23

Physical Strength

The physical strength in the shoulder muscles of the above-elbow amputees was considerably reduced in three patients on the amputated side. The others had, despite existing atrophy, an almost normal physical strength. The force of extension in the elbow-joint was considerably reduced in five below-elbow amputees. The force of flexion of the elbow joint was considerably reduced as compared to the uninjured arm in six patients. Furthermore, in two patients the force in flexing the elbow-joint was extremely reduced. The pronation and supination strength was almost intact in half of the patients with long stumps. The other below-elbow amputees who had a remaining pronation-supination ability, revealed varying strength which, however, was difficult to estimate when the stumps were short.

Sensibility

In all the patients a tuning fork (C 128) was used to check that the sense of vibration was normal. In 12 of the amputees there was reduced sensibility within different areas of the stump. In five of the cases the disturbed sensibility for pricks and touch was localized in grafts or large scars. All the above-elbow amputees, except one, had impaired sensibility in areas of varying sizes. A pathological two-point discrimination as compared to the uninjured arm was found in 11 patients. In five cases the pathological two-point discrimination was also found only within areas of cicatrices or grafts. Among the other six patients, two-point discrimination was measured to range between 30 and 40 mm in two cases. In four patients no two-point discrimination could be measured in the comparatively small areas involved. The size of the skin area where an impaired or a neutralized sensibility was found, varied between 5 and 18 cm^2 and was situated within the distal part of the stump.

The clinical results are summarized in Table 10. The table shows that below-elbow amputees with short stumps on the whole revealed numerous changes. The stumps were irregularly shaped, the skin was thin, often adherent and frequently the sensibility was impaired. There was no pronation or supination left and the mobility of the elbow was reduced, as the physical strength was impaired as a rule. Technically, it is always difficult to apply a prosthesis to a short stump. Due to the above-mentioned reasons, the changes among these patients further complicate a satisfactory fitting of the prosthesis. In patients with long forearm stumps these changes are only found occasionally. However, the length of the stump is decisive for the degree of rotation. The stumps of the above-elbow amputees are all irregular and atrophic, often the skin is thin and the sensibility is disturbed. Despite this, they have

Table 10. Results of the clinical investigation. For further information see the text. + = yes; − = no.

Amputation level	Stump length cm	Inspection: Symmetric stump	Inspection: Normal skin	Inspection: Thin scar	Palpation: Normal muscles	Palpation: Painful neuromas	Palpation: Independ. muscle contr.	Range of motion: Shoulder normal	Range of motion: Elbow extension-flexion	Range of motion: Pronation	Range of motion: Supination	Physical strength: Normal elbow extension	Physical strength: Normal elbow flexion	Physical strength: Normal pronation	Physical strength: Normal supination	Sensibility: Normal pain	Sensibility: Normal two-point discrim.
Below elbow	4	−	−	−	−	−	−	+	180–110	0	0	−	−			−	−
	5	−	−	−	−	−	−	−	180– 90	0	0	−	−			−	−
	8	−	−	−	+	+	−	+	180– 50	10	0	+	+			+	+
	9	−	−	−	−	−	−	−	210– 45	10	10	+	+			−	−
	9	+	+	+	−	−	−	+	200– 60	10	10	+	+			+	+
	10	−	−	−	+	−	−	+	185– 40	40	20	−	−	−	−	+	+
	10	+	+	+	−	−	−	+	180– 70	0	0	+	−			−	−
	12	−	−	+	−	+	−	+	180– 40	0	0	−	−			−	+
	14	+	−	+	+	−	−	+	180– 35	45	40	+	+	+	+	+	+
	16	+	+	+	+	−	−	+	180– 35	60	80	+	−	−	−	+	+
	16	+	+	+	+	−	−	+	200– 35	45	20	+	+	−	−	+	+
	17	−	+	+	+	+	−	+	180– 35	90	50	+	+	−	−	+	+
	17	−	+	+	+	−	−	+	180– 35	45	15	+	+	−	−	+	+
	17	+	+	+	+	−	−	+	180– 35	70	20	+	+	−	−	+	+
	17	+	+	−	−	−	−	−	180– 35	30	70	−	−	−	−	+	+
	18	+	−	−	+	−	−	+	180– 45	20	45	+	+	−	−	−	−
	19	+	+	+	+	−	−	+	180– 35	80	60	+	+	+	+	+	+
	20	+	+	+	+	−	−	+	180– 30	80	45	+	+	−	−	+	+
	20	+	+	+	−	−	−	+	180– 35	75	45	+	+	+	+	+	+
	22	+	+	+	+	−	−	+	180– 35	90	90	+	+	+	+	+	+
	24	+	+	+	+	−	−	+	180– 35	65	30	+	+	+	+	+	+
	25	+	+	+	+	−	−	+	180– 40	50	70	+	+	+	+	+	+
	29	+	+	+	+	−	+	+	180– 35	90	90	+	+	+	+	+	+
Above elbow	11	+	−	−	−	−	−	−								−	−
	15	−	−	−	−	−	−	+								−	−
	17	−	−	−		+	−	−								−	−
	20	−	−	−	−	−	−	−								−	−
	24	−	−	−	−	−	−	−								−	−
	26	−	−	−	−	+	−	−								−	−
	27	−	+	+	−	−	−	−								+	+

a remarkably good and painless mobility of the shoulder joint and the physical strength is only reduced moderately.

Electromyographic Results

The electromyographic results are indicated in Table 11. In the above-elbow amputees three different muscles were studied and in the below-elbow amputees six muscles. In some cases a muscle was excluded from the investigation as the remaining muscle, if any, could not be identified. Pathological changes of the myoelectric signals, picked up from stump muscles, were found in both above- and below-elbow amputees in 40 per cent of these muscles. The type of pathological changes in the myoelectric signal appeared as a peripheral neuron lesion, in all cases except one. This patient had a serious infection which resulted in amputation due to gangrene. In the extensor muscle and the brachioradial muscle of the forearm stump, there were EMG changes which could be interpreted as myopathy. However, similar changes are found in reinnervation.

In three cases of the above-elbow amputees there was a typical peripheral neuron lesion present in the remainders of the biceps muscle and in two cases of the triceps muscle. Among the below-elbow amputees, there existed a peripheral neuron lesion in the brachial biceps in three cases and in the triceps muscle in two cases. The latter amputees obviously had lesions of the lower

Table 11. EMG findings in different muscles of the amputation side, in relation to level of amputation.

Investigated muscles	Above elbow				Below elbow			
	Normal EMG	Lower motor-neuron lesion	Myopathy or reinnervation	No. of muscles available for exam.	Normal EMG	Lower motor-neuron lesion	Myopathy or reinnervation	No. of muscles available for exam.
M. deltoideus	6	1	—	7	23	—	—	23
M. biceps brachii	4	3	—	7	20	3	—	23
M. triceps	2	2	—	4	21	2	—	23
M. brachioradialis					18	2	1	21
M. extensor digitorum					13	8	1	22
M. flexor digitorum					15	6	—	21
Total	12	6	—	18	110	21	2	133

Table 12. EMG findings in the investigated muscles in relation to amputation level and length of stump, + indicates normal EMG, − lower motoneuron lesion, *myopathy or reinnervation.

Level of amputation	Length of stump cm	EMG findings: Stump muscle extensor side of the stump	Stump muscle flexor side of the stump	M. brachio-radialis	M. biceps brachii	M. triceps	M. deltoideus
	4			+	−	−	+
	5	−	−	−	+	−	+
	8	+	+	+	+	+	+
	9	−	+	+	+	+	+
	9	+	+	+	+	+	+
	10	M*	−	M*	+	+	+
	10	−			−	+	+
	12	+	+	+	+	+	+
	14	−	+	+	+	+	+
	16	+	+	+	+	+	+
	16	+	−	+	+	+	+
Below	17	+	+	+	+	+	+
elbow	17	−	+	+	+	+	+
	17	+	+	−	+	+	+
	17	−	−	+	−	+	+
	18	−	+	+	+	+	+
	19	+	+	+	+	+	+
	20	+	+	+	+	+	+
	20	−	−	+	+	+	+
	22	+	+	+	+	+	+
	24	+	+	+	+	+	+
	25	+	−	+	+	+	+
	29	+	+	+	+	+	+
	11				+		+
	15				+		+
	17				−	−	+
Above	20				−		+
elbow	24				−	+	+
	26				+	+	+
	27				+	−	−

motoneuron at a much higher level than that of the amputation. Peripheral neuron lesions were a common finding in the stump muscles of the below-elbow amputees. This was observed in eight cases on the extensor side and in six cases on the flexor side of the stump. Thus, in one third of these patients, the myoelectric signal was pathological in the extensor and the flexor stump muscles. These muscles constitute the muscle pair which so far have been commonly explored as control sites in electric hand prostheses. However, the pattern of a peripheral neuron lesion only appeared in two cases in the brachioradial muscle.

Table 12 illustrates the EMG results and their distribution within various muscles of the same patient as related to the length of the stump. It can easily be seen that pathological changes of the myoelectric signals were found in both long and short stumps. The clinical changes as found in all the patients with short stumps are not in accordance with the pathological EMG changes. The table further shows that of all the muscles examined in the amputated extremity, only ten below-elbow amputees had EMG findings which could be considered as "normal". In seven other patients a peripheral neuron lesion was found in one muscle only. Thus, it could be stated that a fourth of the below-elbow amputees had pathological myoelectric signals in at least two muscles of the injured extremity.

The three muscles which have been examined in above-elbow amputees showed a normal EMG in one patient only. Three patients had a peripheral neuron lesion in one muscle. Half of the above-elbow amputees had no normal myoelectric signals in two of the three muscles which are the most natural to use as control sites.

The changes in the myoelectric signals which are designated peripheral neuron lesions were usually moderate. Pronounced lower motoneuron lesion with remaining single, large potentials were found in three cases only. This is illustrated in Figure 7, where only single, very large potentials were picked up from the stump muscle on the flexor side of the forearm in a patient with an extremely short stump. This patient also had a peripheral neuron lesion in

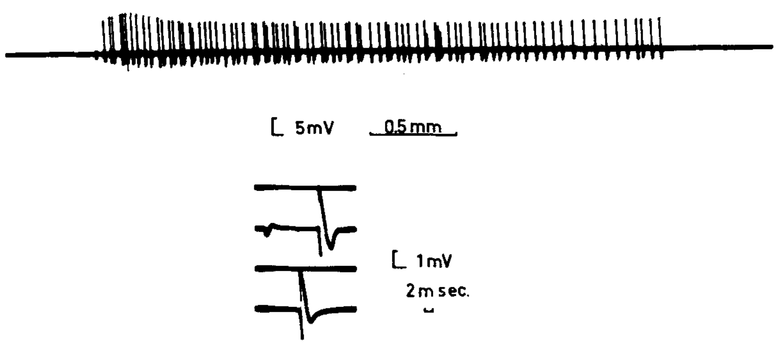

Fig. 7. Advanced lower motoneuron lesion. The myoelectric signals were picked up by a needle electrode from a small, remaining muscle on the flexor side of a short forearm stump. The single motor unit potentials was extremely large with an amplitude of about 4 mV.

the extensor side of the stump and the triceps muscle. There were fibrillations in all these muscles and also spontaneous positive potentials of short duration. This patient was operated on two years ago. Fibrillar action potentials were found, in one muscle of another patient who had been operated upon twelve years ago.

In addition to the cases in which was found a decided lower motoneuron lesion, other deviations in the appearance of the myoelectric signal were observed. Thus, there was an increased number of polyphasic potentials, approximately 10–20 per cent, in nine muscles, as compared to the about 4 per cent commonly stated in the literature as normal for limb muscles. The increased rate of polyphasic potentials was present in the stump muscles of below-elbow amputees, except in one case. A further deviation from the normal EMG was the presence of monotonous motor unit potentials with a duration which was estimated to be slightly increased. Such potentials were observed in seven stump muscles of below-elbow amputees, and in two of above-elbow amputees.

Figure 8 illustrates a number of potentials of short duration, approximately 4–5 milliseconds and an increased rate of extremely polyphasic potentials. These potentials were picked up from the same needle position in the brachioradial muscle of a patient who had been amputated for gangrene caused by a septic embolus. The EMG findings could be indicative of myopathy, but similar changes could be a sign of reinnervation. In view of the cause of amputation, and of the fact that in this case there was myositis, the change in the myoelectric signals was regarded as myopathy.

The electromyographic findings in the stump muscle on the extensor side of the forearm and in the brachial biceps muscle, varied with the intramuscular

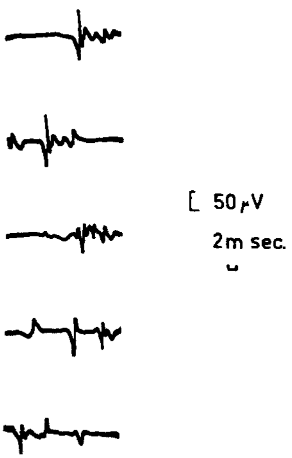

Fig. 8. Singular motor unit potentials picked up from the brachioradial muscle in a below-elbow amputee, needle electrode in the same position. The potentials are of short duration and there is a surplus of polyphasic potentials.

temperature according to Table 13. The table clearly shows that the intramuscular temperature had dropped in all muscles showing peripheral neuron lesions, increased duration, or a surplus of polyphasic potentials. All the myoelectric signals regarded as normal, were picked up in muscles with an almost normal temperature. In nine cases the intramuscular temperature was 36 °C or more. Temperatures between 35 and 36 °C were noted in fifteen cases. Six patients had a very low muscle temperature, which may be due to traumatic injuries of the vessels. There was no close relation between the length of the stump and the intramuscular temperature. Of the six patients who had temperatures below 35 °C, one patient only had a stump shorter than 10 cm.

Table 13. EMG findings in relation to intramuscular temperature of the same muscles

Intramuscular temperatures	EMG findings
36.5	Normal
36.4	Normal
36.4	Normal
36.4	Normal
36.4	Normal
36.3	Normal
36.3	Normal
36.0	Normal
36.0	Normal
35.9	Normal
35.9	Normal
35.8	Lower motoneuron lesion
35.8	Lower motoneuron lesion
35.7	Increased duration
35.6	Normal
35.5	Normal
35.5	Lower motoneuron lesion
35.5	Lower motoneuron lesion
35.4	Lower motoneuron lesion
35.3	Normal (Below elbow; Temp. in m. biceps)
35.2	Normal
35.1	Increased duration
35.0	Lower motoneuron lesion, Increased duration
35.0	Surplus of polyphasic potentials
34.7	Lower motoneuron lesion
34.5	Lower motoneuron lesion
34.5	Surplus of polyphasic potentials
34.5	Lower motoneuron lesion
34.3	Myopathy oreinnervation
32.0	Lower motoneuron lesion

Statistical analyses of variance reveal, with respect to nine right side muscles of the uninjured controls, statistically significant differences of the spectrum parameters of individuals as well as of muscles. This result holds for all the six spectrum parameters. The most important source of variation for all six parameters is to be found within the muscle. The next most important source of variation, also for all six parameters, is the difference between muscles. The least important factor—again for all six parameters—is interindividual variation.

The fact that interindividual variations are significant implies that individuals having high parameter values for one muscle have high parameter values for other muscles also.

Various muscle pairs were compared by means of the *T*-method. The only pairs displaying no statistically significant spectrum parameter differences at all are m. extensor digitorum communis *vs* mm. brachioradialis and deltoideus. Nine muscle pairs show significant differences for two parameters only. The results indicate a higher low-frequency content for the trapezoid muscle and a lower low-frequency content of the soleus muscle. The mm. brachioradialis and deltoideus have the lowest numbers of statistically significant differences displayed by the individual muscles and mm. soleus and gastrocnemius have the highest numbers. The only muscle pair, for which all six spectrum parameter differences are statistically significant, is m. gastrocnemius *vs.* m. trapezius.

The detailed investigation of needle position influence, which was carried out for the brachial biceps muscle revealed, with minor exceptions only, no statistically significant differences. No significant differences at all were found between the average values of the long head and the average values of the short head.

The comparatively small coefficients of correlation between spectrum parameters indicating low-frequency content and parameters indicating high-frequency content may be a reflection of the existence of different types of motor units.

Also on the left-hand side muscles of the uninjured controls, the statistical analyses of variance showed significant differences between the mean values of individuals as well as of muscles. This result holds for all spectrum parameters. As for the right side muscles intramuscular variation is most important, intermuscular variation is next most important, and interindividual variation is least important. The number of statistically significant intermuscular differences is larger for the left side than for the right side. It is interesting to note that statistically significant parameter differences between the left and right sides of mm. interosseus dorsalis I manus, extensor digitorum brevis, biceps brachii, brachioradialis and extensor digitorum communis take place in the same direction.

Temperature measurements revealed no discernible temperature dependence of the spectrum parameters in the long head of the brachial biceps muscle of the 50 uninjured controls. Temperature influence can neither explain the power spectrum differences between two needle positions in this muscle, nor the difference between the left and right sides.

Comparison between muscles yielding normal and neurogenic EMG in amputees was performed for the right and left side, respectively, of the stump extensor muscles and for the right side of the stump flexor muscle. The analyses revealed statistically significant differences for all spectrum parameters, except for one of the right side flexor muscle. Comparisons between muscles yielding normal EMG in amputees were carried out for mm. biceps brachii, brachioradialis, and the stump extensor muscle of below-elbow amputees. The analysis showed statistically significant differences between the mean values of muscles for all parameters, except for one parameter of the left side. When muscles of amputees yielding normal EMG and muscles of uninjured controls were compared, statistically significant differences between the parameter values were obtained. The significant differences for right side muscles deviated in the same direction as those of left side muscles. The low-frequency content of the amputee muscles was, without exception, low. Temperature measurements were performed in the stump extensor muscle of 14 below-elbow amputees. The low frequency content tended to decrease with temperature.

Dynamic Spectrum Analysis of Myoelectric Signals in Amputees

The dynamic spectral analysis of myoelectric signals from four different muscles in 27 arm amputees is tabulated in Tables 14 and 15. The percentage of recovered muscles was calculated on the number of cases where a significant effect of the high frequency decay was found. The percentage of non-affected muscles was calculated on the total number of the muscles examined. The values, obtained at examination of the amputees, was related to the corresponding normal values, earlier indicated by *Kadefors, Kaiser & Petersén* (1968) except for the muscles on the flexor side of the stump where normal values were lacking (Table 15).

Table 15 shows that there was a difference between the remaining extensor muscles of the forearm stump and the corresponding muscles in normal subjects with respect to the decay of the high frequency components. In 53 per cent of the amputees there was no significant decay as compared to 5 per cent

Table 14. Values obtained from amputees. Muscle effect and recovery after 0, 30, 60, and 90 seconds, respectively.

Effect and recovery	Stump muscle extensor side	Stump muscle flexor side	M. brachio-radialis	M. biceps brachii
	-4 4 16 2	3 9 8 4	5 9 7 11	3 12 5 7
	8 7 3 14	4 3 2 1	5 4 8 -3	5 -1 12 14
	-1 2 4 4	15 6 6 6	17 15 15 10	9 16 11 8
D_5–$D_{1,2,3}$	17 8 2 5		10 17 5 3	4 15 14 15
	9 5 3		19 8 6	4 2 9 12
				21 3 6 19
				2 1
	-8 0 -3 2	-1 -2 3 0	0 1 0 3	0 -1 -1 3
	-5 -3 -1 4	2 1 -1 0	3 3 -5 -3	3 -2 -1 -2
	-1 0 -1 4	2 -1 4 0	0 0 4 0	-2 7 7 -5
D_5–$D_{6,7}$	4 1 1 0		-2 4 -5 2	1 2 4 4
	0 0 -2		5 0 0	-5 2 -2 0
				-1 2 0 9
				1 2
	-9 7 7 -1	1 5 3 1	-1 7 -1 7	0 6 5 4
	-1 2 4 10	10 -2 3 0	5 4 1 -2	3 -3 4 9
	-3 -2 2 3	5 3 8 4	12 14 15 4	5 7 10 -3
D_5–$D_{8,9}$	13 3 3 2		4 18 -5 2	2 13 6 7
	10 2 1		11 8 3	-5 3 4 4
				5 1 1 19
				2 2
	-7 10 10 2	2 8 1 1	-6 8 1 9	1 11 4 8
	0 1 5 10	7 -1 1 0	2 0 2 0	3 -1 4 14
	0 2 4 4	6 3 6 4	7 11 13 6	5 10 11 -3
D_5–$D_{10,11}$	14 2 3 3		4 16 -3 2	4 13 6 6
	6 1 1		14 9 7	-2 1 4 5
				8 0 4 18
				3 2
	-5 12 12 2	1 9 3 -1	-8 7 3 10	2 12 3 12
	2 0 4 6	10 -1 1 0	3 -1 9 1	5 -2 5 17
	-1 3 4 7	8 3 7 6	11 11 12 6	7 10 11 -4
D_5–$D_{12,13}$	17 2 4 4		7 13 -3 1	6 13 16 8
	9 1 2		14 9 7	1 0 4 6
				9 1 5 18
				2 1

Table 15. Effect and recovery with respect to the high frequency band decay. Results from normals by permission from the authors (*Kadefors, Kaiser & Petersén*, 1968).

Investigated muscles	D_5–$D_{1,2,3}$ median dB	Immediately recovered per cent	30 s recovered per cent	60 s recovered per cent	90 s recovered per cent	Non affected cases, per cent
Stump muscle extensor side	4	0	56	56	56	53
M. extensor digitorum in normals	11	10	32	42	42	5
Stump muscle flexor side	5	17	67	67	67	50
—	—	—	—	—	—	—
M. brachioradialis in amputees	8	6	50	56	75	16
M. brachioradialis in normals	11	6	22	45	61	10
M. biceps med. in amputees	8.5	17	33	50	56	31
M. biceps med. in normals	10	12	55	50	45	5

in the normal series. The significant effect revealed in the extensor muscle by the amputees, however, was quite comparable with that in the normal series (Fig. 9). There was no significant difference either in recovery. In this figure the distribution of the normal series was plotted, including values within a limit of significance corresponding to 2 σ. The figure shows that more than half of the values obtained from this muscle in amputees lie closely to the origin of coordinates. Despite the powerful contraction found on clinical examination, and the good quality of the signals as observed with the naked eye, these patients obviously could not contract the remaining muscle as much as normal subjects. Furthermore, one case with a muscle effect of —4 dB was excluded from the figure as an adequate contraction was not achieved. In principle, the same result was obtained at the examination of the remaining muscles on the flexor side of the forearm stump as on the extensor side. Thus, in 50 per cent of the amputees there was no significant reduction of the high frequency components in the myoelectric signals picked up from this muscle. In cases where a significant effect was obtained, the recovery could not be compared with that of a normal series, but in principle followed the same

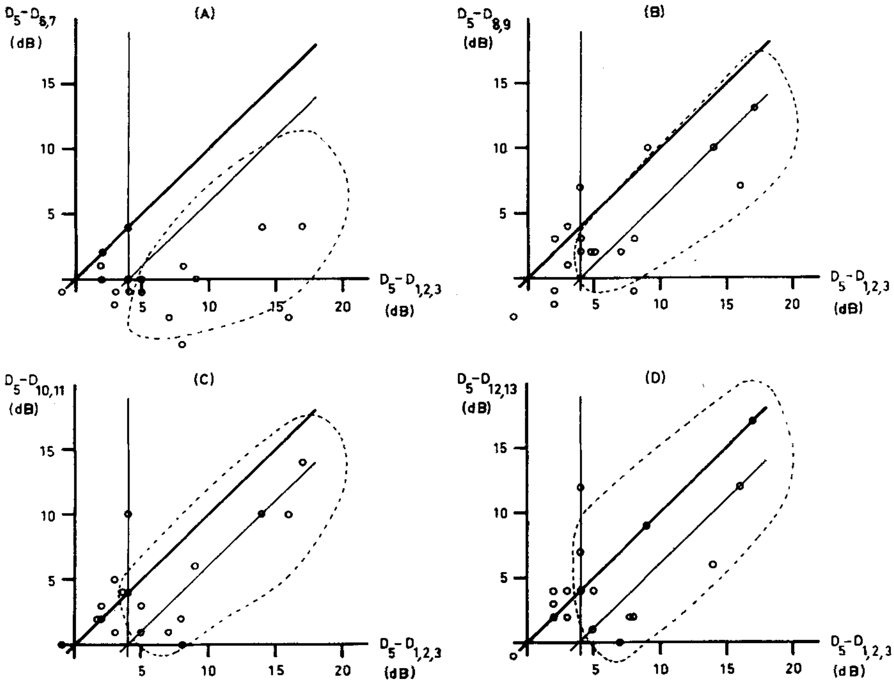

Fig. 9. Recovery of the high frequency decay occurring under the influence of maximum muscle contraction. (A) immediately after the sustained contraction, (B) after 30 seconds of rest, (C) after another 30 seconds of rest, (D) after still another 30 seconds of rest. Stump muscle on the extensor side of the forearm, needle electrodes. Open circles represent 18 below-elbow amputees. The dotted line encircles normal values with a limit of significance 2 σ.

pattern as the three forearm muscles, examined on normal subjects by *Kadefors, Kaiser & Petersén* (1968).

The brachioradial muscle was studied on 19 arm amputees, the number of which corresponded to that of the normal series. There was no significant difference at the 5% level in a T-test between the amputees and the normal subjects either with respect to the effect or the recovery. The median value of the high-frequency decay was 8 dB for the amputees and 11 dB for the normal subjects. The number of non-affected cases was 16 per cent for the amputees and 10 per cent for the normal subjects. The effect seems to be lower for the amputees than for the normal series and the difference is almost significant (Fig. 10). The distribution of the normal series with a limit of significance corresponding to 2 σ has been plotted. The figure shows that three values for the amputees lie on or below the 4 dB limit of significant muscle effect. In two of these cases there was advanced atrophy of the brachioradial

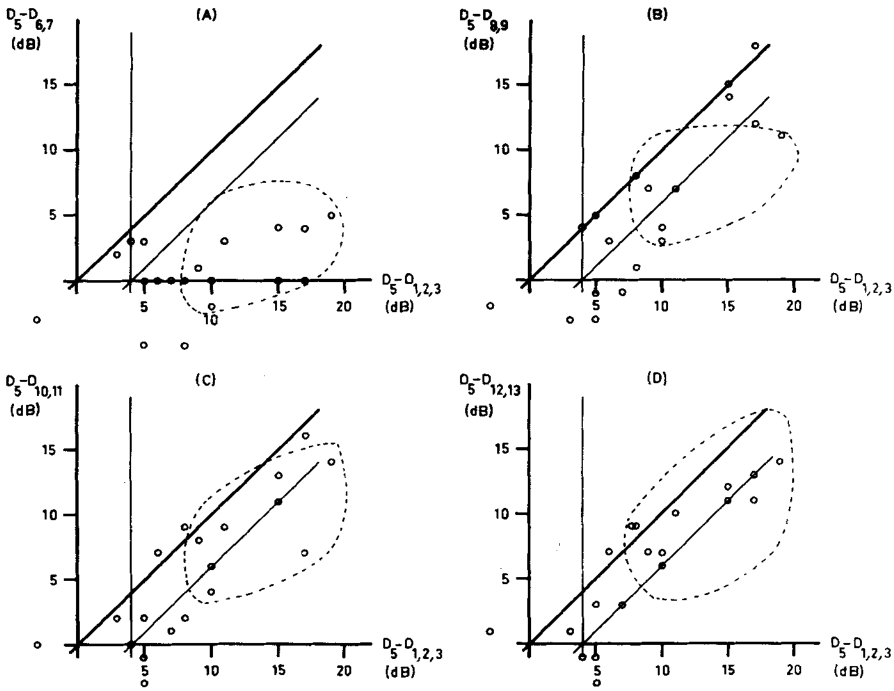

Fig. 10. Diagram identical with Figure 9, but showing effect and recovery in the brachio-radial muscle of 19 below-elbow amputees, needle electrodes. The dotted line encircles normal values with the same limit of significance, 2 σ.

muscle and a very reduced strength which involved difficulties for the patients to maintain maximal contraction during 30 seconds.

The brachial biceps muscle was studied in 26 amputees, five of whom were above-elbow amputated. As the group of above-elbow amputees was very small, the observations were not referred to different levels of amputation. Table 15 shows that in comparison with the normal series there was no obvious difference in the high frequency decay. The muscle effect was 8.5 dB in the amputees and 10 dB in the normal subjects. A T-test did not reveal any significant differences at the 5% level in effect and recovery (Fig. 11). In 31 per cent of the amputees, there was no significant muscle effect as compared to 5 per cent in the normal series. Among the five examined above-elbow amputees, three revealed no significant muscle effect, which partly contributed to the high percentage of non-affected cases. In two of the five remaining amputees, where there was no high frequency decay of the biceps muscle, there was a pronounced reduction of strength at flexion of the elbow-joint. In three of the five cases, no observations of any importance were made. This gives a

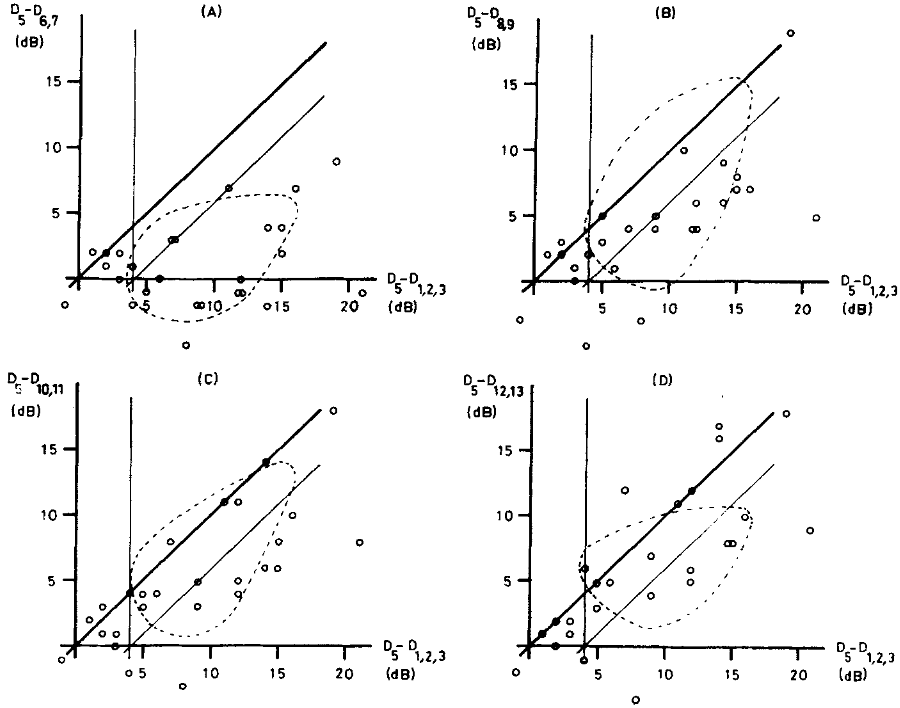

Fig. 11. Diagram identical with Figure 9, but showing effect and recovery in the brachial biceps muscle of 26 above- and below-elbow amputees, needle electrodes. The dotted line encircles normal values with the same limit of significance, 2 σ.

percentage of cases, without any significant muscle effect, which corresponds to the normal subjects.

The results of the four muscles, studied in arm amputees by the use of this method, are not referred to the EMG findings. There was no point in making such a classification of the brachioradial muscle, due to the few cases of pathological EMG findings. No significant difference was noted in the muscle effect and recovery of the muscles on the flexor- and extensor sides of the forearm stump and of the biceps muscle, as based on the EMG findings. As earlier, these conclusions were made after performing a T-test with the 5% level of significance.

Comparison between Electrodes and Implantation of Micro-circuits (Papers II, III)

The impedance of the three types of surface electrodes, utilized in electrical hand prostheses, were measured in different frequency ranges. The lowest impedance was found in the Russian electrode which is due to the use of electrolytic paste. The other electrodes, the Austrian and the Italian, are dry electrodes. The Italian electrode has a much higher impedance as compared to the Austrian as a consequence of its smaller area. All three types of electrodes have a rather high impedance, which at 100 Hz exceed 100 kohm. The phase angle within the different frequency ranges has varied least in the Russian electrode.

The difference between the half-cell potentials of the electrode pairs were measured. The maximum value of the polarization difference, in a series of ten measurements was given for each type of electrode. It was obvious that wet electrodes had considerably higher polarization potential differences than dry electrodes, depending on the abundance of dissociated ions, which exist with the use of electrolytic paste. The polarization potential measured between the signal electrode and its ground plate in the Austrian electrode, was considerably higher than the half-cell potentials of the signal electrode pair. This fact was explained by the signal electrodes and the ground plate being made of different metals, which have different contact potentials.

The signal-to-noise ratio in dB of signals from the electrodes under test was measured. The centre point distance of different electrodes and the area of the electrodes had a direct effect on the recorded signal-to-noise ratio. Thus, this ratio was the most favourable for the Austrian electrodes with the largest centre-point distance and the largest picking up area. A large centre-point distance will give a higher signal potential and, furthermore, signals from more motor units can be picked up with a large area of the electrode.

Spectral analysis of the signals picked up in the signal-to-noise investigation showed small differences between the spectra of the individual prosthesis electrodes.

Using intramuscular electrodes, needle electrodes and wire electrodes, a higher signal activity within the high frequency ranges was recorded as compared to skin electrodes. An implanted electrode had a power spectrum in between the intramuscular and the skin electrodes.

Six electrodes were implanted and worn by subjects for varying periods of up to 15 months (Paper III). The volunteers did not experience the short, surgical procedures, implantation and extirpation of the electrodes as either unpleasant or painful. All the patients returned to full work the day after. The wounds healed without complications and the amputees could utilize their prosthesis without difficulty for their work and without the pressure from the prosthesis causing any discomfort. The final scars were thin and cosmetically quite acceptable.

After removal of the electrodes, they were all found to be encapsulated by a connective tissue membrane, sometimes slightly adherent to the skin but never to the subjacent fascia. Macroscopically, it was found that the covering membrane sometimes was thin and pale and in others hyperaemic and thickened. The varying degrees of slight hyperaemia which could be found, were not related to the period during which the electrodes were implanted. The histological examination revealed a capsule of connective tissue which had developed around the electrodes in varying degrees of thickness. There were layers of histiocytic cells on the surface and some foreign-body giant cells, and diffuse infiltration by lymphocytes as a sign of inflammatory reaction.

The myoelectric signals, received from the implanted electrodes, were all except in one case, of a good quality. The reason of failure in this case was that a break-down in the electronic circuit prevented the electrode from operating after the implantation. According to conventional electromyography, the transmitted signals were normal and of a high quality. Repeated analyses proved that the myoelectric signals had been well reproducible during the periods of the experiments. This fact was also found by frequency analysis of the myoelectric signal. Even after several months, the shape of the power spectrum was unchanged at different levels of contraction. As compared to needle electrodes a more favourable signal-to-noise ratio was obtained with the implanted electrodes.

The main problem of protecting the components of the micro-circuit from surrounding tissues has been solved satisfactorily by the selected method of encapsulation. No technical failure was due to insufficient encapsulation in this investigation. However, there were two cases of small cracks in the plastic material which allowed contact between the semi-conductors and the surrounding tissue. The main failure was fatigue fractures in the gold wires at the surface of the plastic layer. These fractures prevented the myoelectric signals from being transmitted in three of the four cases where the electrodes had stopped operating. The fractures of the gold wires were verified radiographically. In one case the myoelectric signals could not be received due to defects in the external system. The external power supply was reliable. However, the transmission of energy to the implanted electrode was rather sensitive to the relative positioning of the implanted and the external circuits.

Myoelectric Control of the Electro-Mechanical Apparatus (Paper IV)

The results will be given in accordance with the questions discussed in the Chapter of Methods. Table 16 illustrates the off-times of nine muscles in a normal series of 30 uninjured males. The importance of learning in this type of test is illustrated by the difference between the off-times of two successive tests of the same muscle. The results in Paper IV were thus verified between the two investigations with the exception of a small deviation of the mean in

Table 16. Off-times for uninjured males in seconds.

M. extensor carpi radialis I	M. extensor carpi radialis II	M. flexor dig. sublimis	M. inter-osseous dorsalis	M. biceps brachii	M. triceps	M. del-toideus	M. trapezius	M. vastus lateralis	M. tibialis anterior
52	48	7	8	25	52	2	63	18	3
66	18	3	38	10	10	5	80	5	6
61	7	20	78	7	21	4	176	92	11
67	13	10	116	24	55	40	75	65	25
104	2	42	62	90	23	33	91	25	21
81	27	8	5	3	3	3	139	68	8
117	26	16	2	14	9	14	109	12	18
60	14	10	10	71	3	3	7	60	35
79	45	7	59	70	3	21	21	11	23
124	94	20	37	45	16	79	97	48	11
55	5	110	2	33	23	6	136	69	13
144	33	21	2	17	10	39	130	2	8
46	16	9	36	7	15	36	106	45	17
73	8	4	2	3	8	14	7	10	16
80	13	16	43	6	22	25	99	17	3
127	46	40	29	36	27	15	53	86	6
46	9	36	67	28	17	14	109	29	5
41	4	40	37	27	11	76	35	2	12
6	5	3	32	10	16	37	155	5	31
31	16	3	26	175	53	50	205	16	4
89	2	7	7	58	15	3	57	74	15
58	31	6	51	7	14	12	70	34	16
54	26	30	43	45	38	30	68	68	4
101	69	7	35	28	47	125	120	36	10
61	4	10	12	63	10	59	184	12	27
50	13	35	86	8	23	21	62	20	23
129	18	19	84	24	18	48	127	7	26
46	40	6	43	20	25	14	94	79	31
4	19	14	19	27	7	70	38	15	2
118	12	46	107	17	50	30	78	65	8
Mean:									
72.3	22.8	20.2	39 .3	33.3	21.5	30.9	93.0	36.5	14.6
Standard deviation:									
34.9	20.7	21.2	31.2	34.5	15.4	28.0	49.2	28.3	9.4

test I. This deviation may depend on a modified instruction to the patient before the definite examination. By changing the patient's position at the examination (see methods) obvious deviations were caused in the off-times of the deltoid muscle on comparison between the two examinations. The test conditions were the same and comparable for m. extensor carpi radialis longus and for m. vastus lateralis. The off-times of these muscles— obtained at the two examinations—are in good accordance.

Table 16 shows that the trapezoid muscle deviated considerably from the other muscles. Furthermore, it is indicated in the table that there is a tendency towards lower mean values of off-time for distal extremity muscles as compared to proximal, with the exception of the triceps muscle. However, it should be emphasized that there was a large dispersion for singular off-times. The unexpectedly high mean value of the first dorsal interosseous muscle can perhaps be explained by the fact that it proved fatiguing and unnatural to abduct the forefinger continuously during several minutes.

The analysis of variance showed significant intermuscular and interindividual effects. The significance of the individual effect was weak, however. In

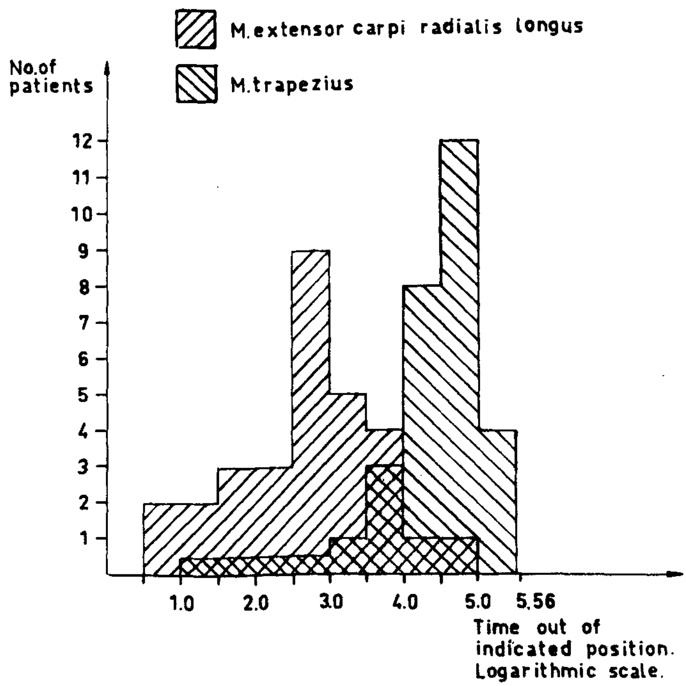

Fig. 12. Histogram showing the significant difference in off-times in the manoeuverability test between the m. extensor carpi radialis longus and the m. trapezius of uninjured controls. The histogram is based on the natural logarithms of the off-times.

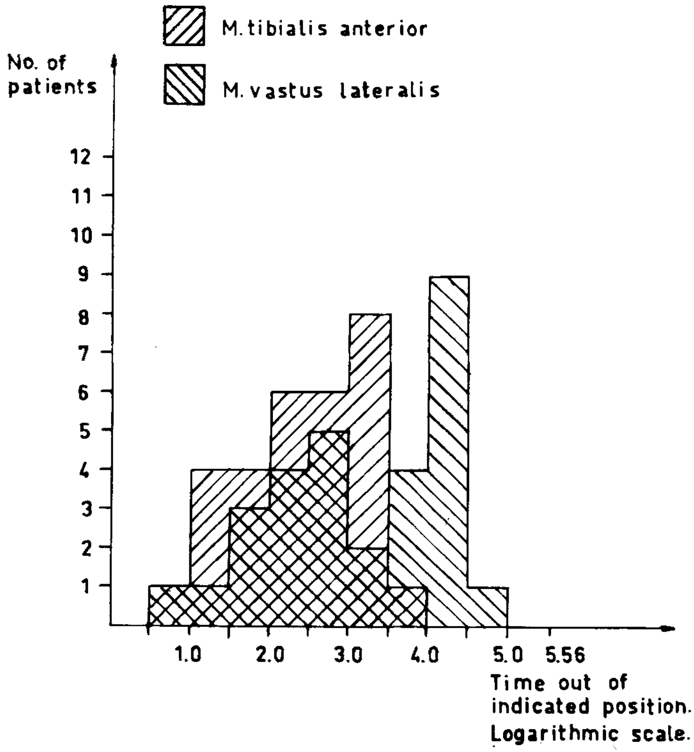

Fig. 13. Histogram indicating a certain, but statistically non-significant difference in off-times between a distal and a proximal extremity muscle of uninjured controls in the manoeuverability. Based on the natural logarithms of the off-times.

the first place this meant that the means of the natural logarithms of the off-times of at least one muscle, deviated from one or a few of the other muscles, and secondly that there were also deviations between different individuals.

The analysis of variance did not show which mean values deviated from each other thus contributing to the significant muscle effect. When comparing different types of muscle contrasts it was found that the trapezoid muscle only had a significantly deviating mean value. In a histogram (Fig. 12) this deviation is clearly demonstrated. The mean values of the other muscles were approximately on the same level. Thus, there were no significant differences in the natural logarithms of the off-times between the proximal muscles (mm. biceps brachii, deltoideus, vastus lateralis) and the distal muscles (mm. extensor carpi radialis, flexor digitorum sublimis, tibialis anterior). There is a certain difference in the natural logarithms of off-times between the anterior tibial muscle and the lateral vastus muscle, when illustrated in a histogram (Fig. 13).

Table 17. Off-times for amputees in seconds. Mean and standard deviation are referred to below elbow amputees, exclusively.

	Stump muscle extensor side forearm		Stump muscle flexor side forearm		M. biceps brachii		M. deltoideus	
Amputation level	Normal EMG	Lower moto-neuron lesion	Normal EMG	Lower moto-neuron lesion	Normal EMG	Lower moto-neuron lesion	Normal EMG	Lower moto-neuron lesion
	79		85		83		88	
	21		7		36		119	
		24	16		6		22	
	3		6		13		85	
	27		2		6		32	
	5			6	10		42	
	8		18		15		37	
	114		179		95		60	
		21	4		30		93	
	15		8		39		51	
	2		30		50		74	
Below		146		20		63	63	
elbow		7	67		81		42	
	3		20		42		151	
	7			20	46		59	
		68	55		67		106	
		43				26	27	
	62			16	52		52	
		5		3	115		15	
						4	30	
	25		7		119		42	
		42		47	42		45	
	42		21		23		35	
						10	23	
						56	208	
Above					17		34	
elbow					5		47	
					11		108	
						82	80	
Mean	29.6	44.5	35.0	18.7	43.3	31.0	59.6	—
	35.0		30.3		41.7		59.6	
Standard dev.	37.4		39.8		30.6		33.2	

The significant deviation—although slight—between different individuals, verifies the clinical impression obtained during the experiments. A difference in dexteriority was thus found, but no systematical individual factor of importance was found at the statistical analysis.

The results of the examination in 30 arm amputated males are seen in Table 17. This table reveals that there were rather few values available for comparison between muscles with a normal EMG finding and those with a peripheral neuron lesion. The dispersion of the individual off-times is large. The statistical analysis showed no significant differences with respect to the EMG findings for remaining extensor and flexor muscles and the brachial biceps muscle, in below-elbow amputees. In above-elbow amputees, the brachial biceps muscles could not be investigated statistically with respect to the EMG findings, due to the small number of observations. Thus, the off-times, obtained for the above-elbow amputees, from the brachial biceps muscle, are excluded from the statistical analysis in a comparison between controls and amputees.

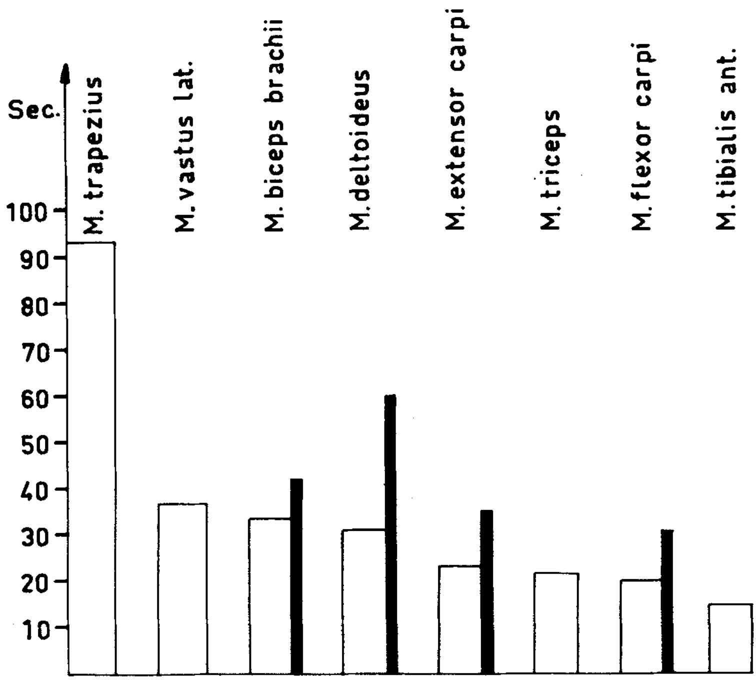

Fig. 14. Diagram showing the off-times in seconds obtained from different muscles in uninjured (open columns) and arm amputees (filled, thin columns). For further discussion see the text.

Besides the reservations made above, all the off-times obtained for the amputees were included in a comparison between uninjured controls and amputees as the EMG findings did not affect the manoeuverability. Table 17 shows that in the group of arm amputees, there were higher mean values of off-time for each of the four examined muscles as compared to uninjured controls. This is clearly depicted in Figure 14. The normal test of the natural logarithms of off-times showed, however, that this difference was significant only for the deltoid muscle.

CONCLUDING REMARKS

During the latter decade, myoelectric signals have been utilized in practice for control of electric hand prostheses and have, furthermore, served as control signals in stimulating paretic muscles. The investigations performed earlier involving a study of the myoelectric signal, plotting of the control sites available, and assessment of the ability to control man-machine systems, were based on normal subjects. However, the myoelectric signals, picked up from the patients who will utilize them in the future, must be studied and accurately characterized. In this work, the characteristics of the myoelectric signals in arm amputated males and normal individuals were investigated. The relative characteristics of different types of electrodes were examined. The description of the myoelectric signal thus obtained, permits the signal to be optimized by suitable filtering before electronic application in prosthesis control.

It is the high unilateral and bilateral arm amputees who are in great need of multiple control sites to control prostheses with several degrees of freedom. Clinical findings revealed, however, that the stump muscles of above-elbow amputees are atrophic and retracted and that the triceps muscle was absent in several cases. The shoulder muscles can also be considerably reduced. Asymmetrical shoulders and scoliosis of the thoracic spine are practically always present in above-elbow amputees, which was also pointed out by *Solonen* (1965). The result of these findings is, unfortunately that only a limited number of control sites are available in above-elbow amputees. It is possible that by attaching the stump muscles distally, in order to prevent retraction and atrophy (*Weiss*, 1960, 1966), the functional end result will be improved. If the stump is not too short in below-elbow amputees, the muscles which are not retracted will only be slightly reduced. Anyhow, it was found that these remaining muscles can be gradually retracted when they are continuously used to control an electric hand prosthesis. A myoplastic amputation procedure, where the muscles are attached distally, may also for this reason be more attractive. This procedure is believed to give a better stump with less pain, normalized circulation and temperature, less decalcification and a better skin (*Dederich*, 1963).

The impaired sensibility which was noted in this series—especially in the above-elbow amputees—must be considered if surface electrodes are to be used. Practical application of the electric hand prosthesis showed that the mechanical irritation, existing with surface electrodes, may cause sores if the sensibility of the stump is deficient. When the sensibility is normal and the subcutaneous tissue of the stump is satisfactory, dry skin electrodes will be tolerated even during long-standing daily use. In many below-elbow amputees, the remaining pronation and supination cannot be utilized due to technical fitting problems of the prosthesis. In order to offer the patient a functionally improved prosthesis, it would be desirable to achieve a normal positioning of the hand. That type of prosthesis should be possible to control with a minimum of effort and unconsciously after some time of practice. Perhaps an active pronation can be obtained with signals picked up from implanted electrodes in the m. pronator teres. Picking up separate activity from deep-lying muscles, a possibility offered by implantable electrodes, is perhaps more realistic than creating control sites by other surgical procedures.

The impaired mobility of elbow- and shoulder-joints in a number of amputees is not of a magnitude which does prevent the use of a prosthesis. The physical strength is only in singular cases reduced to such an extent that it would be of any practical importance, especially as the prosthetic hand always will be the assisting hand in relation to the normal hand.

The myoelectric signals picked up from different muscles in an amputated arm are often pathological according to conventional electromyography. The changes—a peripheral neuron lesion in a large number of muscles—which were recorded in this series, are in accordance with those found by *Petersén* (1966) in 50 arm amputees. The reason of this is probably due to the peripheral neuron being injured in connection with the trauma or at the operation. The extent of the pathological changes depends on the level at which the nerve has been damaged. Fibrillar action potentials rarely occurred which is due to the fact that most of the lesions antedated the examination by several years. In fresh, amputated stumps, denervation potentials probably appear much more often in muscles showing peripheral neuron lesions. In the remaining muscles of very short stumps, large single potentials have been found in a few cases. This is probably a sign of a so-called sprouting phenomenon, which means that growing terminal nerve fibres belonging to a certain motor unit will contact muscle fibres belonging to another motor unit. The number of muscle fibres in a motor unit of that kind will become larger at the same time as the spatial dispersion increases. In practical application of simple electric hand prostheses with on-off control, the interesting result was obtained that these large potentials were satisfactory as control signals. This is the more remarkable since only single potentials could be detected even when surface electrodes were used.

An increased motor unit dispersion due to sprouting, results in more polyphasic potentials. Inactivation of motor units and synchrony of remaining units may according to *Pinelli* & *Buchthal* (1953 *b*) also contribute to the abundance of polyphasic potentials in lower motoneuron lesions. The low intramuscular temperature of the stump may also contribute to the increased number of polyphasic potentials by non-synchronization of otherwise simultaneously activated muscle fibres within the same unit (*Buchthal, Pinelli* & *Rosenfalck,* 1954).

In one single patient, an EMG pattern of myopathy displaying a large number of short duration potentials and an excess of polyphasic potentials was observed. This patient probably had myositis in the stump muscles due to the serious arm infection which caused the amputation.

In the stump muscles, potentials have also been observed with a certain—subjectively assessed—increased duration. These potentials were found when the intramuscular temperature was low. Several authors have shown earlier that a reduced intramuscular temperature causes an increase in the duration of the motor unit potential. This was believed to result from a decreased conduction velocity along the terminal branches of the nerve fibres and the muscle fibres (*Pinelli* & *Buchthal,* 1953 *b*). The low intramuscular temperatures which are found in amputation stumps, probably depend on a change in the circulation, and comparatively reduced muscle bundles. *Eriksson* (1965) showed by means of angiography that the blood supply to leg amputation stumps may be irregular and that the muscle volume is reduced. The same author mentioned that the amputation stump, painful to the patient, had a greater mean blood supply at rest, as compared to the cases without pain. Furthermore, he pointed out that the abundance of serpentine vessels did not necessarily mean that the tissues are well nutritioned, but that these vessels may act as shunts. This is also in accordance with the investigations of oxygen tension, pO_2, present in the venous blood of amputation stumps (*Hulth* & *Högström,* 1961).

An essential point in myoelectric control of externally powered prostheses and orthoses is the dynamic range of the control, *i.e.* the number of well-defined separate levels of the output. Large dynamic output range requires that the full contraction range of the control muscle be utilized. Investigations on the properties of the myoelectric signal can thus not be limited to the individual motor unit potentials, since these are discernible at the very weak contractions only. Even a slight increase of muscle contraction causes the myoelectric signal to take on the character of random noise. Thus the control properties of the myoelectric signal can be characterized throughout the dynamic range by means of its power spectrum.

Two definite methods have been used for measuring power spectra of myoelectric signals from a great number of muscles. With the aid of one method, *Kadefors, Kaiser* & *Petersén* (1967, 1968), modifications of the power spectrum due to sustained, maximum muscle contractions were studied on amputees. The present investigation shows these modifications to be of such a magnitude that they should be taken into account in the design of myoelectrically controlled prostheses.

The other method, *Kaiser* & *Petersén* (1963, 1965), has been used in an extensive study (Paper I) of power spectra for moderate muscle contractions of short duration. This investigation demonstrated the existence of statistically significant intramuscular, intermuscular, and interindividual power spectrum differences. The most important source of the differences consistently was intramuscular variation. This discovery further underlines the necessity of preprocessing the myoelectric signal in suitable filters. It also implies that individual alignment of these filters may be necessary in order to achieve optimum control.

At an investigation of three commercially available skin electrodes in myoelectric hand prostheses, characteristics such as impedance, polarization instability, signal-to-noise ratio and spectral properties were studied. In general it can be said, that skin electrodes used in the control of myoelectric prosthesis should have a low impedance, low polarization instability, high signal-to-noise ratio, and that the reproduction of the signal activity within high frequency ranges should be great. Furthermore, the chemical and mechanical irritation of the skin should be insignificant. It is obvious that the electrodes which were available at this investigation, did not fulfil these demands.

The results of micro-circuit implantation show that the minor surgery required was easy to perform and caused no subjective discomfort. Implantation in the human being of the plastic material selected can thus be tolerated for at least one year. Due to the tissue reactions, it would be justified to carry out further research work in producing more suitable material. Silicone rubber (Paper III) or ceramics (*Reilly*, 1968) have been discussed. From the technical point of view, no failure was due to insufficient protection of the components either mechanically or chemically. However, there were small cracks in the plastic material in some cases which may have increased the tissue reaction. The myoelectric signals, transmitted via the implanted electrodes, were of a high quality and reproducible during long periods. Consequently, it is obvious that the signals are not affected by the electrode being encapsuled in a connective tissue membrane.

The important failure found in this investigation was due to fatigue-breakdown of the gold wires which maintained contact with the tissue. This problem can perhaps be overcome by using thin flexible wires instead. As a result of our experience a modified electrode of that type has been designed. The favourable signal-to-noise ratio depends on the low noise contribution obtained by this method. Consequently, the implanted electrodes are superior to the conventional ones especially at low contraction levels. A limited band-width of the receiver has caused a reduced signal activity within high-frequency ranges. An improvement can be achieved without difficulty by increasing the band-width of the receiver. Both from the technical and the biological points of view, the existing miniaturization was adequate. Security of the function is, however, necessary during a considerably longer time, which should be the aim.

When considering the function of the electro-mechanical apparatus for the manoeuverability test, it can be assumed that the position of the fly-wheel is characterized by an approximately normal distribution. The assumption that the position of the wheel is normally distributed gives a mathematically univocal relation between the off-time recorded and the standard deviation. The instabilities of the input signals of the apparatus can be described by their standard deviations (standard deviation of programme signals=0). The relation between the input and output standard deviations are functions of frequency inasmuch as rapid variations are damped and slow variations are stressed by the apparatus. This can also be described as a time-delay of the apparatus. It is a general control system experience that time-delays tend to create oscillations. The ability to manoeuver the apparatus can partly be regarded as the ability to predict the next phase, which is a cerebral function. The apparatus obviously does not separate the cerebral component and the muscle effect, but measures a combination of these two factors. As a designation of this complex function, the term manoeuverability was introduced. The above reasoning reveals a relation between manoeuverability and recorded off-time.

The analysis of the ability to manoeuver an electro-mechanical apparatus by myoelectric signals shows that the trapezoid muscle differs from a number of extremity muscles. In that connection it should be noted that this muscle participates in the gross activities of the statics of the body. The endurance of the muscle is greater than its ability to perform rapid and precise movements. The difference in the number of muscle fibers per motor unit and in the existence of functionally different types of muscle fibers may explain this result of the trapezoid muscle. The moderate pathological changes, present in the muscles of several amputees, do not affect the manoeuverability. These remaining stump muscles are thus cerebrally well controlled, despite many years having elapsed since the normal function of the muscle ceased.

Arm amputees have greater difficulties to manoeuver a man-machine system with certain muscles as compared to normals. This can partly be explained by less daily activity and less training of the muscles on the amputated side.

There is, no doubt, a changed sensory information, when part of an extremity is missing with an impaired ability to move the extremity with proximal muscles. This should be borne in mind as simulating tests with man-machine systems are generally performed on normal subjects in prosthetic research. The results of these tests are not relevant for amputees with respect to certain muscles. An analysis, necessary to obtain a myoelectrically controlled prosthesis with several degrees of freedom, concerns the ability of a patient to utilize a number of control sites, independently of each other. The activity, present under normal daily and professional conditions in the muscles, possible as control sites, must be investigated. These complex functions must be subject to research before the possibility of controlling a prosthesis with signals from single motor units can be established.

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