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BIOELECTRICAL CONTROL IN A SERVO-SYSTEM

Analysis and Application of Muscle Action Potentials in an Experimental Hand Prosthesis

By

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By incorporating a motor and source of current into a hand prosthesis, a system can be created for performing a programmed work. It is then a matter of engineering to make function to subordinate to will. The motor's source of current must therefore in some way receive impulses from the neuromuscular system.

The questions we have asked ourselves have been:

1. Do myopotentials exist in an amputated arm some time after the amputation has been performed and are these suitable as impulses for the servo-hand?
2. Can a motor be constructed that is sufficiently small and light to be inserted in a hand prosthesis and still be effective enough to perform useful work?

The problems outlined above have been studied in several research centres and hand prostheses have been constructed which have functioned during laboratory tests.

In our efforts to develop practically useful prostheses, we have drawn up the following ideals:

1. The myopotentials should be derived so that a permanent and steady arrangement results which does not harm the patient.
2. The servo-hand should be so constructed that its functional scope

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can be simply adapted to cover a sufficiently wide register of movements and power output.

3. The impulses should preferably not be initiated only by the will but also be regulated or in some way quantitized in order to make a wider functional scope possible. The amputated person should be able both physiologically and psychologically to adapt himself to the system.

When controlling a prosthesis it is important that the control signals are "natural" for the individual in question. One type of such natural signals can be taken from the mechanical activity of another muscle which is intact or from the muscle stump of an amputated extremity via transducers. The fact that there is a proportion between the mechanical activity of the muscle and its action potentials is a prerequisite for the use of action potentials when controlling a prosthesis. This had also been taken advantage of, as may be seen from works by *Battey, Nightingale & Whillis,* 1955, *Kobrinsky et al.,* 1960, and *Horn,* 1963.

Muscle action potentials derived from the muscle of an extremity show considerable variations with regard to the amplitudes, even though conditions have been constant. These variations are so great that they might possibly render it difficult to use myopotentials in controlling a prosthesis. We have therefore wanted to investigate whether filtration of the myosignals could reduce this difficulty. Furthermore, we have wanted to get an idea as to the psycho-physiological possibilities of test persons to use their myopotentials quickly and correctly in a test programme. Finally, we have converted our experience into a servo-prosthesis, the actuator system of which shows force-velocity relations comparable to those that are to be found in a striated muscle.

RESULTS

1. *Frequently-amplitude Relation in Muscle Action Potentials.*

When skin electrodes are used, it is possible to derive muscle action potentials within the frequency range 25 Hertz to 1,000 Hertz and to obtain amplitudes from some microvolts to millivolts. Tre frequency range of the potentials varies in different muscle groups. Figure 1 shows the course of the frequency amplitude in myopotentials derived from M. biceps brachii. There is maximum activity in the range about

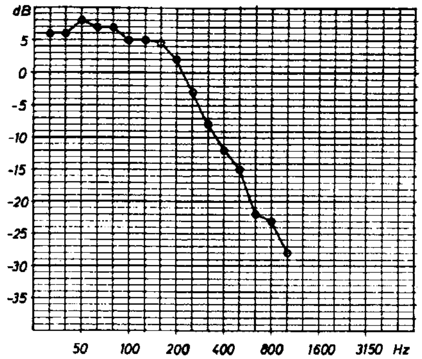

Fig. 1.

Frequency-amplitude responses obtained with a number of ⅓-octave filters. Abscissa: frequency in Hertz: ordinate: relative output amplitude in dB.

50 Hertz. At higher pass-band frequencies there is a strong reduction of the amplitude.

In a previous work (*Kaiser & Petersén,* 1963), an analytic method to obtain simplified data with regard to the frequency profiles of the myopotentials has been described. On the basis of the results of this work, an automatic broad-band frequency analyser for EMG was constructed. This analyser (to be published, *Kaiser & Petersén,* 1964) has been used in the present work utilizing primarily three one-octave filters.

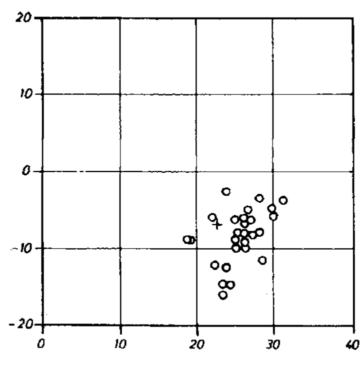

Fig. 2.

Shape of frequency profile obtained via three broad band filters (one octave). M. biceps brachii. Skin electrodes. Abscissa: Ratio between filter output 200 Hertz and filter output 800 Hertz, expressed in dB. Ordinate: Ratio between filter output 200 Hertz and filter output 50 Hertz, expressed in dB.

Fig. 2 shows a number of loci which indicate the form of some frequency profiles of the same type as in Fig. 1. As a reference amplitude, the area about 200 Hertz has been used. The abscissa shows the ratio between amplitudes in the 200 Hertz range and the 800 Hertz range, expressed in dB. The point marked + corresponds to the curve in Fig. 1. It can be observed that the activity at 800 Hertz lies 22 dB lower than the activity at 200 Hertz (the abscissa) and that the 200 Hertz activity lies 7 dB lower than the 50 Hertz activity (the ordinate). The circles in Fig. 2 show the frequency profiles in 27 derivations with skin electrodes from biceps muscles. It can be observed that the greatest amplitude reductions below 200 Hertz amount to approximately 30 dB, *i.e.* to about 3 per cent of the 200 Hertz activity. The greatest difference between the 50 Hertz and the 200 Hertz areas amounts to about 16 dB, *i.e.* the 50 Hertz activity can be up to six times greater than the 200 Hertz activity. The greatest difference between the 50 Hertz activity and the 800 Hertz activity is 40 dB, *i.e.* the 50 Hertz activity is about 100 times greater than the 800 Hertz activity.

2. *Choice of Proper Filter Range.*

The great amplitude reduction at high frequencies referred to above might indicate that the range about 50 Hertz would be the one most suitable for the purpose. There are two factors, however, which must be recognized at this point: *a.* the myopotential fluctuations of an erratic nature mentioned in the introduction and *b.* outside interference such as alternating current interference (power-line noise) of 50 Hertz and their harmonics. Both of the difficulties mentioned under *a.* and *b.* may be overcome by choosing a proper filter range.

Fig. 3 shows, in a logarithmic scale, the output voltages from four different filters which each have a band width of one octave. These voltages have been obtained logarithmically direct from the analyser. The uppermost curve shows the activity at 50 Hertz; curve no. 2 at 200 Hertz. Curve nos. 3 and 4 which partly coincide, show the activity at 400, resepectively 800 Hertz. The fact that these two lines partly coinside, does not mean that the two voltage levels are alike. Actually, the 800 Hertz level lies 11 dB lower than the 400 Hertz level. The test person has tried to keep the muscle contraction at a constant level during the course of ten seconds. The distance between the horizontal lines corresponds to a change of the potential level of about 40 per cent. The peak to peak deviations in the 50 Hertz filter correspond to

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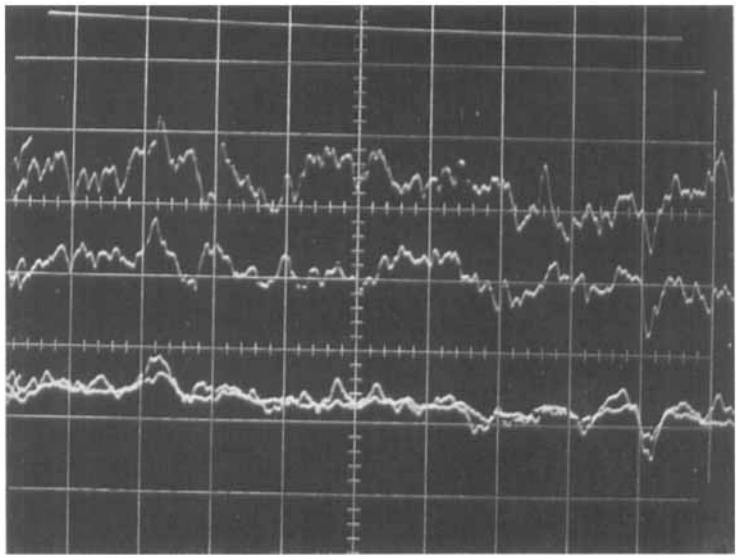

Fig. 3.
Logarithmic output voltages from four filtres; band with one octave. Logarithmic scale. Above: filter output for 50 Hertz centre frequency. Middle: filter output for 200 Hertz centre frequency. Below: two partly coinciding curves: filter output for 400, respectively 800 Hertz centre frequency. Abscissa: the distance between vertical lines = 1 sec. Ordinate: the distance between horizontal lines = 4 dB.

a ±20 per cent voltage change; at 200 Hertz ±12 per cent, at 400 Hertz ±7 per cent and at 800 Hertz ±4 per cent. This shows that filters with passband which cover 300–1,000 Hertz can reduce the tendency towards interference in a very efficient manner because the interference reduces more than the potential level itself when the frequency increases. As a result of this, the difficulty due to myopotential fluctuations have been reduced to a tolerable level. The improvement is probably due to the fact that the filters with higher passband frequencies give single potentials with shorter duration than the filters with low frequency, which reduces the chances of interference.

b. A.C. interferences, too, are controlled by choosing filters with passband. The cut-off frequency 300 Hertz protects against 50 Hertz A.C. interferences and against up to the 5th harmonic.

3. *Myoelectrical Level Changes in a Test Programme.*

The following tests were made in twelve normal test persons for the purpose of determining the possibilities of a normal arm muscle to

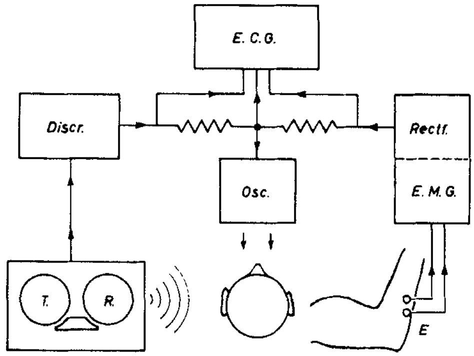

Fig. 4.
Schematic presentation of the test set up for programmed muscle activity under bioelectrical control. E = skin electrodes, EMG = EMG apparatus, Rectf. = rectifier for output potentials, OSC = cathode ray oscillograph in bridge coupling, ECG = ECG apparatus, Discr. = frequency discriminator, T.R. = tape recorder.

give myopotentials quickly and accurately in accordance with a predetermined programme. The investigation which is illustrated in Fig. 4 was carried out in the following manner: Skin electrodes, E, were attached to the forearm of the test person and were connected to a Disa EMG apparatus with a rectifier unit, which produced a D.C. voltage proportional to the muscle action potentials (not filtered). At the same time, a simple melody (Brother Jacob) was played on a tape recorder (T.R.) which was connected to a frequency discriminator, *i.e.* an electronic circuit which produced voltages that were proportional to the pitch of the tone played. These output voltages and the voltages of the myograph could compensate one another via a bridge circuit connected to a cathode ray oscillograph. The test person could both hear the melody and simultaneously follow visually on the screen of the cathode ray oscillograph how he compensated with his muscle activity the potentials reproduced in conjunction with the melody. All potentials were registered on an ECG apparatus. The results of such an investigation may be seen in Fig. 5. Curve A shows the time lapse of the output voltages of the discriminator which reflect a part of the melody. Curve B 1 shows the rectified and somewhat smoother myopotentials which are received from the test person when by visual control he tries to compensate the voltage niveau programme produced with the melody. Curve B 2 shows how the test person succeeds in trying to compensate. As may be seen, the deviations from the ideal result (a stright line) amount to 20 microvolts at a mean level of 50 microvolts. The deviation

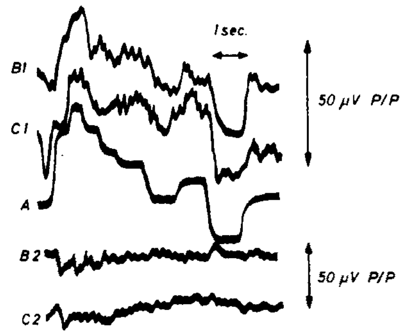

Fig. 5.

Time lapses of integrated potentials in a test programme. A = test programme. B 1 = course of the potential under visual control. B 2 = result of the compensation. C 1 = course of the potential without visual contdol. C 2 = result of the compensation. Abscissa: time. Ordinate: amplitude.

at lower levels (down towards 10 microvolts) was procentually just as great and had a reasonable proportion to those determinations which have been mentioned above (± 20 per cent). Curve C 2 shows the person's ability to reproduce his contraction programme when he hears the melody but has no access to visual control. The nature of the compensation is here of another type. In the successful tests, it can be seen (C 1) that the level changes without visual control are developed quicker but with less precision than with visual control. The result of the compensation may be seen by C 2, which shows less quick deviations but greater level changes than in B 2.

When no visual control is used, an overshooting occurs in the compensating activity at the onset of pitch changes. This activity pattern with its tendency to overshooting is actually advantageous for the utilization of a motor-operated prosthesis, as the initiated force should be greater than the stationary. Muscle potentials and, possibly, peripheral nerve action potentials have dynamic advantages in a prosthesis servo-system because they are in themselves intended for servo-systems and are therefore to be preferred to signals generated more central in the nervous system.

4. *Servo-prosthesis.*

The principle for the servo-system that we have arrived at is shown in Fig. 6. The myopotentials are derived by means of bi-polar skin electrodes. The registered potentials are amplified by means of a difference

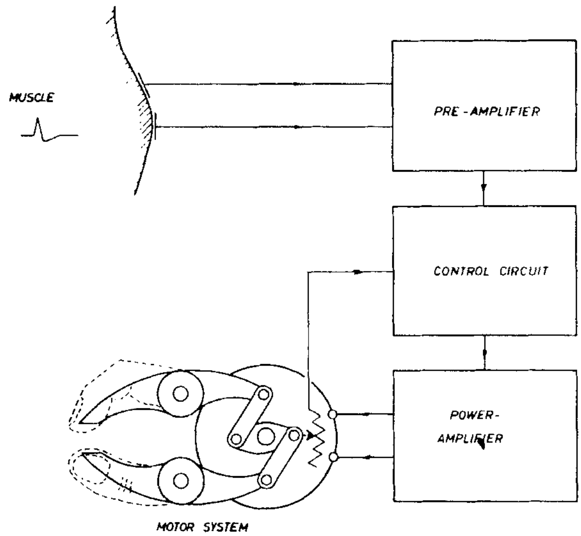

Fig. 6.
Schematic presentation of a bioelectric controlled servoprosthesis.

pre-amplifier with a range of 300–1,000 Hertz. Input impedance is 300,000 Ohm. The voltage gain is adjustable up to a maximum of 100 dB. The output impedance is 1,000 Ohm. The outgoing voltages are transferred to a servo-system, the control circuit of which contains a rectifier for the myosignals. The analog direct current component produced in this manner is led to a bridge circuit which compares the myosignals to a change of resistance created by the position of the prosthesis. If there is a discrepancy between the myosignals and the corresponding equilibrium position in the prosthesis, the bridge connection will produce d.c. potentials. These potentials are amplified in the power amplifier and are transferred to the servo-motor. This gives an automatic adjustment of the prosthesis in relation to the controlling signals. By means of the servo loop, the working range of the prosthesis may be adjusted according to need.

The static force of a grasp obtained through the current is within certain limits proportional to the difference between the actual position of the prosthesis and the position which the prosthesis would assume at the same activation but without outer load. The servo-amplifier produces voltages of up to ± 2.5 volts d.c. The current goes up to 600 milliampere. The maximum output effect is obtained at control voltages which exceed 1.5 volts d.c.

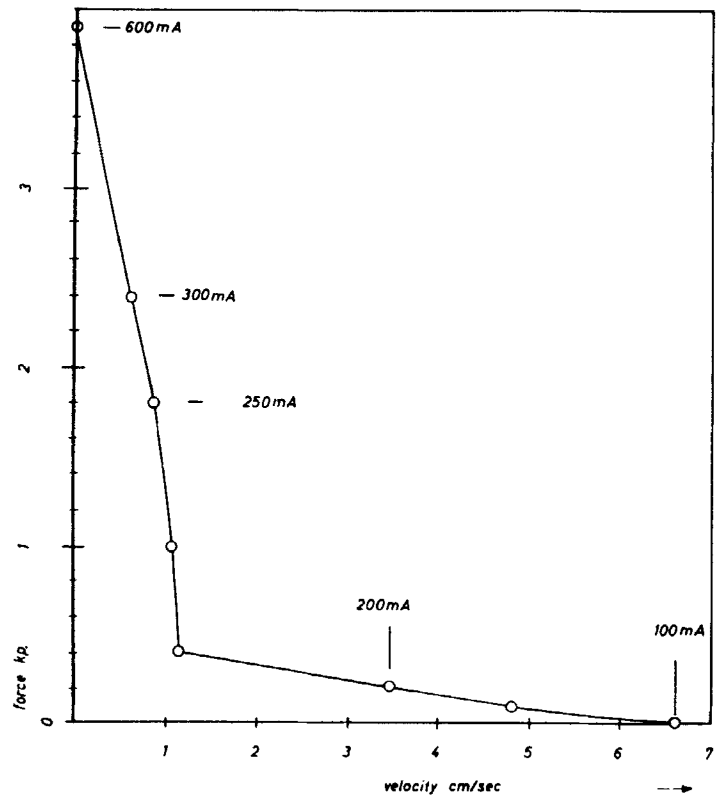

Fig. 7.
Force-velocity at maximum activation of the prosthesis motor system. Abscissa: velocity in cm/sec. Ordinate: force in kp.

5. *The Motor System.*

An ideal force-velocity relation would be obtained by using a so-called series motor of principally the same type as the one used for starting an automobile motor. Such a motor, however, is rather uneconomic in a miniature design. It is necessary to use motors with permanent magnets in order to obtain a reasonable electro-mechanical efficiency. The use of motors with permanent magnets means that the velocity-force relation will be less suitable for these purposes. In order to counteract this difficulty, we have used an automatic coupling to combine two miniature motors in such a manner that there is a suitable distribution of work between the two motors. Motor no. 1 produces the quick movements with relatively little force. Motor no. 2 additionally produces a relatively great force in combination with slow movements. During purely static conditions (*i.e.* when the prosthesis is fixed in a grip around an object) motor no. 2 need not be activated as motor no. 1 keeps the prosthesis locked with a minimum of current. A real relaxation is not possible until there is a change of the current direction in the motor system. By using this arrangement, there is an optimal electro-mechanical function both during dynamic and static conditions. Furthermore, a force-velocity relation similar to that of a striated muscle is obtaineed. Fig. 7 shows the current which is necessary to give maximum force at different speeds expressed in cm./sec. At the present transmission ratio the maximum force during purely static conditions is 3.9 kp. and that it is obtained at a current of 600 milliampere. In addition Fig. 7 shows that a maximum velocity of 6.6 cm./sec. without load is obtained at 100 milliampere. From the maximum velocity down to 1 cm./sec., a force which increases with reduced velocity up to about 0.4 kp. (400 gm.) is obtained. This part of the diagram corresponds to the effect of motor no. 1. The next part of the diagram for velocities below 1 cm./sec. (after the knee of the diagram) shows the effect of motor no. 1 + motor no. 2. Here one can see a smooth relation with additionally increasing force as the velocity decreases. The current at the point where motor no. 2 is just starts its contribution is about 200 milliampere. At lower activation levels, one obtains curves which are parallel to the one shown and which lie between the one shown and the two coordinate axes.

DISCUSSION

Regarding the Suitability of Electrical Signals in Servo-System.

The choice of signalling muscle activity via the action potentials or via the mechanical activity produced is dependent on several factors, for instance: *a.* the availability of the signals, *b.* its usefulness seen from the point of view of graduation and speed.

a. Regarding the availability of the signal, the prerequisites probably are the same whether one uses action potentials or the mechanical activity. Both skin electrodes as receivers of electrical activity as well as an external transducer for mechanical signals require a mechanical contact with the extremity. The possibility of outside interferences of a mechanical nature is probably greater when the mechanical transducer is used, as electrical interferences only can come into question when the electrical signals are to be received. Such interferences, however, can easily be avoided by means of a suitable filtration technique.

When implanted units are used for receiving mechanical or electrical activity, the mechanical problems of the transmission are reduced, and in both cases there is a possibility of obtaining a more undisturbed electrical transmission of both the mechanical and the purely electrical signals. In this case, there will be interferences only if the electro-mechanical transducer is subjected to an unintentional external pressure; the purely electrical system, howeever, is insusceptible in this respect.

b. The usefulness of mechanical and electrical signals from the point of view of graduation is expected to be somewhat better than is the case with the mechanical signals. On the other hand, the myosignals are quicker than the mechanical ones. In cases of level changes, a quickness via the myopotentials is therefore obtained, which is reduced only by the dynamic characteristics of the servo-system.

Regarding Electrodes.

In electromyography needle electrodes usually are used. Owing to the risk of pain and infection this type of electrodes of course is impossible to use in controlling a prosthesis.

Another possibility is skin electrodes. These, too, would involve considerable inconvenience when used for a prolonged period of time.

Another possibility which may be realized thanks to modern micro-electronics is to create a small electronic circuit for implantation under the skin so as to obtain a closer contact with the muscle. Such a circuit may be energized from the outside and can transmit electro-magnetic signals of high frequency through the skin. We are in the process of working on such a circuit.

Regarding Choice of Filter.

The potentials which can be registered by skin electrodes represent statistically a small part of the total activity in the different motor units of the muscle. It is therefore clear that among the few registered potentials, individual potentials will play a greater role in relation to the registered muscle activity. This is one of the reasons for the erratic nature of myopotentials derived from a small area. Another reason is the severe interference between the components with low frequencies. The chance of interference between potentials in different motor units grows with the quotient of the duration of the single potentials, and the time interval between action potentials. One would expect that an increase in the contraction force would give a relatively smaller erratic activity owing to the greater number of potentials per time unit, but unfortunately this is compensated by the interference mentioned previously. Therefore, we chose to investigate whether the interference phenomenon could be reduced by a suitable choice of filter.

We found that filters with passbands 300–1,000 Hertz were especially suitable for solving certain problems, *viz.* both the power-line noise amounting to 50 Hertz and the unsuitability of the unfiltered muscle action potentials (owing to their erratic nature). This is especially interesting considering some of the statements made in literature on the subject. *Battey et al.* (1955), state that the most favourable range is 100 to 1,000 c/s but do not justify their statement. *Horn* (1963) also uses the range 100 c/s to 1,000 c/s. This author considers it justified to neglect the information from the range below 100 Hertz owing to power-line noise. As may be seen from the above, we have, however, shown that contrary to this it is rewarding to neglect information all the way up to 300 Hertz, as the information in the range below 300 Hertz contains erratic components which are too great.

Regarding the Transmission System.

The fact that our transmission system holds the prosthesis locked in a grasp until there is a change of the current direction in the motor system is a pre-requisite for good Watt-economy in the servo-system during static conditions; as a result, the advantage of saving energy that is to be found in pneumatic systems during static conditions has been matched.

Figs. 8 and 9 show the artificial hand with its electromechanical system.

Our solution with miniature motors placed in the artificial hand involves rather great mechanical demands of a qualitative nature. The possibility of placing electrical motors of larger size outside the hand but with a purely mechanical or hydraulic transmission should not be overlooked. On the whole, it is our opinion that with respect to different types of prosthesis systems, one should not commit oneself to one type or another without gaining experience based on research or practice which will indicate the most suitable system for the work to be done at home, at work, or during spare-time occupation. It may also be necessary to try a combination of several systems.

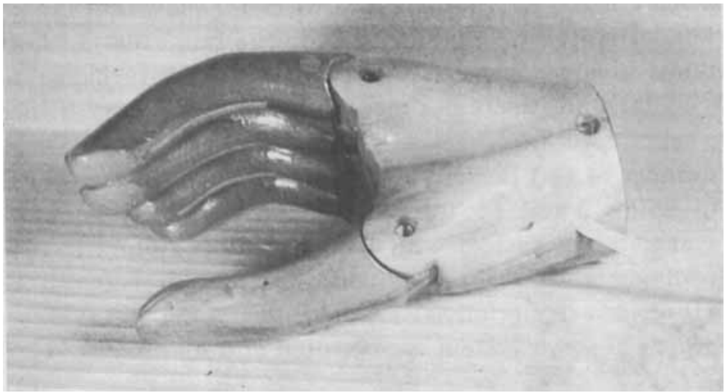

Fig. 8.
The exterior of an artifical hand containing a servomotor system.

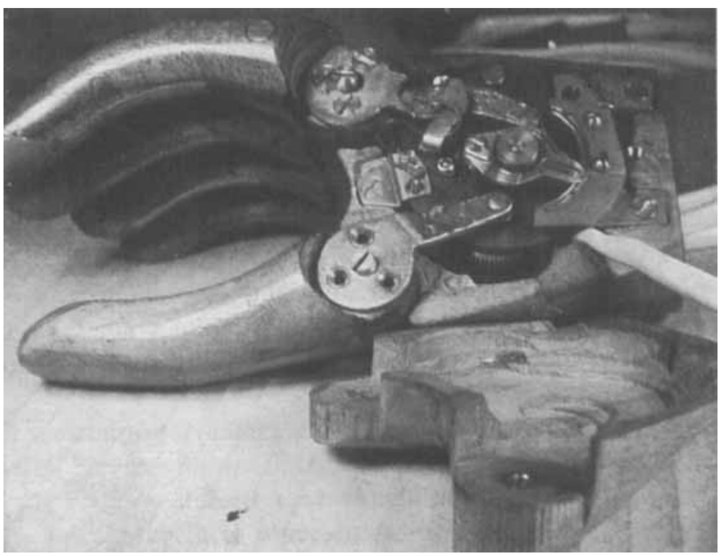

Fig. 9.
The electromechanical feedback system in the hand.

SUMMARY

1. Muscle action potentials from M. biceps brachii were derived with skin electrodes. They were analysed with octave band filters in order to determine the informational value in the different frequency ranges between 30 and 1,000 Hertz. Although the greatest activity is to be found

at frequencies of 200 Hertz and below, we could show that the informational value of muscle action potentials in the range 300–1,000 Hertz is considerably greater, because the interference disturbance between different motor units decreases more when the frequency is increasing than the signal level itself.

2. A number of test persons were investigated with respect to their ability to follow a test programme by means of muscle contraction.

3. A transmission system with two miniature motors is presented. The advantages with regard to work distribution between the two motors (force-velocity relation) are discussed.

4. A proposed servo-system is presented and discussed.

RESUME

1. Des potentiels d'action musculaire à partir des M. Biceps Brachii furent dérivés avec des électrodes en peau. Ils furent analysés avec des filtres ayant une largeur de bande d'une octave, afin de déterminer la valeur d'information dans les différentes gammes de fréquence entre 30 et 100 Hz. Bien que la plus grande activité se retrouve à des fréquences de 200 Hz et moins, nous avons pu montrer que la valeur d'information des potentiels d'action musculaire dans la gamme de 300–1000 Hz est sensiblement plus grande en raison du fait que la perturbation d'interférence entre différentes unités motrices diminue beaucoup plus quand la fréquence croît que le niveau même des signaux.

2. Un certain nombre de personnes furent testées pour déceler leur aptitude à suivre un programme de test au moyen de la centraction musculaire.

3. Un système de transmission à deux moteurs miniatures fut présenté. Les avantages quant à la distribution du travail entre deux moteurs (rapport puissance/vélocité) sont discutés.

4. Un servo-système proposé est présenté et discuté.

ZUSAMMENFASSUNG

1. Muskelaktionspotentiale vom Musculus biceps brachii wurden mit Hautelektroden abgeleitet. Sie wurden mit Oktavbandfiltern zur Erforschung des Informationswerts in den verschiedenen Frequenzbereichen zwischen 30 und 1000 Hz analysiert. Obwohl die grösste Aktivität bei Frequenzen um 200 Hz und niedriger liegt, konnte nachgewiesen werden, dass der Informationswert der Muskelaktionspoten-

tiale im Bereich 300–1000 Hz erheblich grösser ist, weil Interferenzstörungen zwischen verschiedenen Motoreneinheiten bei steigender Frequenz stärker abnehmen als das eigentliche Signalniveau.

2. Eine Anzahl von Versuchspersonen wurde im Hinblick auf ihre Fähigkeit untersucht, einem Testprogramm durch Muskelkontraktionen zu folgen.

3. Ein Impulsübertragungs- und Kraftüberführungssystem mit zwei Motoren wurde beschrieben und die Vorteile bei Berücksichtigung der Arbeitsverteilung zwischen den beiden Motoren wurden diskutiert.

4. Eine Lösung zum Servosystem wurde vorgeschlagen und diskutiert.

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