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CLINICAL APPLICATIONS OF A MULTIFUNCTIONAL HAND PROSTHESIS

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ABSTRACT

The lack of control sites and necessity of extensive patient training entail difficult problems in the control of multifunctional hand prostheses. A promising approach to the solution of these problems is the use of pattern recognition techniques in conjunction with digital computer facilities. Further developments of this previously described method have now led to the realization of a portable electronic control system. An individually adapted portable pattern recognition network as well as a portable power control unit have been constructed.

Since no commercially available electrode complies with the requirements of our signal processing, an active surface electrode has been developed. The electrode provides amplification, filtering, and rectification. The gain from rms to dc is variable, in the range 60 to 90 dB, and input impedance and CMRR are both very high. The complete prosthesis system has been fitted to a patient. Careful manufacture of the socket has been necessary to obtain optimal signal patterns. The results of the laboratory evaluation show that the patient can successfully operate the prosthesis without introductory training.

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INTRODUCTION

Externally powered prostheses for the upper extremities have been commercially available for a good number of years. The vast majority of these devices provide a single function only - mainly grasp or release - in spite of the amputees' evident demand for additional functions. Great efforts have been made in many countries to solve the control problems involved with multifunctional prostheses [1, 2, 3]. The main obstacle is the difficulty of furnishing a control system which does not require excessive effort and tedious training to operate the assistive device. This problem is becoming increasingly important as engineering advances make the hardware production of multifunctional prostheses increasingly feasible.

A method holding considerable promise of a material reduction in the burden of training imposed on the amputee fitted with an advanced prosthesis has recently been described [4, 5]. In these studies the control problem is solved by applying pattern recognition techniques to myoelectric signals in conjunction with digital computer facilities.

In the present project, a portable electronic system for operating six movements of a hand prosthesis has been developed. The system includes a set of six active surface electrodes for picking up myoelectric signals, a control unit based on pattern recognition technique, a battery unit, and power circuits for motor control. The present hardware is adapted to the Swedish hand prosthesis [6], which has six active movements, as follows: grasp, release, pronation, supination, wrist flexion, and wrist extension. It is our belief that the functional capability of this hand is sufficient to make it beneficial, provided it can be properly controlled.

METHOD

The control method is based on the perception of a phantom-limb, prevailing in every amputee [7]. When below elbow-amputees are asked to perform specific movements of their phantom hand, specific muscle contractions are produced within the stump. If a number of surface EMG electrodes are applied around the stump, different relative intensities - different patterns - of rectified myoelectric signals can be picked up for each individual movement, as illustrated in Fig. 1. The information carried by such patterns can be interpreted by employing pattern recognition techniques.

Our pattern recognition technique is based on the use of discriminant functions evaluated by a digital computer. The zeros of these discriminant functions describe decision surfaces separating the patterns; mathematically the functions are

$$f(E) = \sum_{i=1}^{n} w_i e_i + w_o = 0,$$

where $E = e_1, e_2, \ldots, e_n$,
e_i = rectified EMG signals from channel i,
w_i = weighting factor associated with channel i,
w_o = constant term,
n = number of input channels (in our case $n = 6$).

If the myoelectric signals obtained in a particular stump muscle contraction yield a value of $f(E) > 0$, that contraction is considered to belong to a corresponding particular class of movement, if $f(E) < 0$, the contraction does not belong to this class. Thus, each movement of the prosthesis has its own discriminant function. These functions decide whether the myoelectric signals originate from the corresponding movement of the phantom

Engineering particulars of the control system

The pattern recognition is performed in two phases. The first phase is the one of training in which the discriminant functions are computed and the weighting factors are determined. The second phase is the one of testing, in which the computer classifies new, unknown myoelectric patterns. In this phase, by performing natural phantom limb movements, the amputee can operate the prosthesis through the computer. The next step is to replace the computer in the classification tasks with a portable electronic network. Fig. 2 illustrates the classification procedure. This operation can be realized in an electronic network, as illustrated in Fig. 3. In stage one, the weighted sum is computed. Six of the input resistors are proportional to the inverse of the weighting factors, and the seventh determines the constant term. High resolution is obtained by using precision resistors (one per cent tolerance). The offset trim potentiometer makes it possible to adjust the constant term (such an adjustment is equivalent to a parallel displacement of the decision surface). Stage two is the threshold unit, realized by means of a comparator. The circuit shown in Fig. 3 controls a single particular movement; the entire system requires six separate such circuits. Circuits for opposing movements, such as grasp and release, say, are connected in such a way that if a pattern is classified as belonging to both movements, the output is zero. Different patients have different decision surfaces. It is thus necessary to ask the computer for individual weighting factors, and to adapt the input resistors to each individual patient.

The pattern recognition circuit is connected to a power control circuit driving the prosthesis motors. The control is of the on-off type, and simple switching transistors are used. Locking of the grasp-release function is accomplished by motor EMF feedback. The complete electronic network, the battery unit, and the portable box are shown in Fig. 4.

Electrode for signal processing

The pattern recognition network operates on essentially DC analog input signals proportional to the contraction level of the muscles. This implies that the myoelectric signals picked up on the surface of the skin have to be processed in several stages before reaching the control unit. First, they must be amplified and band-pass filtered, then rectified, and after that low-pass filtered. The maximum rms voltage obtained between two skin electrode plates of 20 mm separation is between 50 and 1000 μV. To obtain an output DC signal with a dynamic range of a few volts, the required gain, from rms to DC, must be of the order of 60 to 90 dB. Suppression of signals appearing at both electrodes under identical phase and amplitude conditions - common mode signals - must be efficient in order to reduce disturbances. This suppression is obtained by a high common mode rejection ratio (CMRR) and a high input impedance in the first differential stage; preferably the CMRR should be more than 70 dB, and the input impedance should be more than 100 MΩ. The interface impedance between tissue and electrodes may fluctuate quite considerably (as much as ±50 per cent). If the input impedance of the amplifier is not high compared to the interface impedance, the fluctuations will cause troublesome errors when common mode signals occur [8, 9].

No commercially available electrode satisfies the requirements of our signal processing - neither the on-off types, which must clearly be rejected, nor the linearly amplifying types, which have too low a CMRR or too low an input impedance. The electrode we use is one of our own design, featuring a proportional output, a CMRR of 80 dB, an input impedance of 10^{12} ohms, and a gain variable in the range 60 to 90 dB. The circuit diagram of the electronics in the electrode is illustrated in Fig. 5. The input consists of two voltage followers followed by a differential stage and a subsequent coupling capacitor. To avoid problems arising from polarization voltages and the late AC-coupling, the voltage gain of the differential stage has to be low (a gain of ten, say), and the input bias current drawn from the tissue must be very low. The voltage followers have a bias current less than 1 pA and an extremely high input impedance. The high CMRR is accomplished utilizing a potentiometer in the differential stage. In the following two stages the myoelectric signal is amplified and band-pass filtered. The filter has a lower cut-off frequency of 120 Hz, with a low frequency slope of 12 dB/octave and an upper cut-off frequency of 1200 Hz, with a high frequency slope of 6 dB/octave. Adequate speed and negligible ripple is obtained by the full-wave rectifier and subsequent low-pass filter with a time constant of 0.1 s. Negative weighting factors of the pattern recognition equation are obtained from an additional output stage of the electrode, yielding the inverse of the output signal. Since each electrode serves a six-channel load, its output may, in the extreme case, face an impedance of less than 1 kΩ. Accordingly, the outputs are provided with driver stages precluding attenuation of the output signal arising from heavy load.

The complete electrode circuit is molded in araldite, which makes the electrode waterproof and resistive to mechanical impacts. The skin-connecting surfaces are made of gold-covered brass buttons.

Socket design

The socket is of the conventional Münster design. It has to be carefully manufactured to ensure amputee comfort as well as good myoelectric pattern acquisition. It is important that the socket fits perfectly, so that all electrodes remain in contact with the skin whatever the position of the arm. This perfect fit is facilitated by mounting the electrodes on elastic springs, providing a resilient contact. Since signal patterns depend on electrode positions [4, 5], the electrodes must be accurately and stably located within the socket in order to yield optimal separation of the different patterns. A typical socket and electrode mounting is illustrated in Fig. 6.

CLINICAL APPLICATIONS

The control system and a socket have been adapted to one patient, a 35 year old male who lost his left forearm in an accident 25 years ago. The length of his stump as measured on the dorsal side of the forearm from the tip of the olecranon is 17 cm. The patient has a remaining phantom limb perception of his lost hand. He distinctly perceives the possibility of opening and closing the phantom hand and of extending and flexing the wrist. A simple belt enables him to wear the box of batteries and electronic gear under his jacket. The complete system, worn by the patient, is illustrated in Fig. 7.

The gain of the electrodes was adjusted so that the output signal amplitudes normally reached a level of a few volts. At very heavy contractions the signals reached maximum output voltage. Signal patterns of the different phantom hand movements were recorded while the patient was sitting and standing with the arm in various positions. These patterns were put together and analyzed, and the weighting factors were tested in the second computer phase. After this phase was successfully carried out, the weighting factors were converted into resistor values and the complete system was evaluated.

EVALUATION AND RESULTS

At present, the system has been subject to laboratory evaluation only, and only one single patient has been tested. The patient, who had no introductory training, was asked to perform each separate prosthesis movement, i.e. grasp (G), release (R), pronation (P), supination (S), wrist flexion (WF), and wrist extension (WE). The output signals of the pattern recognition network - in other words the binary control signals of prosthesis motors - were recorded by the computer as the patient responded to the required performance. Whenever a movement command was given, computer recording was initiated and was kept running until the movement was completed. The span of time needed by the patient to select the correct pattern of each movement - the selection time - was defined as the time elapsed from the moment of command until the correct output signal was present. Since all six output signals were recorded, the computer also provides information on non-desired simultaneous movements.

In order to evaluate patient control of the prosthesis in various positions the investigation was carried out under different conditions. The following common arm positions were chosen: (1a) sitting with the arm supported at the elbow, with an elbow flexion of 90°, and with the hand open; (1b) same position as in (1a), but with the hand closed; (2) standing with the upper arm along the trunk, with an elbow flexion of 90°; (3) standing with a shoulder flexion of 30°, and with an elbow flexion of 120°; (4) standing with a shoulder flexion of 90° and an elbow flexion of 120°; (5) same position as in (2) but with a prosthesis hand load of 1 kg. In no case was any abduction or rotation of the shoulder involved.

The selection times are presented in Fig. 8. All movements could be performed at once. In previous investigations [4, 5], in which the computer was used for the control, selection times of less than one second were considered excellent. As indicated in the figure, selection times attained with the portable control unit are well below one second for all movements, in all various positions tested, including those in which the prosthesis was loaded.

In Fig. 9 the signals for simultaneous movements are illustrated; in (a) the movements should be performed with an open hand and in (b) with a closed hand. It is obvious that the patient can perform the intended movements without any interfering movements. The output signals appearing at Release in Fig. 9(a) during pronation and supination are quite expected as well as non-detrimental since the instructions were to keep the prosthesis hand open. In Fig. 9(b) the hand should be closed and it is therefore not surprising that output Grasp signals occur in some movements. In Fig. 9(b), however, when grasping, wrist flexion occurs for 4.5 per cent of the time. When supinating, wrist extension occurs for 5 per cent of the time. In extending the wrist, release, pronation, and supination occur for 2 per cent, 4 per cent and 2 per cent of the time, respectively. Such short signals will not, however, produce visible prosthesis movements. Some of them might occur as a result of mechanical vibrations of the prosthesis, induced when the intended movement is stopped by the corresponding end position switch.

DISCUSSION

The new electrodes as well as the portable pattern recognition system have completely held up to evaluation requirements. Great efforts have been devoted to developing a reliable electrode, insensitive to disturbances. The high input impedance and the high CMRR of the amplifier have accomplished negligible errors arising from common mode signals. As the electrode is molded in araldite and highly reliable electronic components are used, high mechanical and electrical withstanding has been achieved. Despite the considerable number of components needed for satisfactory signal processing, the physical size is adequate. The total cost of the components is approximately $30.

Evaluation has demonstrated that the present configuration of the pattern recognition network is feasible. Using modern miniaturized components, it is possible to further reduce the size and weight of this unit. The final individual adaption of the circuit has been accomplished by means of the offset trim potentiometers. This is a feasible way of making adjustments. It gives an opportunity for immediate adjustments when the complete system is worn by the patient. Furthermore, it may also be necessary to compensate for successive changes in the signal patterns arising from training effects [10].

The pattern recognition can also be implemented by a micro-computer. In this interesting possibility the decisions are performed by a central digital processing unit, and the weighting factors are stored in programmable read-only memories that can be rapidly reprogrammed (with the patient in the waiting room). The costs of micro-computer components are fairly low, and new improved devices are continuously developed.

The manufacture of the socket requires great care; it must be of perfect fit and have exact electrode positioning. These requirements are possible to meet, but before a socket is perfect many may have to be rejected (a previously [11] reported experience). There can be no doubt, however, that any skilled orthopaedic workshop experienced in applying commercially available prostheses should be able to make the sockets.

Previous results [5] have indicated that the pattern recognition method is feasible. The results of the current development show that the complete portable control system is a reality. Without any kind of training, the patient is able to operate the prosthesis successfully. It is important to note that performance is as good with the patient standing with his arm in various positions as it is in the sitting position. Intended simultaneous movements are possible by this method, something that never occurs with individual myoelectric control sites despite intense training [12]. Loading of the prosthesis does not affect the patient's control capability, reflecting the fact that the influence of moderate muscle load on myoelectric patterns is negligible.

An advanced technical organization is needed for the individual adaption of this system. A digital computer has to be employed for the on-line analysis of the myoelectric signal patterns, simulation control, and individual adjustment of the electronic circuits. This requires a center of good medical and engineering resources.

The project is now continuing with an evaluation of the system from a functional point of view. It is, however, necessary that further development and production of multifunctional prostheses take place before more clinical experience can be achieved.

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CAPTIONS

Fig. 1. Amplified, rectified and smoothed myoelectric signal patterns from six electrode pairs on the forearm stump. The patient is performing six consecutive movements with his phantom hand.

Fig. 2. Classification of myoelectric patterns. Each of the six prosthesis movements has its own unit, deciding if the pattern originates from the corresponding phantom hand movement. Symbols W_{ij} denote weighting factors, W_{io} constant terms.

Fig. 3. Electronic network for classifying myoelectric patterns. Unity ("1") at the output implies a signal to the corresponding prosthesis motor, zero ("0") implies no signal.

Fig. 4. The electronic circuits, the batteries, and the box in which the system is carried. The individually adapted resistors are mounted on the circuit board shown to the left.

Fig. 5. Electronic network for amplifying, filtering, rectifying, and smoothing the myoelectric signals.

Fig. 6. The socket with electrodes mounted.

Fig. 7. The complete system with the prosthesis applied to the patient.

Fig. 8. Selection times in various positions; see text.

Fig. 9. Diagram showing output signals obtained for various intended movements. Vertical axes show output signals, horizontal axes show intervals of each intended movement. Interval length corresponds to the time needed to complete the movement.

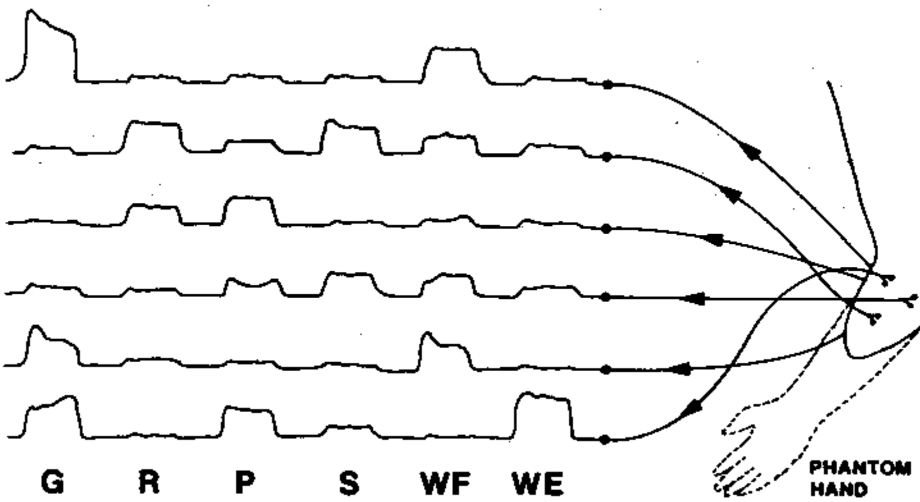
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Fig. 1.

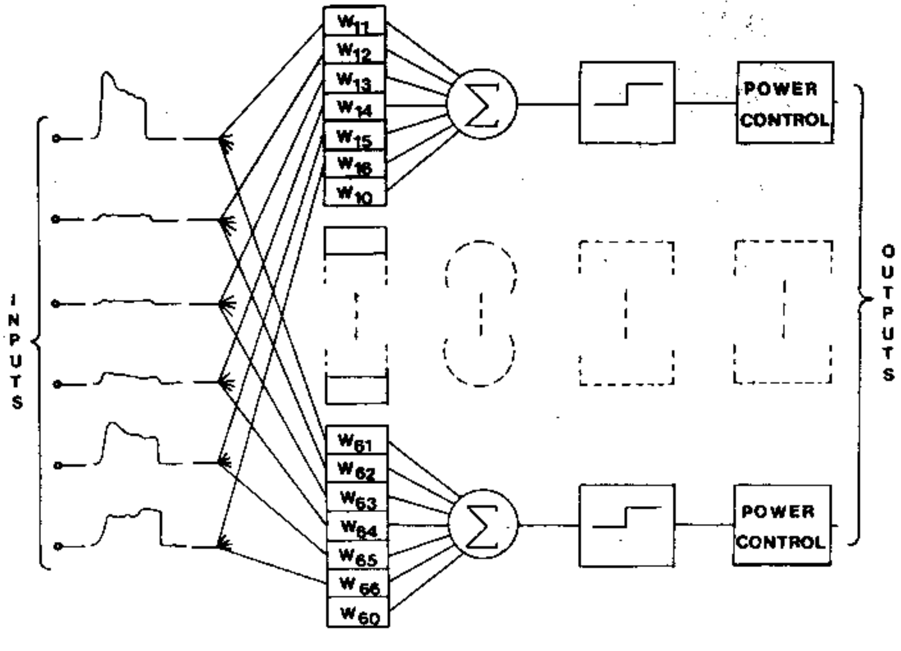
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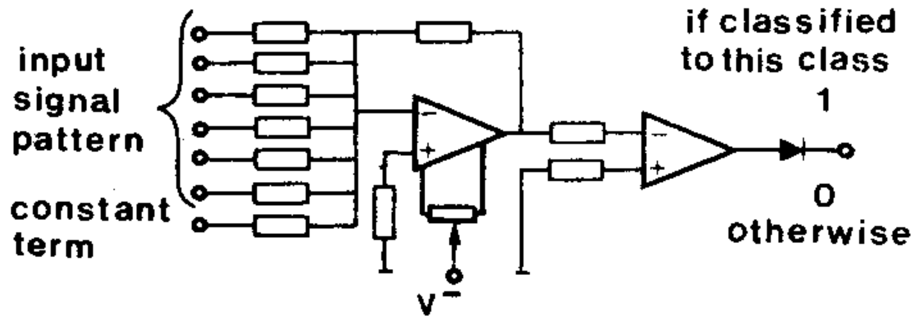
Fig. 2.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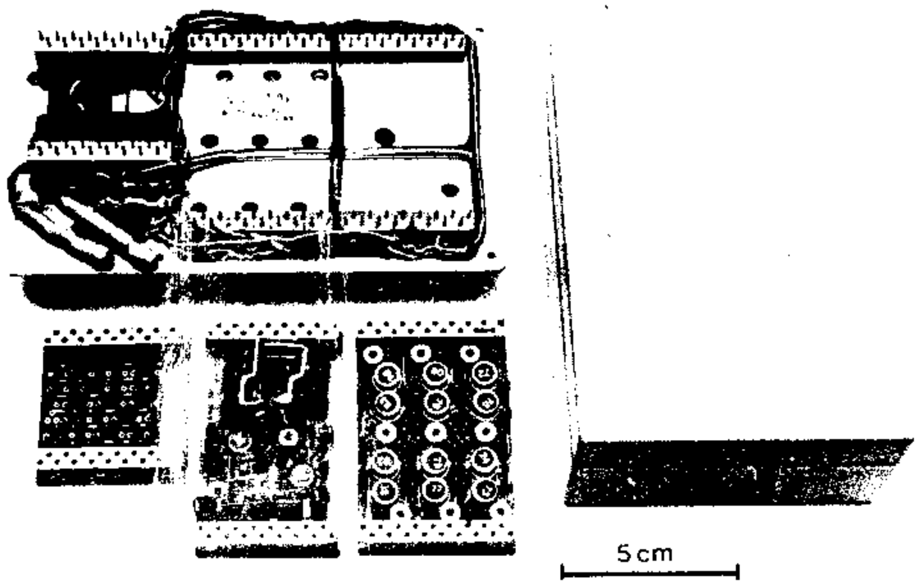
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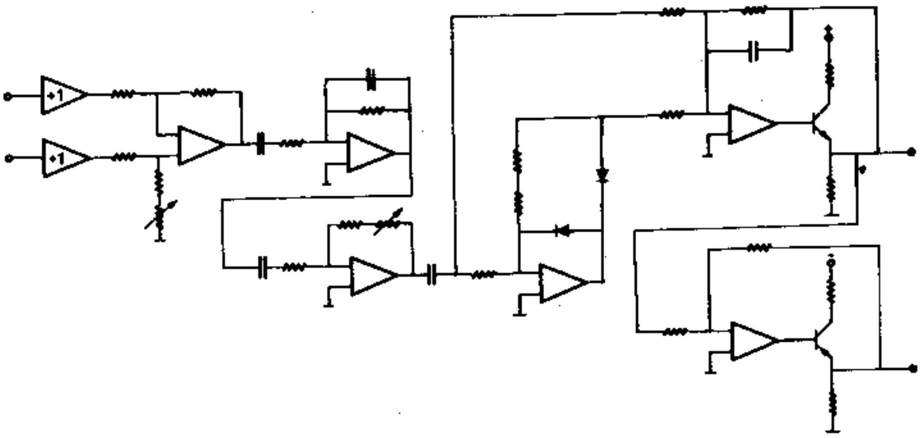
Fig. 4.



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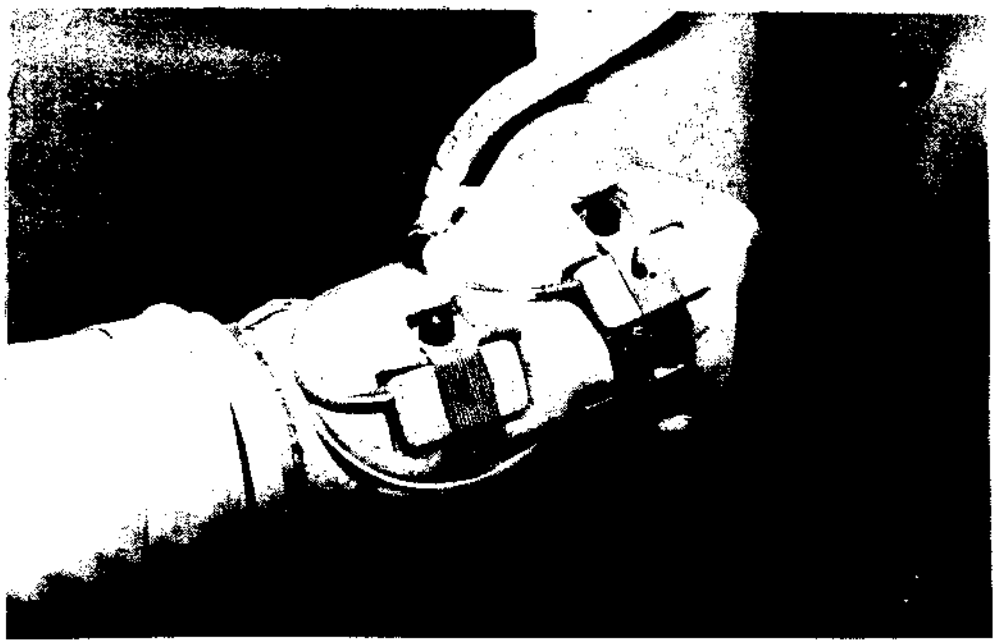

Fig. 6.

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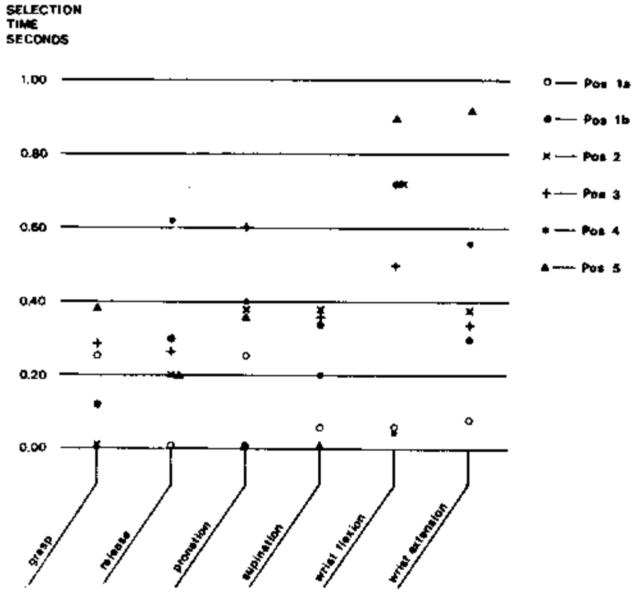
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Fig. 8.

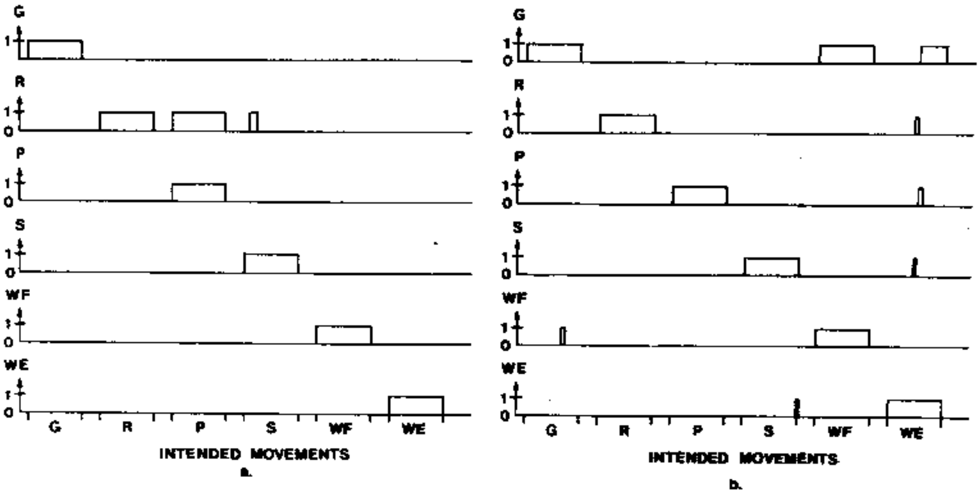
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Fig. 9.

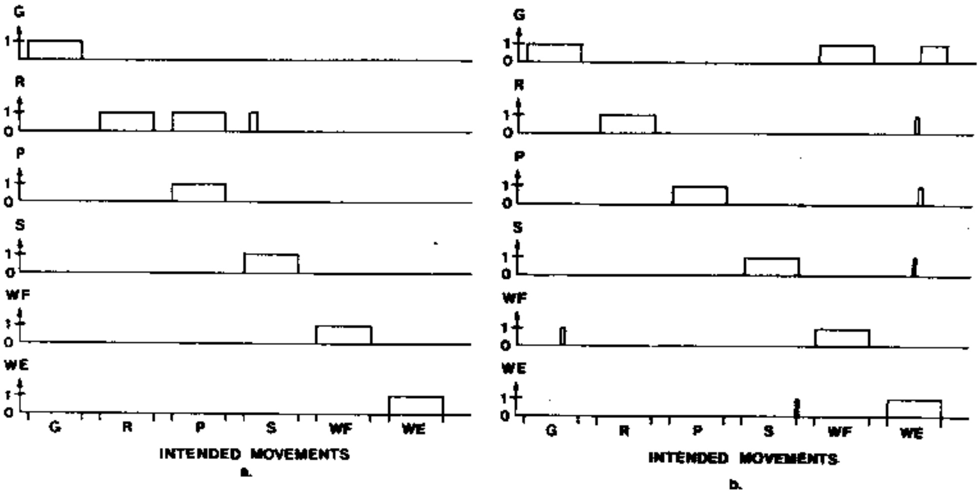

Fig. 9.