DEVELOPMENT OF MULTI-FUNCTION HAND PROSTHESIS

A DISSERTATION
submitted in partial fulfilment
of
the requirements for the award of the Degree
of
MASTER OF ENGINEERING
in
ELECTRICAL ENGINEERING
(Measurement & Instrumentation)

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ROORKEE
1977

CERTIFICATE

Certified that the dissertation entitled, [illegible] OF [illegible] - [illegible] PROCESSOR which is being submitted by Sri C.S. [illegible], in partial fulfilment for the award of the degree of Master of Electrical Engineering (Measurement and Instrumentation) of the University of Roorkee, Roorkee is a record of bonafide work carried out by him under the supervision of guidance undersigned. The matter embodied in this dissertation has not been submitted for the award of any other degree or diploma.

It is further certified that he has worked for six months from January 1977 to June 1977 for preparing this dissertation at this University.

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With regards to all,

Dated
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INTRODUCTION

Myoelectric energy is very insignificant in magnitude and this cannot be used, even after suitable amplification, to drive the cybernatic mechanism. Hence the driving energy is electric (like electric motor driven by rechargable battery) or pneumatic (like carbon-dioxide cylinder) and the myoelectric potential is used, after amplification, to control the driving energy.

In this introductory portion, we will discuss the nature of myoelectric potential.

Human skeletal muscle is composed of many thread-like fibres. These fibres do not exert a constant contractile force, but rather contract and relax repeatedly at rates as high as 50 times/sec. The fibres rarely, if even, act individually. Rather, they are innervated in groups. Each group, which contains from 2 to 2000 fibres, depending upon the muscle function, is innervated by a single nervė axon. The group of fibres, together with the axon and the nerve cell body, is referred to as a motor unit, and is considered the basic functional unit of a muscle.

The muscle fibre may be regarded, for the purpose of this discussion, as consisting of a cylindrical membrane with fluid both inside and outside. In its resting state, the membrane is thought to be selectively permisble to ions, with the result that an ion concentration difference and an

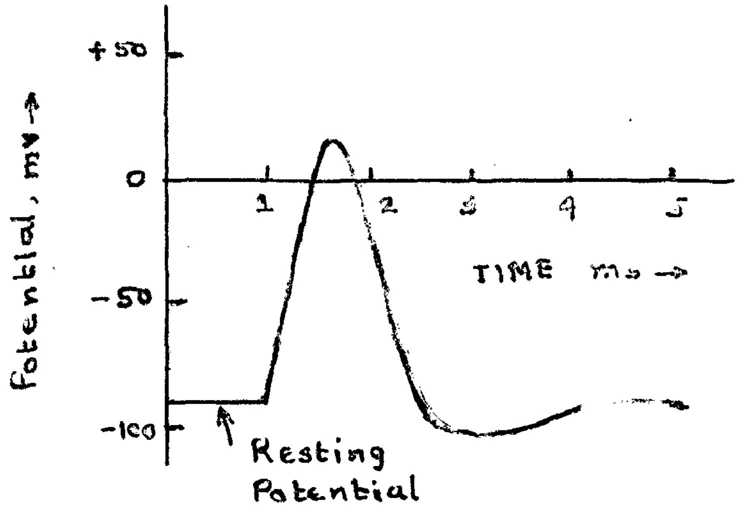


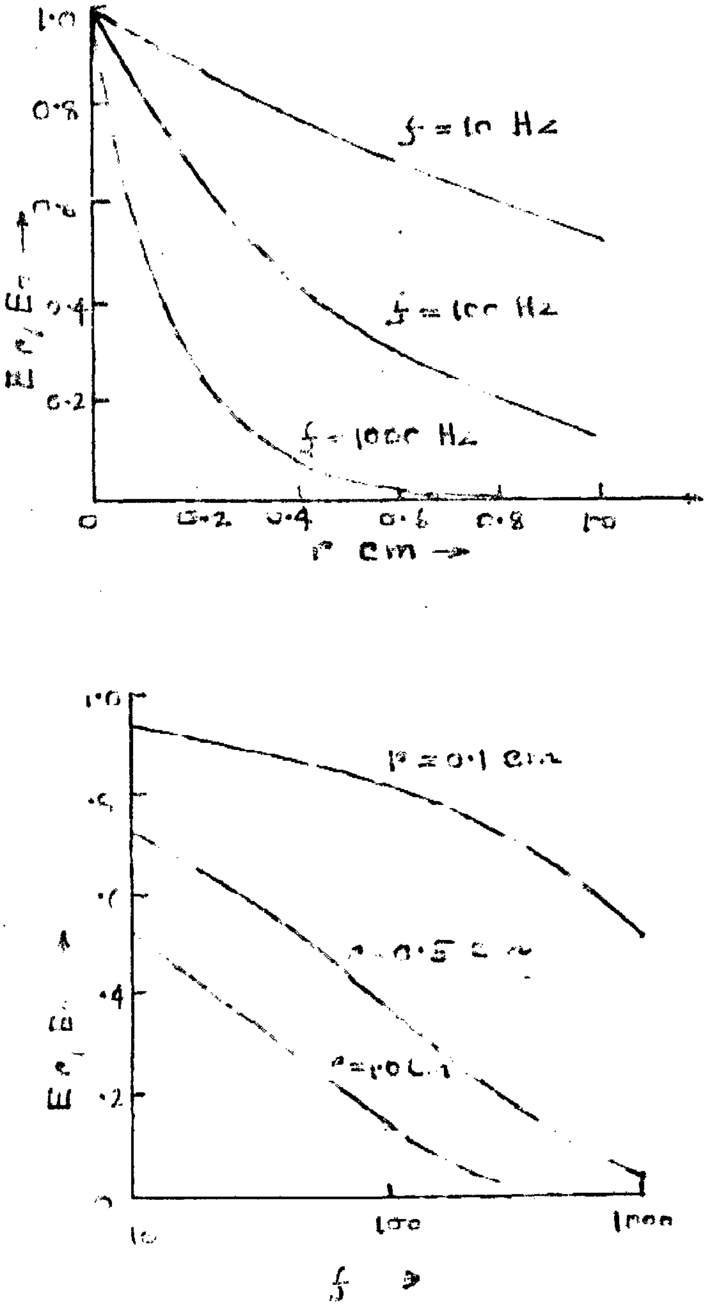
FIG-1.

accompanying electric potential difference exist across the membrane, i.e. between the inside and outside of the fibre. The potential difference is relatively large, nearly 0.1 Volt.

When the fibre is to contract, a rapid increase in the membrane permeability is thought to occur at the point where the motor nerve terminates on the muscle fibre. This increase in permeability permits an exchange of ions between the intracellular and extracellular fluids and results in temporary elimination or reversal of the resting potential difference. Moreover the disturbance of permeability and of ion concentration propagates rapidly in both directions longitudinally throughout the fibre. The contraction is followed promptly by a return to the normal resting state, with the whole depolarization - repolarization cycle lasting only a few milliseconds.

If an electrode could be placed within the intra-cellular fluid (without damaging the fibre) and if another electrode were placed in the extracellular fluid, a potential variation similar to that shown in fig-1 would be observed between the two electrodes each time the fibre contracts.

When a muscle fibre contracts, a flow of ions occurs in the extra-cellular fluids immediately surrounding it, as mentioned previously. Consequently, an electric potential variation may also be observed between a pair of electrodes located nearby in the extra-cellular fluid.



1·0
0·8
0·6
0·4
0·2
0
0·2
0·4
0·6
0·8
1·0
f = 10 Hz
f = 100 Hz
f = 1000 Hz
r cm →
·4
·2
10
100
1000
f →
r = 0·1 cm

FIG - 2.

Because of the high conductivity of that fluid, this potential variation is attenuated rapidly with increasing distance from the fibre. Recent experiments directed by R.N. SCOTT verify that the high frequency components are attenuated more rapidly than the low frequency components, the relation being roughly [1]

$$Er = Eo \exp(-0.2\, r\, f^{1/2})$$

where Er is the potential difference observed at a distance r cm radially from the fibre, Eo is the potential difference at the fibre, r is the distance in centimeters, and f is the signal frequency of interest in Hertz. The effect of this attenuation is illustrated in fig-2.

If a pair of tiny electrodes is inserted among the muscle fibres, contraction of a fibre very close to the electrodes will produce a ' sharp ' potential change, of relatively high amplitude. A fibre located further from the electrodes will produce a smaller potential change, having relatively less high frequency content. While it is unlikely that a single muscle fibre will contract alone, it is not uncommon for a single motor unit to contract in isolation. The potentials from several fibres of a motor unit will be observed by a given electrode from as many different directions and distances. Further, the contractions of the various fibres of a motor unit are not exactly synchronous. Consequently, the potentials observed from contraction of single motor units are often quite complex

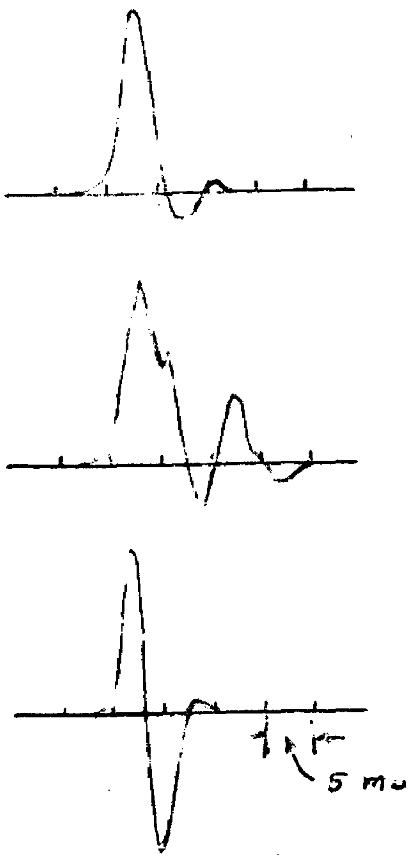


FIG. 3.

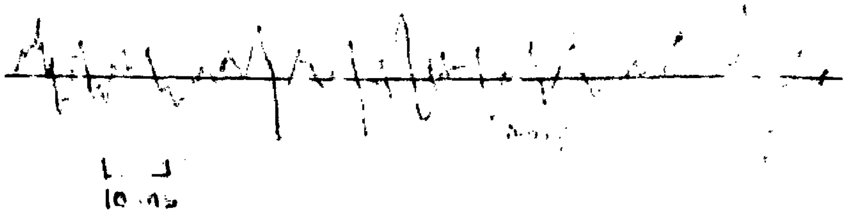


FIG-4.

and are usually of longer duration than single fibre potentials. Typical single motor unit potentials are shown in fig-3.

When a muscle contracts normally, various motor units act asynchronously, each unit contributing from 5 to 50 brief twitches per second. The resulting disturbances of electric potentials in and near the muscles are very complex. When plotted as a function of time, they constitute the familiar interference pattern of electro-mygraphy (fig-4).

One property of this signal which is of importance in the design of myo-electric control systems is the relative energy content at various frequencies. It is generally agreed that there is little useful energy below 20 Hz or above 1000 Hz. About the distribution of energy in between these two limits there is significant difference in opinion. Hirsch et al. maintains the view that energy distribution is more above 100 Hz, whereas R.N. Scott finds it below 100 Hz.

In our present work ~~we have provided~~ a band-pass filter with lower cut-off 100 Hz and higher cut-off about 300 Hz has been provided. For the control purpose it has been found satisfactory. It is to be noted that EMG potentials as that of fig-4 is of not much use. After rectification and low-pass filtering it takes a distinct shape as will be explained afterwards.

The last questions remains, whether an individual can voluntarily control the electric activity in a muscle.

It was at first viewed with scepticism, but it is now generally agreed that conscious voluntary control of the myoelectric activity in a muscle is possible. It is even possible to excite a single motor unit though it requires considerable training. Since the fibres of a single motor unit are distributed within a radius of 5 mm, a single muscle can provide many control sites.

PREVIOUS DEVELOPMENT

Before the myoelectric control, the artificial arms were body-operated, either by harness or by cineplasty. The external power source like carbon-dioxide cylinder came latter in use.

The force of harness control is obtained from relative motion between two or more segments of the body.[2] Force and excursion usually are harnessed by means of 1-in-wide webbing and a Bowden cable, a flexible cable running through a wrapped wire housing. Three mostly used sources are : Biscapular abduction (using scapular abductor muscles), Arm flexion (using humeral flexor and scapular abductor) and Arm extension (using humeral extensors).

In cineplasty a tunnel is made through the belly of a muscle (like biceps or triceps) through which one end of flexible cable passes. The movement of the muscle provides required force and excursion to the cable which ultimately controls the terminal device.

External power pack like carbon-dioxide cylinder can provide more than one operation. Switches are arranged on the shoulder and by operating different switches, the high pressure carbon-dioxide is directed in various channels to obtain different types of movements. Electro-mechanical control in the similar way is possible by operating different micro-switches.[3]

Ofcourse all these methods lack that 'novel advantage' - the ' control by will ', which is possible in the case of ' phantom sensation ' only. If a relatively unused muscle can be trained to excite at 'will' then also that novel advantage is retained. These muscles may be very small in size and hence cannot provide harness control of cineplasty. The pick up of myopotential becomes a necessity.

Monofunctional Prosthetic Hand :

To obtain a single-functional prosthetic hand by myoelectric control is relatively easy. Multifunction cannot be obtained simply by increasing number of switches as in the case of carbon-dioxide control or electro-mechanical control. It involves the use of ' pattern recognition ' as will be explained afterwards.

The first suggestion that myoelectric control was feasible is generally attributed to Professor Norbert Weiner, in the late 1940's. The first published account of research on this topic seems to be that of Berger and Huppert in 1952, describing a brief feasibility study carried out at New York University. That account was quite optimistic. However, Alderson, in discussing the results, concluded that while advancing technology might alter the situation, it was easier in 1954 to use the mechanical output of muscles for control. The next development in the United States was the report, in 1957, of the use of myoelectric potentials for automatic control of a respirator. This work was carried out at the school of Medicine, Vanderbilt University and was followed in 1959 by demonstration of voluntary myoelectric control of a carbon-dioxide - powered artificial muscle at Baylor University. Meanwhile the first report of a working myoelectric control system had been published by a research group at Guy's Hospital, London, England in 1955. This laboratory demonstration of the feasibility of myoelectric control was followed by the development in London of increasingly complex systems.

In the U.S.S.R., myoelectric controls research began at the academy of sciences, Moscow, in 1957 and a working model fitted to a patient was demonstrated in 1960. Clinical application was immediate, and by 1966 over 1000 patients had been fitted with this equipment.

Thus it will be seen that myoelectric control was first studied in the United States, first demonstrated in England and first used for voluntary control of artificial limbs in the U.S.S.R. There has been an understandably rapid growth in interest in this new control technique.

A simplified block diagram of the U.S.S.R. system is shown in fig-5. Two muscles are used in direct on-off control of the motor which operates an artificial hand. The equipment is intended for use by a below-elbow amputee, with muscles of the flexor group in the stump providing the control signal for the ' closing ' channel and extensor muscles controlling the ' opening ' channel. Metal electrodes in contact with the

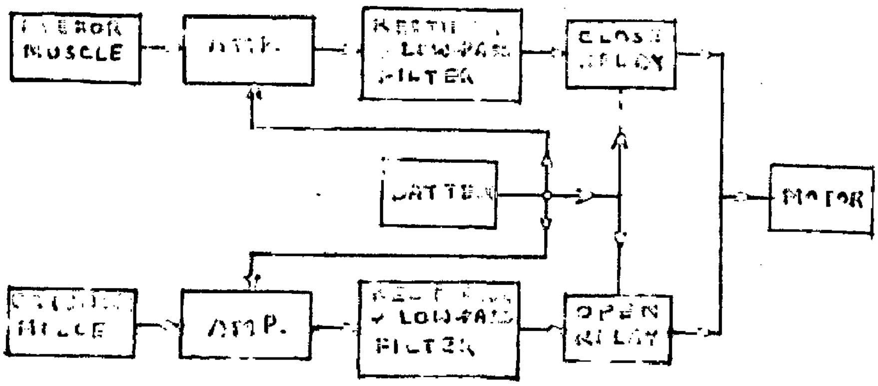


FIG-5.

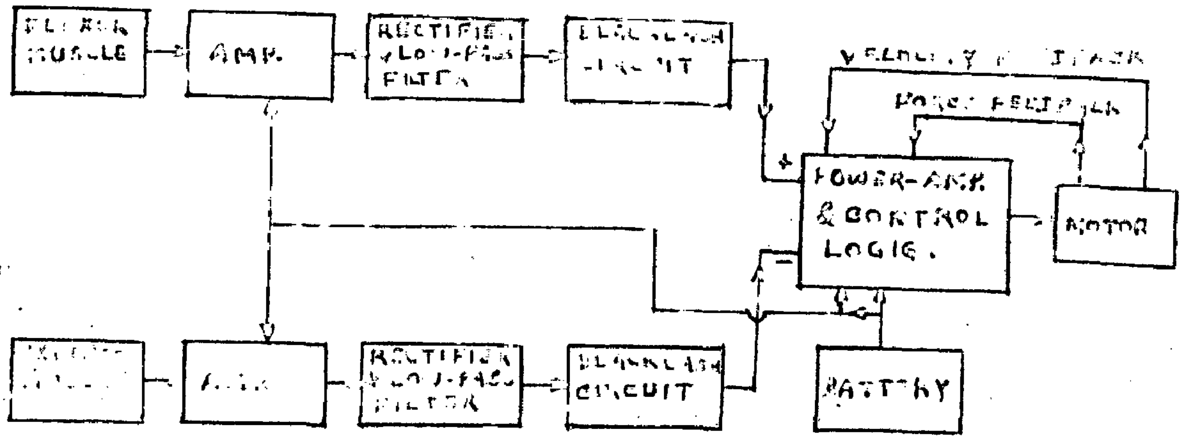


FIG-6

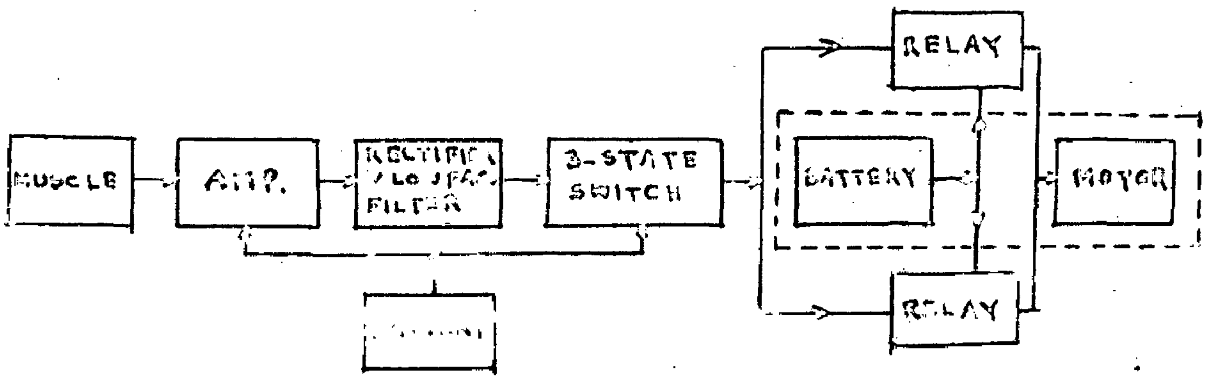


FIG-7.

skin, with electrode paste for improved electrical conduction, are employed. The electronic control unit (for two channels) measures 82 x 46 x 11 mm and weighs 120 g. The 13.5 volt rechargeable battery which provides energy for the prosthesis and the control unit weighs 320 grams. Miniature electro-magnetic relays are used here.

In England, the myoelectric control system developed by Bottomley and manufactured by Atomic Weapons Research Establishment in 1965 was more complex. It isshown in the simplified block diagram in fig-6. One basic feature is that the difference between the myoelectric outputs of the flexor and extensor muscles is used as the control signal. However, the major difference from the USSR system is the provision of continuous control of prehensile force or velocity, instead of simple on-off control of the motor.

This added function is not gained at no cost. The control unit is having the approximate dimensions 118 x 118 x 32 mm, weighing 425 g (roughly 5.6 times the volume and 3.5 times the weight of the USSR Control Unit) with a battery package weighing 680 gm (roughly 2.2 times the weight of the USSR battery). It is also inevitable that any system which provides a continuously variable motor speed or torque from a dc supply will be less efficient than a system which merely provides on-off control. Consequently, for the same operational life batteries for the former system will be considerably larger and heavier.

Two prosthetic systems which closely resemble the USSR system have become available in Europe. One is in Italy by INAIL, its specification and general description are like that of USSR system except that it is slightly large and heavier. The other is an Austrian System. The control unit and the battery pack are combined into one flat package 107 x 130 x 20 mm, weighing approximately 510 g, and switching transistors are used instead of relays to provide on-off control of the motor-driven hand. The available prehensile force, 5-6 kg, greatly exceeds the 1-5 kg reported for the USSR system. Gold-plated electrodes are used with no electrode paste.

A myoelectric control system developed at the Bio-Engineering Institute of the University of New Brunswick incorporates the level discrimination technique mentioned previously. Independent on-off control of two functions from one site is provided. A block diagram is shown in fig-7. A slight time-delay in the initiation of function-1 is provided, in order that the transition between the 'off' and function-2 states may be accomplished without activating function-1.

There has been considerable experimentation on feedback.[4] The magnitude of opening or closing of prehension is converted into proportional frequency and it is feedback on the skin of the subject by electrical or mechanical vibrator. The experiments of Osamu Sueda on mechanical vibratory feedback and of University of New Brunswick on electrical stimulation is of considerable importance.[5]

The griping force can also be fed back by suitable strain-gauge mounted in the thumb. The Otto Bock hand has been tested by semiconductor strain guges.

Multifunctional Prosthetic Hand :

To produce multifunction by myoelectric control is considerably difficult. If wrist is flexed on extended by similar magnitude as that of fingers, what change is there in EMG signals? Obviously the band of frequency as well as the amplitude (average) remain more or less same. Then what is to be taken as discriminating factor? Only the pattern of EMG is changed and thereby it requires a method of pattern recognition. Initially it seems to be the only possibility and the researchers proceeded towards that.

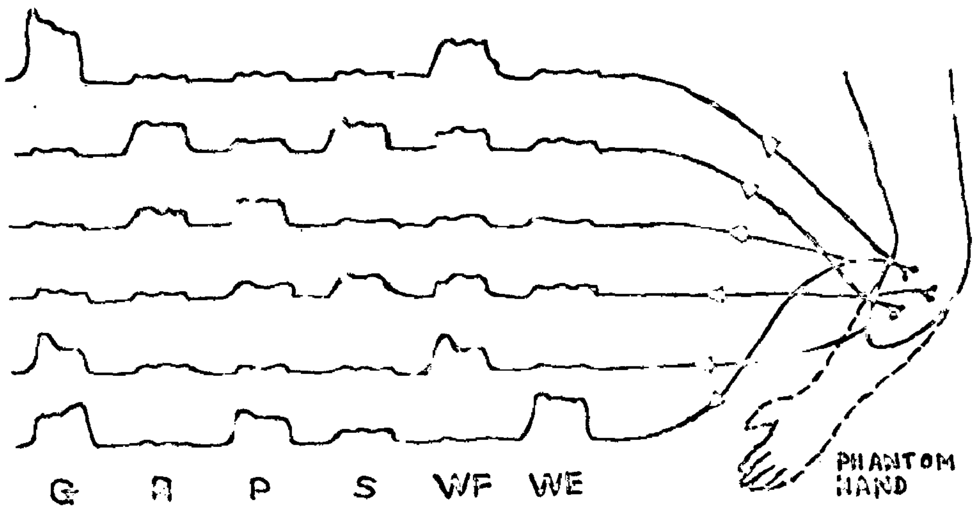
Of course the very wide range of motions of ~~figures~~ fingers, wrist and elbow cannot be produced in artificial hand. Apart from the difficulty of achieving such complex control, the electro-mechanical arrangement of such various movements on different axes are not realizable. Hence the attempt has been to produce the following six basic movements in the case of below-elbow amputation :

FF - Finger flexion

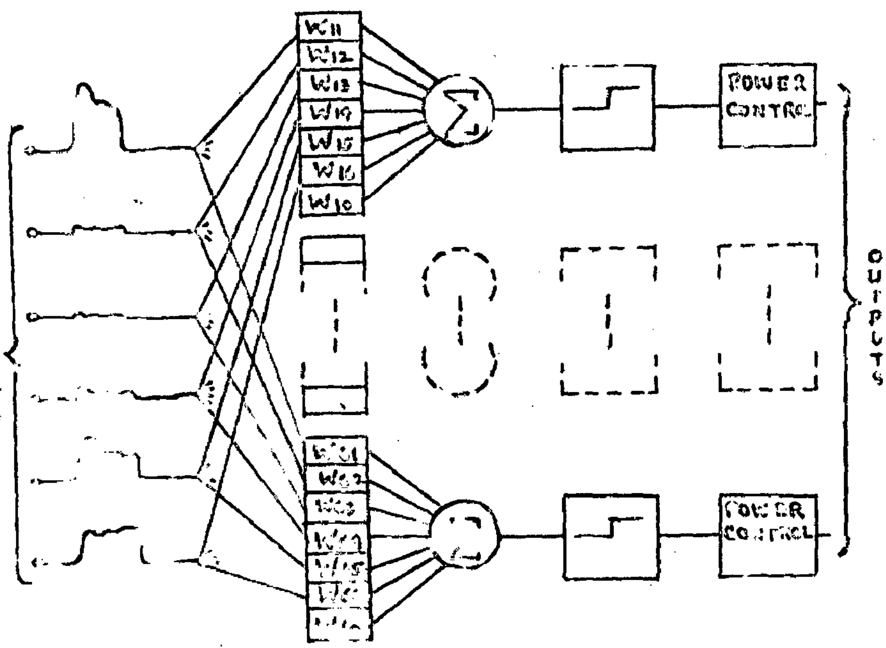
FE - Finger extension

WF - Wrist flexion

WE - Wrist extension


G
R
P
S
WF
WE
PHANTOM HAND

FIG—8.


W11
W12
W13
W14
W15
W16
POWER CONTROL
POWER CONTROL
INPUTS
OUTPUTS

FIG—9

P - Pronation and

S - Supination.

Upto the present time two different research groups have attempted multifunctional motions in two different ways. AGA Orthopaedics from Scandinavia has adopted the method of pattern recognition by digital computer and then realising it by ' weighting factors ' decided by computers.[6] The other group is from Colorado State University, they have separated the signals for different movements by statistical analysis of EMG. The typical method used by them is called ARMA (or auto-regressive moving average) calculation.[7]

Since our approach is also for multifunction, these two already-existing methods are shortly described below.

The Scandinavian team is headed by Christian Almstrom and Peter Herberts. Their control method is based on the perception of phantom-limb, prevailing in every amputee. 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-3. The information carried by such patterns can be interpreted by employing pattern recognition techniques.

Their 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 $E = e_1 , e_2 \ldots\ldots e_n$,

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 their 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 prostheis has its own discriminant function. These functions decide whether the myoelectric signals originate from the corresponding movement of the phantom-limb.

In the actual procedure, the EMG signals from all six channels are fed to a computer with the desired movement specified. The output of the computer is the optimized ' weighting factors '. These weighting factors are realizable by electronic networks. The way of connection is shown in fig-9 for two desired movements of hand. The corresponding output terminal will be at 1-state if that particular motion is desired, rest all five output terminals will be at 0-state.

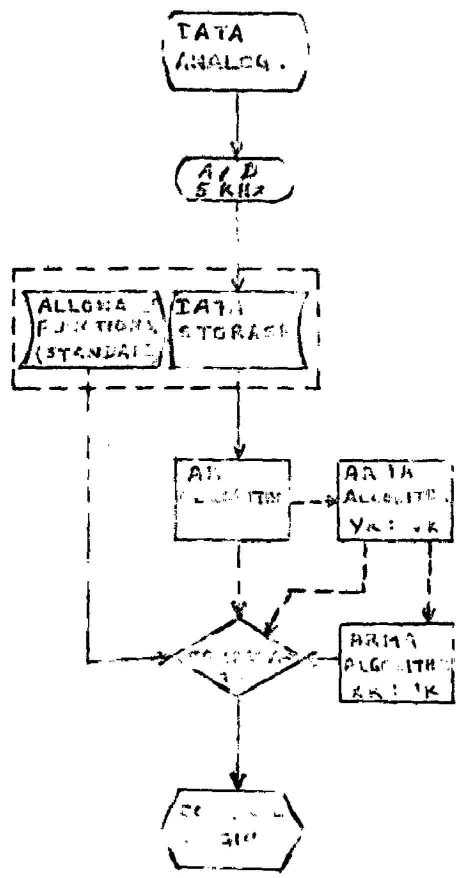


FIG-10.

The analysis of the Colorado State University group has been based on the nature of time series in that their pattern can be parametrized into a finite set of parameters of a linear signal model that form a reduced minimal set as compared with the (almost) infinitum of values of the pattern itself. These parameters must not be all stationary or unique per each function for function separability. However if at least some of these parameters are such that their range of variation for a given limb function does not overlap with that of other functions, then such separation is possible, as is shown later to be the case in all their surface-electrode EMG recordings (over 250 records).

An approach to the aforementioned recognition problem is given in terms of deriving a fast parametric-recognition algorithm whereby the autoregressive moving-average (ARMA) parameters and the Kalman filter parameters of the EMG time series are identified. It is shown that the resulting identified parameters yield sufficient information to discriminate between a small number of upper extremity functions.

The hardwave realization of this technique, as suggested by the authors, is shown in fig-10. It requires A/D converters memory, comparators etc. and is realizable by micmoprocessors.

Both these systems are susceptable to the following drawbacks :

(a) Hardware realization is a very complex process as well as costly. Either it will require a digital computer or microprocessors.

(b) Apart from the complexity, the volume of the hardware is considerable. In Scandinavian system six EMG channels are required whereas in Daiel Grape's method several channels have been suggested for use to increase the reliability of the method.

(c) The position of the electrode on the muscle is to be accurately maintained. In Scandinavian system, it has been reported that, the displacement of electrode by 2 mm from the required position has been critical. This matter is due to the fact that pattern of EMG is changed due to changed position of electrode, and since the whole of their procedure depends on pattern of EMG itself, the recognition system cannot handle a new pattern. Usually the electrodes are fitted on the inner surface of the socket and slight movement between the socket and the stump is inevitable when the prosthetic hand will handle some load.

(d) Even if electrodes are kept absolutely in position over the muscle, the EMG pattern will be changed due to fatigueness of the muscle after working ~~from~~ for sometime.

Keeping all these factors in view, we have developed a system which does not depend much on the pattern of EMG. Even if electrodes are displaced by centimeters or pattern is drastically changed by fatigueness, our system is supposed to work, because here the rectified signals will have to satisfy a minimum required amplitude only. For the separation of functions it requires only a few logic gates. The

OUR APPROACH

The method of picking up myoelectric signals can be of following three types :-

(i) to train an auxullary muscle which is otherwise not performing any other function,

(ii) to train a muscle which is performing some other function of some other parts of body, and

(iii) to use ' phantom sensation '.

We have adopted the third method, though the first two methods also have been found generally successful. The number of auxullary muscles are in the body are very few in number, also these are smaller in size and deeply located. Usually it is more difficult to train these. On the other hand a muscle, which is already performing some function can be trained for additional function also. For example deltoid, trapezius, scapular muscles etc. can be trained to operate a prosthetic arm if the amputation is of above-elbow type and this is only the possible control if amputation starts from the shoulder region. Obviously, the amputee will be seen to move these muscles (shrugs, jerks etc.) while operating his elbow or wrist joint or prehension movement. Though these are also termed as voluntary control, the subject looks grotesque for these jerky body-movements and is to be employed only when there is no other alternative.

Our case is that of below-elbow amputation and it is assumed that sufficient length of stump is there below the elbow to fit the prothetic device. By rotating that consequent electronic circuitry is very simple and thus provides high reliability.

In the next chapter we have described our system in general and the details of design are given in the next two subsequent chapters.

stump along its axis, the two movements - pronation and supination - can partially be obtained. Hence we attempt to provide the other four movement FF, FE, WF and WE. The two muscles, extension Carpi radilis longus and its antagoinst one (flex or carpi ulnaris) suffer contraction for the above-mentioned four motions and part of these two muscles are available even in the elbow-zone. Thus in our case the phantom sensation in quite useful to avoid the grotesque movements of body and to work the prosthetic device more by the voluntary 'will'. The phantom sensation is the sensation of a phatom hand which almost every amputee feels (though there are cases of neative phantom also) even after long time of losing the organ. They distinctly feel of closing or opening the palm, bending the wrist etc., and contractions in the muscles of the s tump are noticed. Evidently those become the signal-sites.

The most important advantage which phantom sensation provides is that almost no training is required for the muscles.

Picking up the signal -

Usually there are three types of electrodes : surface electrode, percutaneous electrodes and totally implanted telemetry equipment.

We have used surface electrodes for its ease of application and keeping in view the patients' acceptance.

An electric potential difference exists across any boundary between dissimilar conducting materials, including the boundary between a metallic conductor and the conducting fluids of the body. This potential difference is dependent upon the materials involved and is often considered to be constant for a given metal-electrolyte system. However it is subject to variation of considerable amplitude, some of which are sufficiently rapid to fall within the frequency spectrum of the myoelectric signal. Since electrodes must be used in pairs, the average contact potential is effectively cancelled out and does not affect the measuring system, but this cancellation will not occur for the random fluctuations of that potential. It is desirable, therefore, to employ as electrodes, materials which are known to have low and stable contact potential with respect to body fluids. For external use, a silver-silver chloride electrode, made by anodizing metallic silver or by compressing a mixture of silver and silver chloride into a pellet, is generally considered best in this respect.

Physical movement between the electrode and body tissue often contributes a sizable electric potential disturbance. This may sometimes be reduced by suitable mechanical design of the electrode assembly. One factor contributing to this motion-induced potential, at least with surface electrodes is modulation of the electrode-tissue contact potential.

On the other hand, [illegible] has reported that extensive experience with ECG monitoring does not indicate the practical superiority of ' nonpolarizable ' over ' polarizable ' electrodes.

Keeping all these factors in view, we have employed silver electrodes which are cheap as well as it can be easily made.

The outer layer of the skin is reasonably good electric insulator. While this layer may be removed, either by general abrasion or by local ' skin drilling ', these procedures are not acceptable for the long term, repeated application necessary for myoelectric control. The effective electrical resistance of the outer layers of the skin can be reduced by application of a conducting paste which penetrates that layer to some extent and also increases the effective electrode-skin contact area. At the frequencies of interest, surface electrodes used with suitable electrode paste give an effective electrode-tissue impedance of only a few thousand ohms, which is quite satisfactory. If high input impedance amplifiers are employed, high electrode-tissue impedances can be tolerated. However, high-impedance systems are much more susceptible to electrical interference, and are generally avoided when possible. This is particularly true when the system is to be fitted to a patient who is free to travel outside the controlled environment of the laboratory.

The use of electrode-paste, although desirable for good electrical performance, constitutes a nuisance when the electrodes are applied daily for many years. This factor has been considered by some as sufficiently important to make use of electrode paste totally unacceptable for chronic myoelectric control applications.

Surface electrodes should have a minimum diameter of 5 mm. Hence, where independent outputs from adjacent muscles or from closely spaced areas in one muscle are required, spatial selectivity may be a problem. And finally, a fundamental limitation of all surface electrode systems regardless of design is that only superficial muscles can be used.

A relatively new electrode developed at the NASA Flight Research Centre seems promising for applications where high electrode-tissue impedance may be tolerated in exchange for a thin, flexible, low-mass, highly reliable electrode. This electrode is ' constructed ' on the subject using a rather intricate spray on technique. Application by trained technicians, is said to be accomplished in only 20 secs.

Percutaneous electrodes overcome the problems due to isolation by the skin from the signal source. That is, they permit contact directly to the muscle and a high degree of spatial selectivity. Because the electrode area is usually very small, electrode-tissue impedance is not appreciably lower than with surface electrodes. In fact, if very selective electrodes are required, impedances may exceed 50,000 ohms.

The electrode for control-applications, must be flexible and consequently the needle electrodes widely used for clinical electromyography are not suitable. Instead a small diameter wire is usually employed. [illegible] has been found particularly useful, both bare and with nylon or polyurethane insulation, in diameters ranging from 0.001 to 0.003 inch. Insertion is usually accomplished with a small hypodermic needle, although for some applications a suture needle has been used.

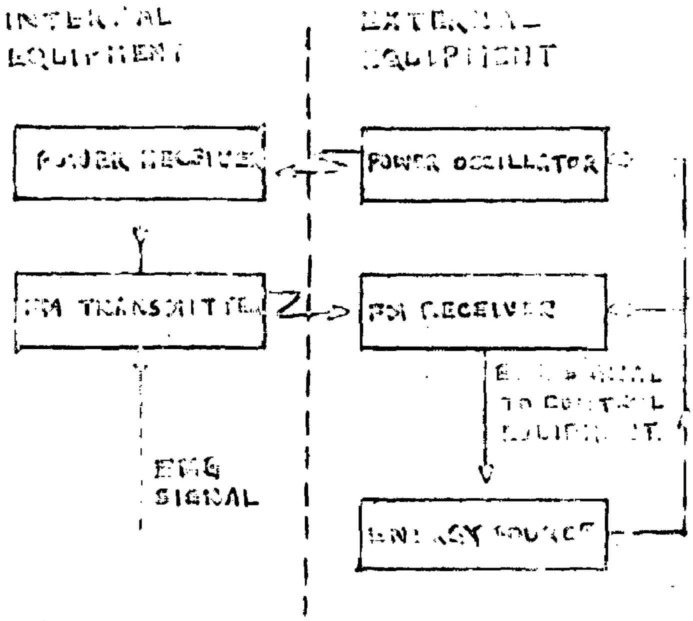

INTERNAL
EQUIPMENT
EXTERNAL
EQUIPMENT
POWER OSCILLATOR
FM RECEIVER
EMG SIGNAL

FIG — 11.

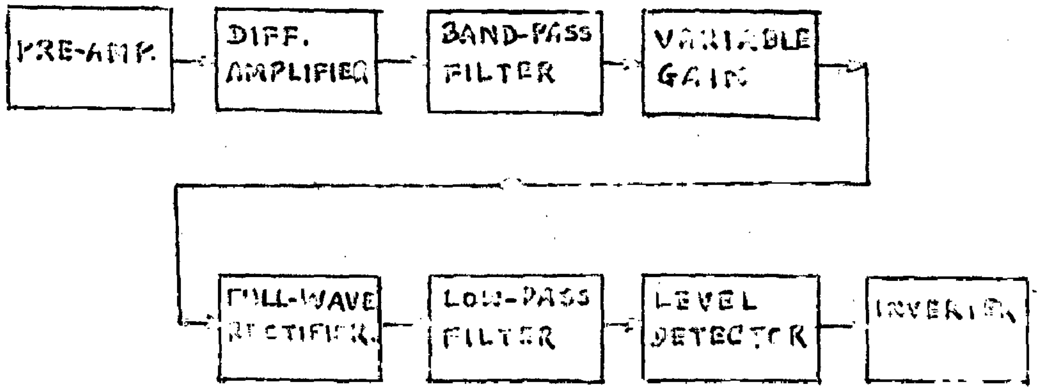

PRE-AMP.
DIFF.
AMPLIFIER
BAND-PASS
FILTER
VARIABLE
GAIN
FULL-WAVE
LOW-PASS
FILTER
LEVEL
DETECTOR

FIG — 12.

used.

Infection has not been a serious problem with fine-wire percutaneous electrodes, but wire-breakage does constitute a problem.

A technique recently developed in Cleveland produces a tiny helix of 0.0011 inch diameter [illegible] wire, polyurethane insulated, with a silicon rubber filler inside the helix. The helix is sufficiently small that it may be passed through a 23 gauge hypodermic needle for insertion, and it is sufficiently elastic, due to the rubber filler, that breakage is effectively discouraged. Development of this electrode may well prove to be a major contribution to myoelectric control research. There has been considerable experimentations on implanted telemetry equipment. Fig.-11 shows such a scheme. Power receiver and FM transmitter are located beneath the skin. The former receives the power by electromagnetic induction from a power oscillator. This power drives the FM transmitter. Myoelectric signals are collected by FM receiver via radio telemetry.

No doubt, this method is more efficient, but costly as well as complex. It also requires surgical operation which may be discouraging from the point of view of patient's acceptability.

Processing the signal -

It has already been mentioned that EMG signals, as it is, is unsuitable to use. Fig-12 shows the block diagram of processing with the level detector at the end of the channel. The functions of individual blocks are shortly explained below.

Pre-amplifier provides large input impedance in order of 10^{12} ohms. Such that source loading is in order of 1 pA (10^{-12} amps.) This large input impedance is required also for the fact that source impedance is in order of kilo-ohms.

Next stage is differential amplifier. It is required for the reason that EMG signals remain almost imerged in noise, particularly the power-frequency noise which the body picks up. Human body, in fact is a good antena for power line frequency noise. The variable potentiometer provides very high CMRR. The gain obtained at this stage is low (about 10). It is for the late capacitor coupling as well as for the fact that polarization potentials, if any, cannot be amplified much at the output terminal of this stage.

The next stage is a band-pass filter. We have employed the resonant type, lower cut-off frequency is 100 Hz and higher cut-off is about 300 Hz. These are for the fact that above few hundred Hz, the EMG signals are very weak. With the lower cut-off as 100 Hz, it does not require the 50-Hz strong rejection filter also. Gain obtained at this stage is 50, thus making the total gain 500 upto this stage.

The next block represents the variable gain stage. Our total requirement of gain is about 90 dB ($\sim$30,000), hence at this stage a provision of variable gain between 4-100 is kept. With the potentiometer in the minimum position, total gain is $10 \times 50 \times 4 = 2000$. By employing a total gain of about 20,000, the EMG signals look as shown in fig-13.

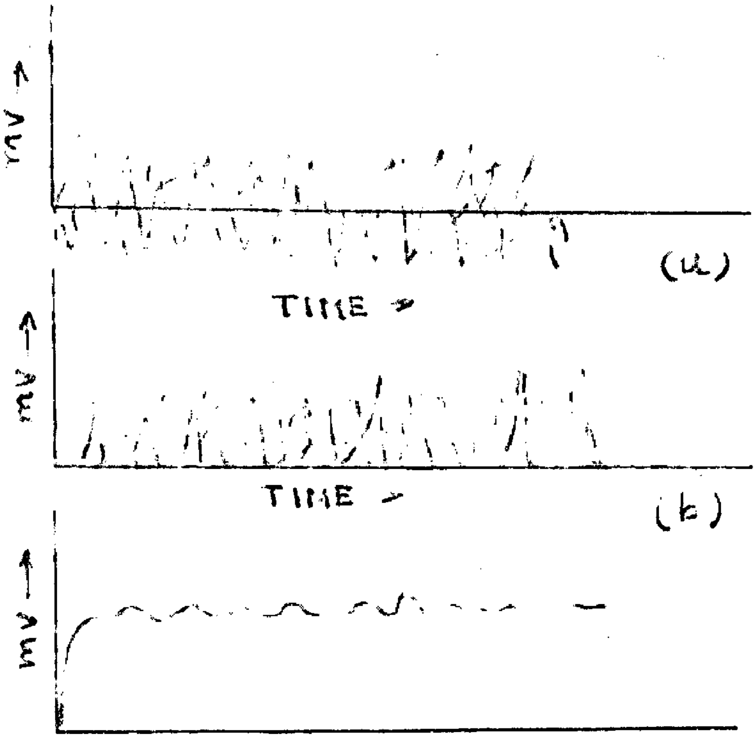
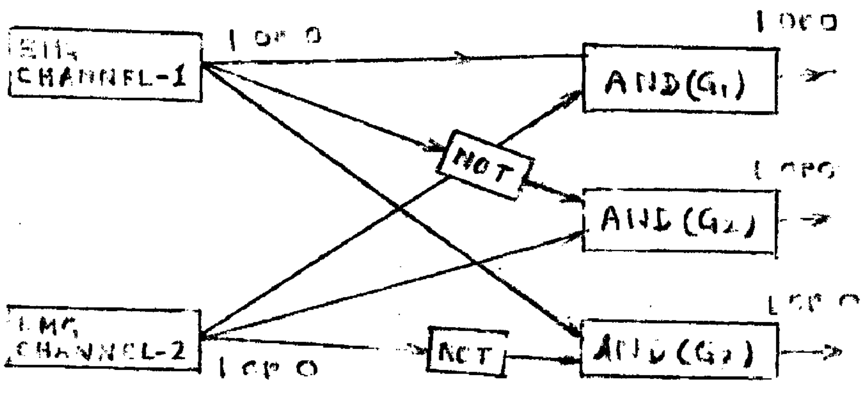


FIG_13.



CHANNEL-1	CHANNEL-2	G_1	G_2	G_3
1	1	1	0	0
0	1	0	1	0
1	0	0	0	1

TRUTH TABLE

FIG_14

It is rectified at the next stage and after rectification the signals look as shown in fig-13. It is evident that, in that pattern also, the signals are difficult to use.

Hence it is low-pass filtered. To reduce the ripples sufficiently and for obtaining suitable speed the time-constant of RC Combination is kept 0.1 sec. As shown in fig-13, the potential is now in utilizable form. At this stage the CMRR can be adjusted such that common-mode potential which is of the nature of d.c. can be kept between 1-2 volts. This is prevented to pass to the next stage by capacitor coupling. With moderate contraction of muscles, the peak of EMG (as shown in fig-13) varies between 1.5-2 volt. The level at the level-detector, at the subsequent stage, is kept about 0.5 volt. With no excitation of muscle the potential at the output terminal is + 12 volt, and with moderate excitation potential is -12 volt (OP-AMP is used in level detector). These two states have been described as 0 and 1 state respectively. An inverter is used as the final block, such that at 0-state, output voltage is -12 volt and at 1-state output voltage is + 12 volt.

Now it is shown how these binary states have been used for separation of functions (eg. LE, UE etc.).

<u>Separation of functions using binary states</u> -

Let P = potential at the end of EMG channel after amplification, rectification and low-pass filtering.

This potential can be expressed as a product of four functions,

$$P = p.b(f).\ e(l).m.$$

where p = participation factor of that muscle in particular function.

$b(f)$ = bland-width of frequency allowed to pass through band-pass filter.

$e(l)$ = position of electrodes on the muscles

and m = magnification allowed. This m can be taken as constant assuming that magnification linearly increases the amplitude of P.

Thus for n number of channels,

$$P_1 = p_1.b_1(f).\ e_1(l).\ m_1$$

$$P_2 = p_2.b_2(f).\ e_2(l).\ m_2$$

. . . .

$$P_n = p_n.b_n(f).\ e_n(l).\ m_n$$

Hence for each type of function, P_1, P_2 etc. will be such that some of the channels will be at 1-State and some at 0-State. For discrimination it is required that the binary combinations will be different for different types of functions.

For n number of channels, total number of combination is 2^n, out of this, all-zero state, otherwise indicate the rest-state, hence cannot be used. Therefore total number of functions by n-channels are (2^n-1). Thus by one channel, only one function is possible, with 2 channels 3 functions and with 3 channels 7 functions.

Since we are to produce six functions of the artificial hand, we require at least 3 channels. For below elbow amputee, four functions are even sufficient as shown earlier. If the fingers are kept extended by spring (such that by myo-electric control it can be closed only, and can be kept closed as long as the muscle remain under contraction) the total number of functions are reduced to three only i.e. FF, WE and WF. These three functions can be produced by two channels and that has been done presently.

The two sets of electrodes have been placed over two muscles, the first one is extensor carpi radialis longus and the second one is flexor carpi ulnaris. Participation factor of these two muscles are 100% for FF. But for WE, participation factor (p) of second muscle is zero and for WF p for first muscle is zero.

If the first muscle is connected to the first EMG channel and the second muscle to the second, then the following binary combinations are obtained

For FF binary combination is 11

For WE binary combination is 10

and For WF binary combination is 01

These three states are separated by logic gates as shown in figure-14.

If more than three functions are required, we are to employ more than two channels. In that case it will require finding out the functions b(f), c(l), p etc. by direct experimentation. Then the potential P can be written in terms of the component functions b(f), c(l) etc, provided

DETAILS OF DESIGN (ELECTRO-MECHANICAL)

The electro-mechanical hand is shown in fig-18. It consists of the basic structure and three magnetic clutches.

The super-structure is made of aluminium sheet to make it as light as possible, the fingers are wooden with aluminium sheet on both sides to make it compact. The basic structure resembles Otto Bock hand but internal design to move the joints are different.

The hand can perform all six basic movements, i.e. FF, FB, WF, WB, P and S, if suitably excited. FF and FB are performed by first magnetic clutch, WF and WB by the second and P and S by the third. Finger movements are performed by the cord and pulley arrangement, and the motion is transferred from the first magnetic clutch. Wrist movements are transferred from the second magnetic clutch via lever, whereas P and S are directly performed by the third magnetic clutch which is mechanically connected to the body of the hand. In all cases, motion in the magnetic clutches are obtained due to rotation of motor (not shown in the figure).

Motor

It is a reversible drive d.c. fractional H.P. motor. Maximum input voltage is 28 volt. In our case the extreme position of any movement brings the motor to blocked-rotor condition. The motor is very old and neither its wattage or current is mentioned. Any how we performed some test on blocked rotor condition to know the rotor current. All these obey at least some sort of emphirical relationship. Afterwards b(f), e(l) etc. of each channel will have to be adjusted in such a way that for different functions of prosthetic hand a new binary combination is produced.

This work is quite lengthy as well as complex and will be dealt with in future.

our operations are of short-time and hence overloading was possible. The following table shows the blocked-rotor test :

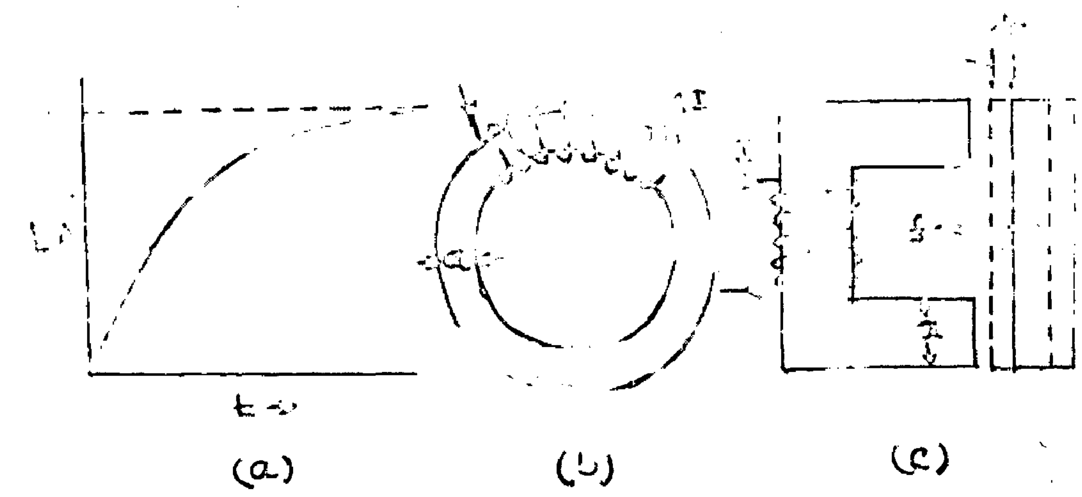
Supply voltage	Running current	Blocked-rotor current
11 V	0.9 Amp.	1.8 Amp.
12 V	0.9 Amp.	2.0 Amp.
14 V	1.0 Amp.	2.2 Amp.

Any how we limited the rotor current upto 2.5 Amp. in all cases to avoid overheating. The motor does not run below 5V and minimum current to run is 0.8 Amp.

The motor is having two field winding. One clockwise and other anti-clockwise. These give reversible drive by change of terminal. Total resistance of the motor including armature and field is 5 ohms.

Magnetic Clutches

This is the main part to design in electro-mechanical system. Here we have used toothed type clutch, hence the air gap between two magnetic parts are more, this leads to more number of turns i.e. to bigger size of clutches. Ultimately it was found that toothless type is also sufficient, the friction between two parts is sufficient to transfer reasonable griping force to the prehension. Hence the size could have been much smaller. However, since this is the first model, optimization of weight was not maintained



FIG–15

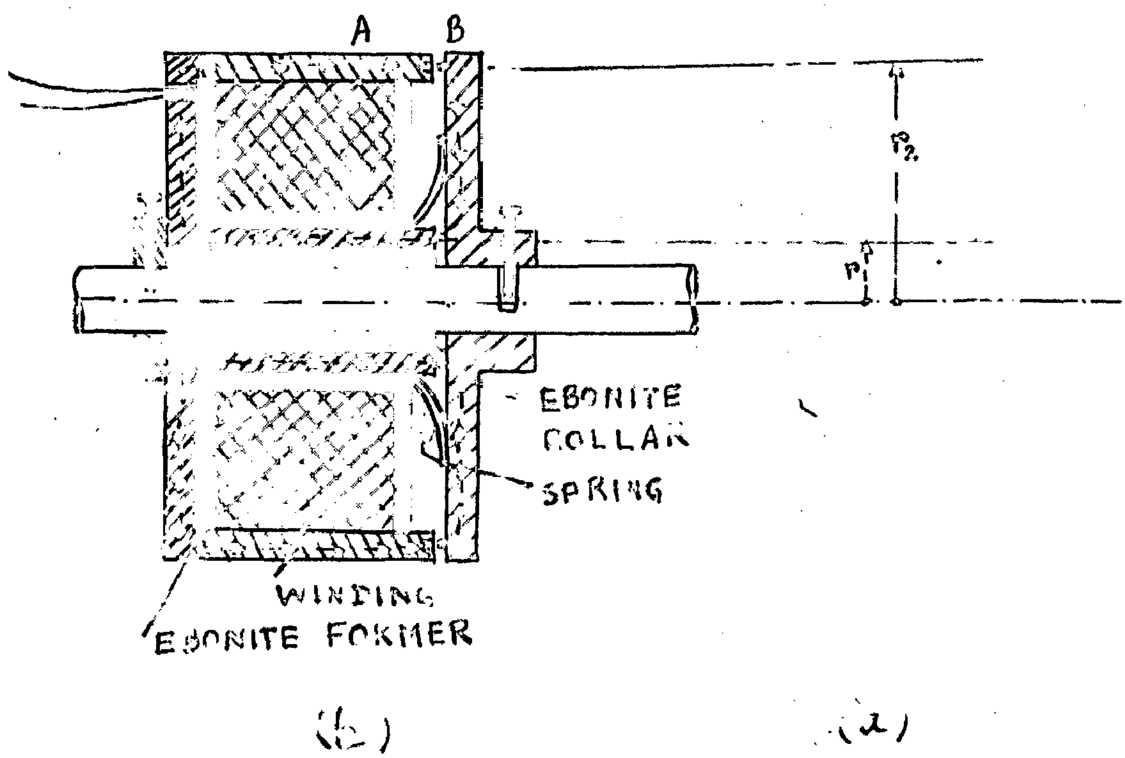


FIG – 16

during the design.

(c) Pulling force of an electro-magnet

Current i increases exponentially in an inductive circuit (fig-15(a)).

Voltage across the coil at any instant $= L \frac{di}{dt}$.

$\therefore$ Power at that instant $= L\, i \frac{di}{dt}$.

Energy stored in time 'dt' $= Li \frac{di}{dt} . dt. = Li\, di$

Let I = Ultimate value of current

By that time, energy stored $= \int_0^I Li\, di = \frac{1}{2} L I^2$.

The ring in fig-15(b) is considered,

l = total peripherial length

I = Current in winding

μ = relative permeability

$H = \frac{nI}{l}$, hence $B = \mu\mu_0 H = \mu \mu_0 nI/l$

$\therefore \phi$ = flux $= \mu \mu_0 a\, nI/l$

$\therefore L = n. \frac{\phi}{I} = n \times$ (flux/amp) $= n^2 \mu \mu_0 a/l$

W = energy stored $= \frac{1}{2} L I^2 = \frac{(nI)^2 \mu \mu_0 a}{2 l}$

Hence energy-density $= \frac{W}{(al)} = \frac{(nI)^2 \mu \mu_0 a}{2l\, a.l} = \frac{1}{2}(nI/l)^2 \mu \mu$

But $B = \mu \mu_0 H = \mu \mu_0 nI/l$

$\therefore$ Energy-density $= \frac{B^2}{2 \mu \mu_0}$.

Let us now consider fig-15(c).

Say, the iron core is shifted by a distance d_x ,

Pulling force (by one pole) = f .

Energy = force x distance = $f.\ d_x$.

Change of volume in air-space = $a.\ d_x$.

Hence, change in stored-energy = $a.d_x.\ \frac{B^2}{2\mu\mu_o}$.

$$\therefore\ a.d_x\ \frac{B^2}{2\mu\mu_o} = f.\ d_x$$

$$\text{or,}\quad f = \frac{B^2 a}{2\mu\mu_o}\ N_w\ .\quad \left[\text{all in SI Units}\right].$$

In our case, the magnetic clutch does not exactly form the ring-path, but the above calculation can be carried out for first-cut design.

In fig-16, the cross-sectional view of magnetic clutch is shown. It is having two parts, A and B. Part B is fitted on the shaft by a pin. The hole in the shaft is slightly bigger than thediameter of pin, such that part B can have axial motion on the shaft at least for 2 mm, but otherwise it will rotate with the shaft when the latter is driven by motor. Part-A is loosely fitted on the shaft and does not share the motion of it. But when the coil is energised, part B is pulled by part-A, the peripherial pins enter into the holes of part A and can forcibly rotate part-A. From this part-A, motion is transferred to the joints of hand by cord or lever as the case may be. Latter we removed the peripherial pins of part-B, and it was found that even frictional force between the two parts is sufficient to exert reasonable force on the joints of hand. All the movements of hand reach the extreme position within one-fourth revolution of part A, such that electrical connection to the coil can be maintained by flexible wire.

The formation of magnetic path is shown in the figure. The collar on the shaft, made of ebonite is provided such that flux cannot enter into the shaft. In that case it will reach the other clutches through the shaft. Part-B is kept at a distance of 1 mm by sprip type of spring, and we require to exert a force of atleast 0.5 N_w on it.

According to the previous calculation we at first find out the area which will exert the force.

From fig-16(a).

$$Y_1 = 3/16 \text{ in.} = 4.75 \text{ mm} = 0.48 \text{ cm}$$

$$Y_2 = 9/16 \text{ in.} = 14.2 \text{ mm} = 1.42 \text{ cm}$$

$$\text{Total area} = (2\pi \times 0.48) \times 0.16 + (2\pi \times 1.42) \times 0.32$$

$$= 2\pi \times 0.32\ (0.24 + 1.42)$$

$$= 2\pi \times 0.32\ (1.66)$$

$$= 3.32 \text{ sq. cum}$$

$$= 3.32 \times 10^{-4} \text{ sq. m.}$$

$f = 0.5\ N_w$ (required) .

$$f = \frac{B^2 a}{2\,\mu\,\mu_0} = 0.5$$

$$\text{or, } B^2 = \frac{\mu \mu_o}{a} = \frac{4\pi \times 10^{-7}}{3.32 \times 10^{-4}} = 37.8 \times 10^{-4}$$

$$\therefore B = 6.15 \times 10^{-2} \text{ Wb.}$$

$$\text{Hence, } H = \frac{B}{\mu_o} = \frac{6.15 \times 10^{-2}}{4\pi \times 10^{-7}} \text{ AT/m} \simeq 50 \text{ AT/mm.}$$

The flux will have to cross air-gap of total length 2 mm.

Hence AT for 2 mm = 200.

At that low flux density, AT required for iron path is insignificant. Therefore a total of 200 AT may be provided.

Considering the space available inside the prosthetic hand, the diameters of various parts are set as follows :

Shaft dia = $1/4''$ = $\frac{4}{16}$ inch

Magnetic insulation = $\frac{1}{16}'' + \frac{1}{16}'' = \frac{2}{16}''$

Iron path(inner) = $\frac{1}{16}'' + \frac{1}{16}'' = \frac{2}{16}''$

Former = $\frac{1}{16}'' + \frac{1}{16}'' = \frac{2}{16}''$

Total = $\frac{10}{16}$ inch.

The inner diameter of clinder = 1 inch, and overall outer dia is $1\frac{1}{4}$ inch.

Therefore, winding space available = $\frac{16}{16}'' - \frac{10}{16}'' = \frac{6}{16}''$

Hence, winding space in one side is $\frac{3}{16}$ inch or 4.8 mm.

Let us use 23 SWG, which is having a normal current carrying capacity of 10 amp, if used as static coil and 1.5 amp. if used in dynamic part. However for short time operation over-loading is possible upto 2 or 3 amps.

Dia of wire = 0.63 ~ 0.64 mm

Winding space excluding insulation = 4.5 mm (Say).

As shown in fig. 17, vertical distance for two layers is (1.73 x radius).

$$\text{No. of vertical layers} = \frac{4.5}{0.32 \times 1.73} \simeq 8$$

$$\text{No. of horizontal layers} = \frac{200}{8} = 25.$$

Length of coil = 25 x 0.64 mm = 16 mm ~ 5/8 inch.

However, the length of former made is 3/4 inch.

The dimensions of various parts of magnetic clutch can now be fixed as shown in fig-16.

It is obvious that, if teethless magnetic clutch is made, air-gap of the order of 0.2 mm is sufficient, and this will reduce the AT to one-fifth of present value. Consequently the size and weight of the clutches will be highly reduced.

(b) Calculation of length and resistance of wire

$$\text{Length of each turn} = \pi D = \pi \times \frac{13}{16} \text{ inch} = 2.56 \text{ inch.}$$

For three windings, Length = 512 x 3 = 43 yds.

For 23 SWG, weight is 5.67 lbs for 1000 yds.

$$\text{Weight for 43 yds} = \frac{5.67 \times 43}{1000} \text{ lbs.} = 110 \text{ gms.}$$

Resistance is measured by ammeter, voltmeter method and is found to be 0.6 ohms.

(c) Inductance and time constant of the coil

$$\text{Inductance} = L = \frac{N^2 \mu \mu_o a}{l} \text{ Henry.}$$

$\mu = 1000$ (for such low flux density), assumed.

$N = 200$.

$a = \pi r^2 = \pi (5/16)^2$ sq. inch $\simeq$ 2 sq. cm. $= 2 \times 10^{-4}$ sq.m.

$l = 3'' = 3 (25.4)$ mm $= 3 \times 25.4 \times 10^{-3}$ m.

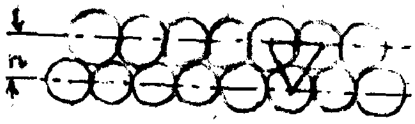
$$\therefore \quad L = \frac{(200)^2 \times 1000 \times 4\pi \times 10^{-7} \times 2 \times 10^{-4}}{3 \times 25.4 \times 10^{-3}} \text{ H}$$

$$= 0.132 \text{ H}$$
$$= 132 \text{ mH.}$$

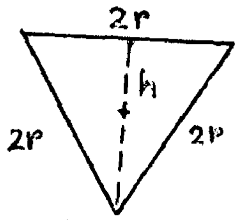
Hence time constant $= \frac{L}{R} = \frac{0.132}{0.6}$ sec. = 222 m sec.

Hence time required for the current to come to 95% of the final steady-state value = 3 x time constant i.e. 0.67 sec.

This is the major component causing delay between contraction of muscle and execution of function by prosthetic hand. Delay caused by electronic circuit is negligible, also delay by rotation of motor is not much. In any case execution of function will be within 1 sec after the ' desire for it ' which is thought to be good and in accordance with the published data by other method like pattern recognition.



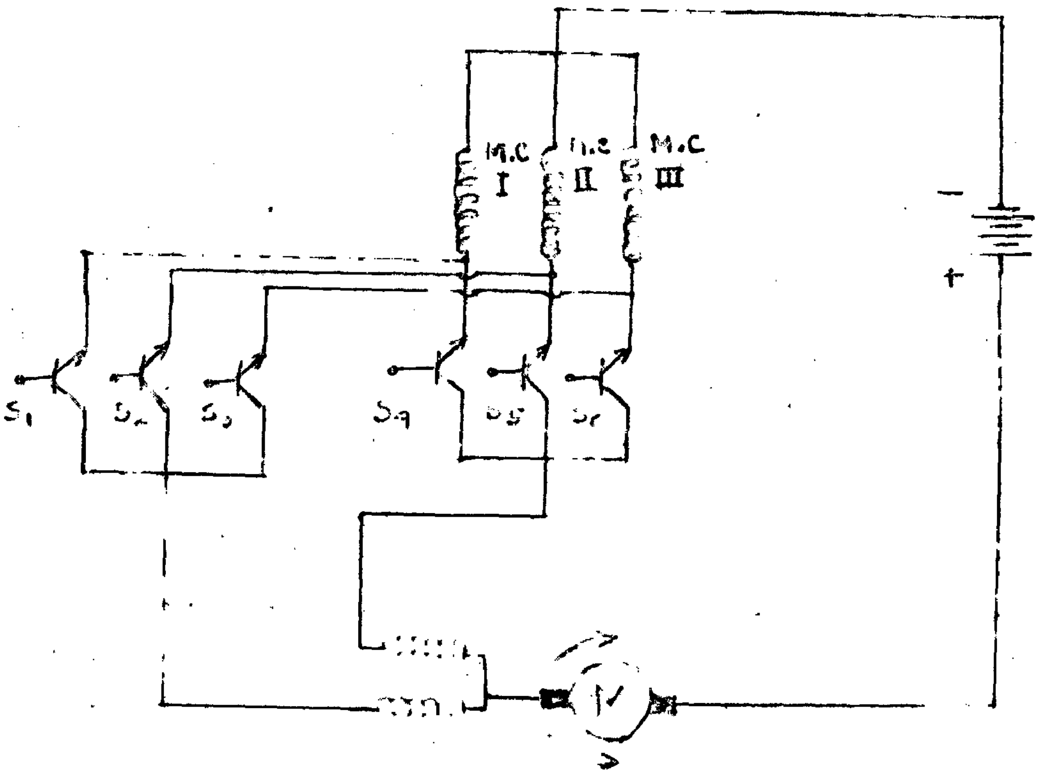
r = Radius of wire



$$h = \sqrt{(2r)^2 - r^2} = 1.73\,r$$

$$\frac{h}{2} = 0.87\,r$$

FIG—17.



FIG—18.

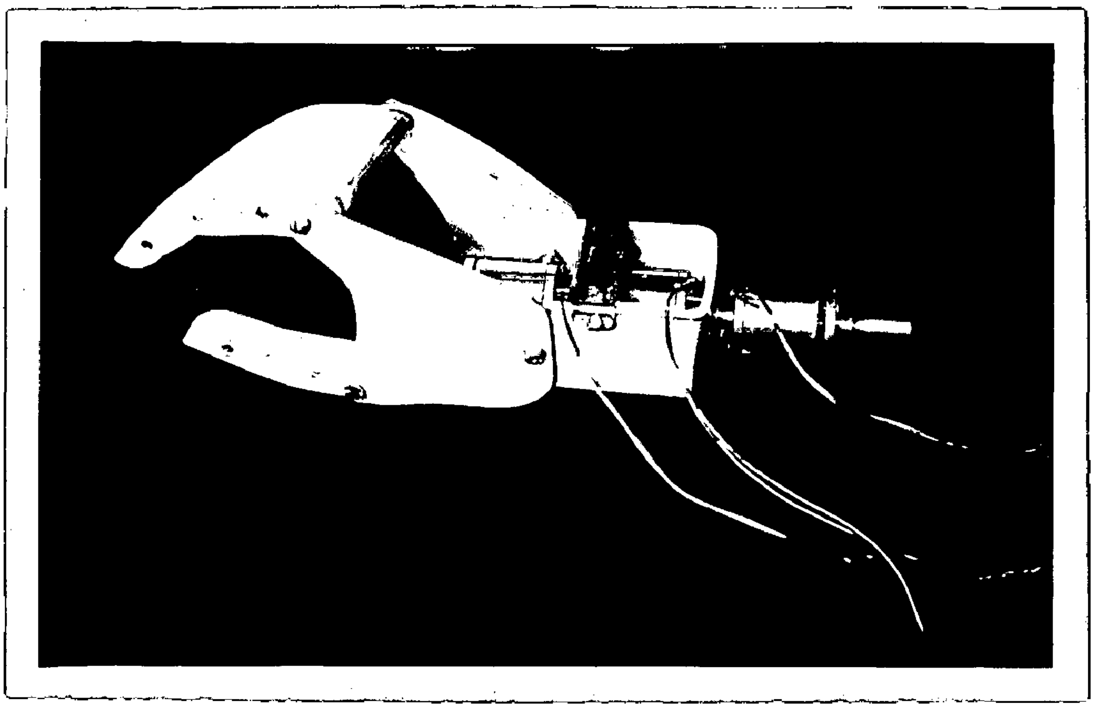


FIG-18.

Directional Control of Current

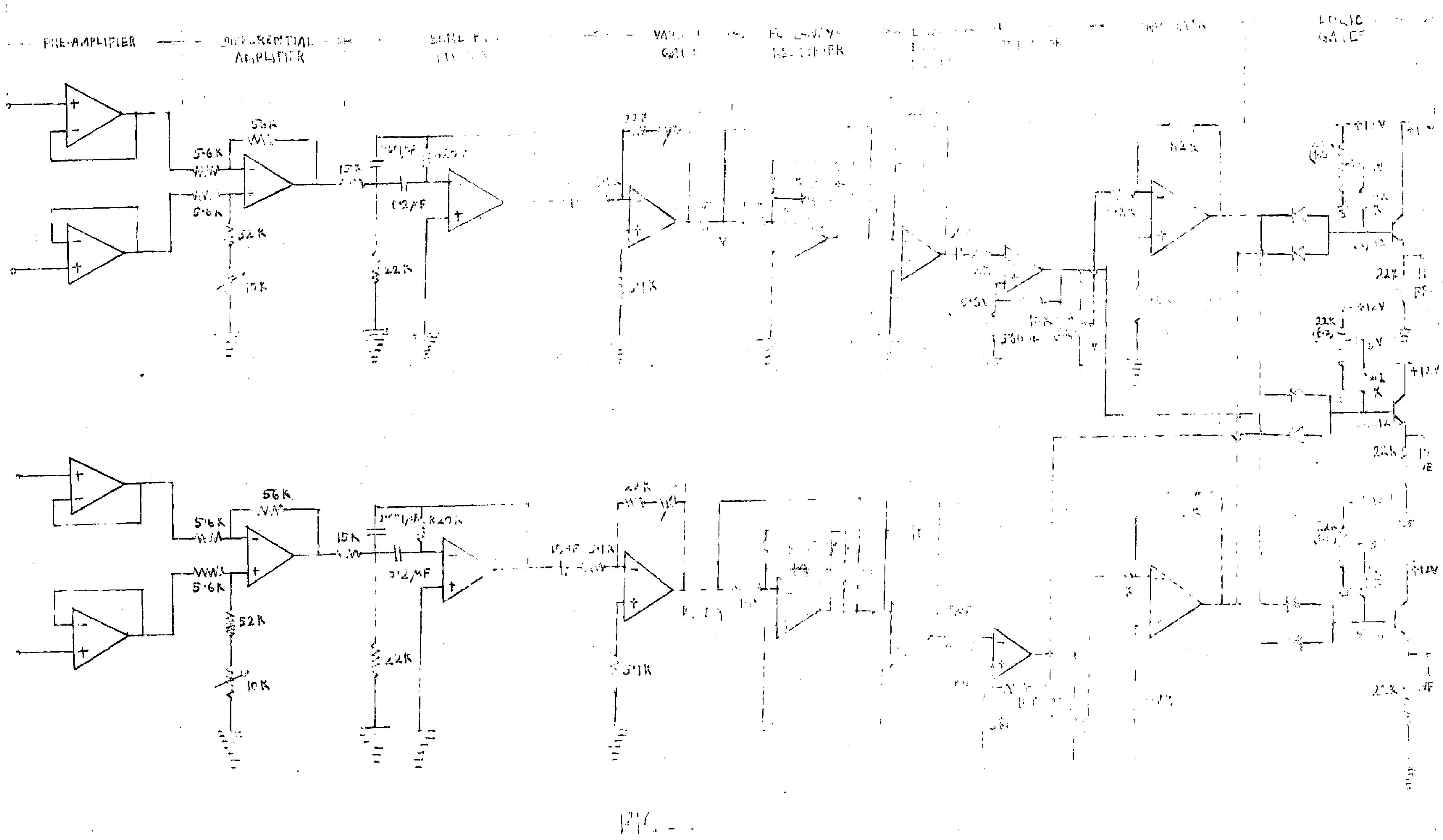
The overall electro-mechanical prosthetic hand is shown in the fig-18 with all pulley, lever and gear connection. During experimentation, the hand is kept clamped which mimics the movement of normal hand. The movement of phantom hand and the movement of a normal hand is said to produce same pattern of EMG in the same muscle. With this in view, the prosthetic hand is moved by the similar movement in normal hand.

Fig-19 shows the connection of motor, battery and magnetic clutches. Here the motor-resistance (5 ohms) is the collector resistance of the transistors, the resistances of magnetic clutches form the emitter resistances. If a 12 volt battery is used, maximum current through the armature is 12/5.6 or approximately 2.1 amp, when transistors are ' on '. For this, voltage required at the base is

$$0.6 + 2.1 \times 0.6 \text{ volt} = 1.85 \text{ volt} \approx 2 \text{ volt only.}$$

Hence if the base is given a minimum potential of 2 volt, transistor is 'on' and motor starts running.

Direction of currents are also automatically controlled. If base S_1 is excited, magnetic clutch-I is energized and motor runs in anti-clockwise direction. If base S_4 is excited, the same M.C.-I is energized but motor runs in clockwise direction due to reversal of flux. Hence the pair S_1-S_4 can control FF and FE. Similarly the pairs S_2-S_5 and S_3-S_6 control WF-WE and P-S respectively.


PRE-AMPLIFIER
AMPLIFIER
RECTIFIER
LOGIC
GATES
5·6K
56K
52K
10K
15K
0·2/MF
22K
+12V

DETAILS OF DESIGN : ELECTRONIC

The complete circuit diagram is shown in fig-20 with all the values of the components. Block diagram is already shown in fig-12 and short explanations were also given. The design aspects are given below :

Follower

It is simply an OP-AMP with 100% feedback,

Gain = $1 + \frac{0}{R} = 1.$

But since the OP-AMP takes very small amount of current-input, the consequent input-impedence is very high. We obtain here the input impedence of the order of 10^{12} ohms.

Differential Amplifier

This is of conventional design with the gain of 10 only. The reason of such low gain at this stage is already given. The potentiometer gives very high CMRR of the order of 80-90 dB.

Band-Pass Filter

The resonant type band-pass filter is used here. Schematic diagram is shown in fig-21.

For calculating the values of the components, the following relationships are used :

$$R_1 C_1 = \frac{Q}{\omega_0 A_0} \qquad \ldots (i)$$

Where A_0 = mid-band voltage gain

$\omega_0 = 2\pi f_0$, f_0 = mid-band frequency.

$$Q = \frac{f_o}{B}$$, where B = Bandwidth (in frequency)

$$R_3 \frac{C_1 C_2}{C_1+C_2} = \frac{Q}{W_o} \qquad \ldots \text{(ii)}$$

$$R^1 R_3 C_1 C_2 = \frac{1}{W_o^2} \qquad \ldots \text{(iii)}$$

$$R^1 = \frac{R_1 R_2}{R_1 + R_2} \qquad \ldots \text{(iv)}$$

Say, lower cut-off frequency = 30 Hz
Higher cut-off frequency = 230 Hz } assumed.

Hence B = 230 - 30 = 200 Hz.

$$f_o = 30 + \frac{200}{2} = 130 \text{ Hz}$$

$$W_o = 2\pi \times 130 = 820 \text{ radian/sec.}$$

$$Q = \frac{130}{200} = 0.65$$

Let $A_o = 50$, $C_1 = 0.001\ \mu F$, and $C_2 = 0.2\ \mu F$.

$$R_1 = \frac{Q}{A_o W_o C_1} \qquad \text{from (i)}$$

$$= \frac{0.65 \times 10^6}{50 \times 820 \times 0.001} = 15.8 \text{ K} \simeq 15 \text{ K}$$

$$R_3 = \frac{Q}{W_o . \frac{C_1 C_2}{C_1+C_2}} \qquad \text{from (ii)}$$

$$= \frac{0.65 \times 10^6}{820 \times 0.001} \text{ K}$$ 800 K (We have used 820K

$$R^1 = \frac{1}{w_o^2 R_3 C_1 C_2} \quad \text{from (iii)}$$

$$= \frac{1}{820 \times 820 \times 800 \times 10^3 \times 0.001 \times 0.2 \times 10^{-12}} \text{ ohm}$$

$$= 9.3 \text{ K}$$

$$R_2 = \frac{15.8 \times 9.3}{15.9 - 9.3} \approx 22 \text{ K}$$

Using $R_1 = 15$ K, $R_2 = 22$ K, $R_3 = 820$ K, $C_1 = 0.001$ µF, and $C_2 = 0.2$ µF (all the components within an accuracy of ± 20%, as was available), we obtained by experiment.

Lower-cut off	=	100 Hz.
Higher cut-off	=	270 Hz.
Mid-band freq.	=	170 Hz.

The discripency between the calculated and experimental values are quite high. Any how, the range obtained experimentally was sufficient for our purpose.

Variable gain stage

It is also of conventional design. The total gain before this stage is 10x50 = 500. Minimum gain at this stage is 4, thus making the total gain 2000. The 1M ohm potentiometer can give high gain upto 200, though it was found that experimentally such a high gain was difficult to obtain. The overall gain the amplifier could provide was approximately 100 dB, which was sufficient for our purpose. High gain also increases the common-mode potential. For

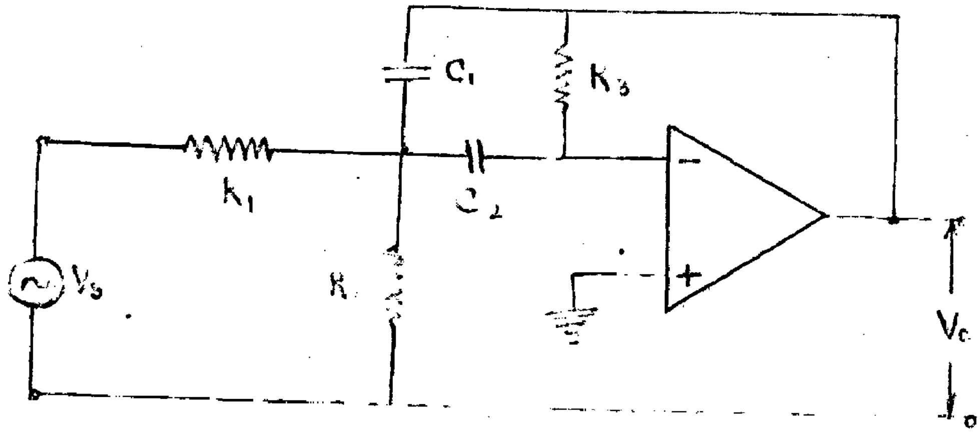


FIG-21.

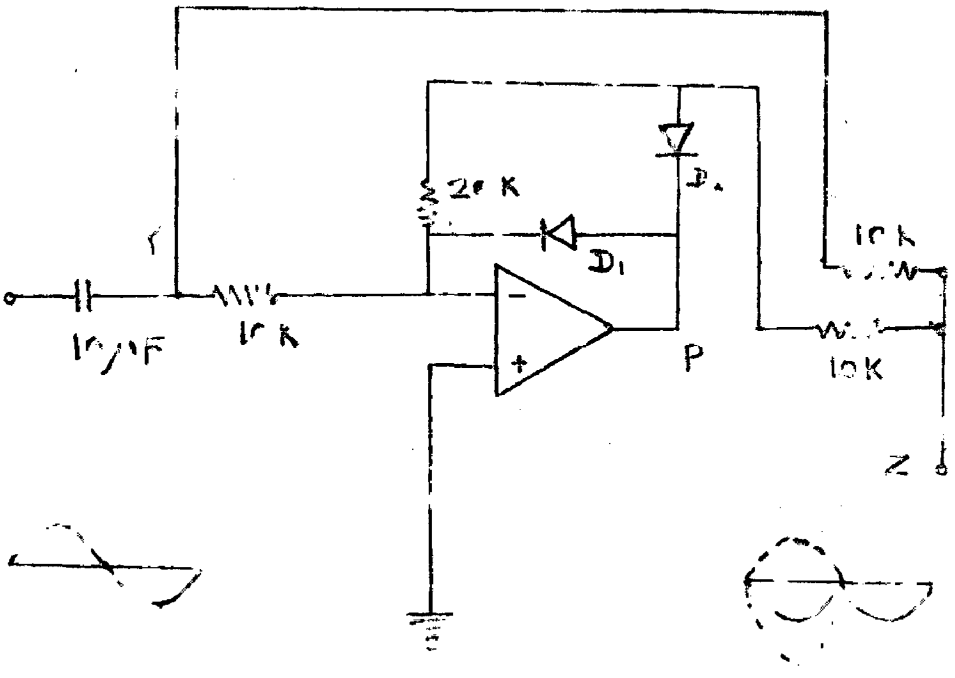


FIG-22.

example, if the amplified common-mode potential is 5 volt, and supply voltage is ± 12 volt, the differential signal can have a maximum amplified amplitude of (12-5) = 7V. When the 5V d.c. is blocked by capacitor, differential signal appears as of maximum 7V output, and more and more gain effectively reduces the amplitude of it.

We adjusted the gain in such a way that amplified common-mode potential remains between 1-2 volt.

Coupling Capacitor

At lower cut-off, impedance $= \frac{1}{2\pi f_c} = \frac{1}{2\pi \times 100 \times C}$

$$= \frac{10^6}{2\pi \times 100 \times 10} \text{ ohms}$$

$$= 160 \text{ ohms.}$$

Hence for C = 10 μF, a series impedance of 160 ohms is put effectively and this impedance is not much compared to the other components.

At higher cut-off, impedance is $\frac{160}{3}$ ohms or 53 ohms only.

Rectifier

OP-AMP is used in the process of rectification such that the same ground point is maintained in the next stages also. The process of rectification is shown in figure-22. When the terminal is having some positive potential, the output terminal P of OP-AMP will have negative potential, since the input is given to the inverting terminal. The diode D_1 will be reverse-biased but D_2 will be forward-biased. Hence the feed-back resistance 20K will come into action. Thus the output point Z will have two potentials, one is direct input positive potential, and the other is negative potential double the magnitude of positive one. Hence the whole positive half-cycle of input-point Y will be converted to negative half cycle at the point Z.

When the point Y will have negative potential, the point P will have positivity. Hence diode D_1 will be forward-biased, and the point P will remain at earth-potential. But due to direct connection between Y and Z, the latter point will continuously trace the same negative potential of Y. Hence the overall connection will give the effect of rectification, because the negative half is retained and as if the positive-half is inverted.

Low-Pass Filter.

Figure-23 shows this section individually. We are to keep the time constant 0.1 sec. or

$$RC = 0.1 \qquad \ldots (1)$$

Let d.c. gain = 2. (assumed)

$\therefore$ R = 10 x 2 = 20 K (we used 22 K)

$$\text{From (1), } C = \frac{0.1}{R} = \frac{0.1}{20\times10^3} F = \frac{10^5}{20 \times 10^3} \mu F = 5 \mu F .$$

In actual circuit, however, we had to use C = 10 μF for better result , i.e. time-constant was actually kept

0.2 sec.

We can calculate the cut-off frequency as follows :-

Say Z = equivalent impedance of R-C combination

or $\frac{1}{Z} = \frac{1}{R} + j\omega C = \frac{j\omega RC + 1}{R}$

$\therefore$ $Z = \frac{R}{j\omega RC + 1}$

At cut-off frequency gain is $\frac{2}{\sqrt{2}} = \sqrt{2}$

or $\frac{R}{j\omega RC + 1} - \frac{1}{10 x 10^3} = \sqrt{2}$

or $\frac{R}{j2\pi f_{\circ}RC + 1} = \sqrt{2} x 10^4$

Putting R = 2.2 K, and RC = 0.2

$\frac{22 x 10^3}{j2\pi x 0.2 f + 1} = \sqrt{2} x 10^4$

or 1.56 = 1.26 f + 1

or f = 0.45 Hz .

With this cut-off frequency the d.c. component of rectified EMG comes out at the output terminal subsiding all the ripples. It is also to be noticed that the common-mode signal will also produce d.c. voltage after low-pass filtering, its magnitude being 1-2 volt. It is blocked by 30 μF series capacitor. The output terminal after the capacitor remains at zero volt when there is no differential input. When the muscles contract, the potential rise is between 1-3 volt (more common 1-2 volt) and creates change of state in level detector.

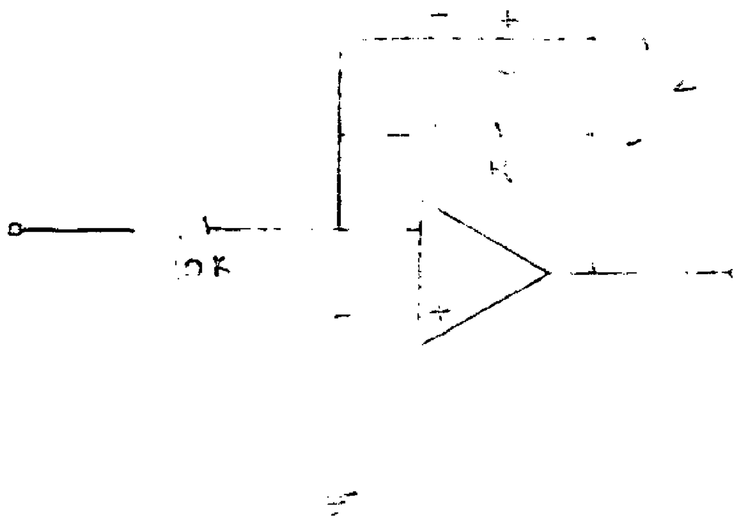


FIG – 23

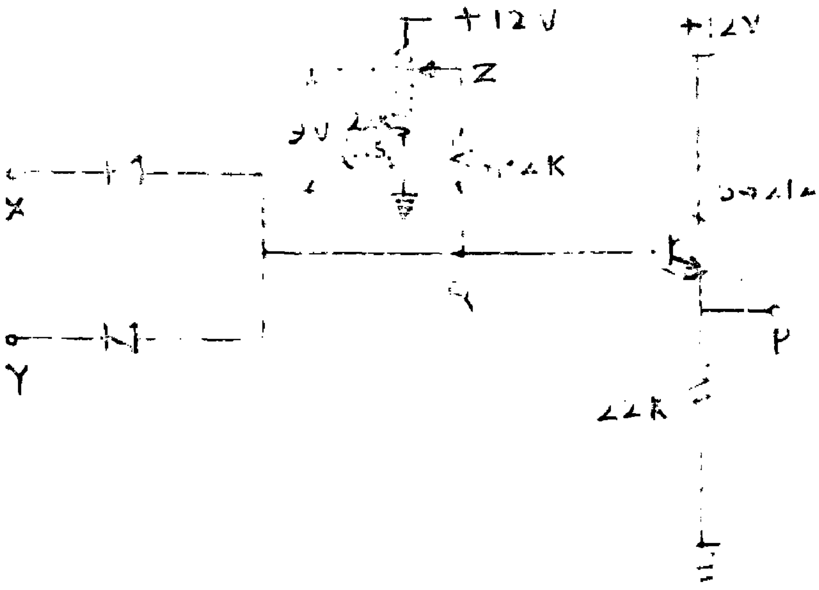


FIG – 24.

The Level Detector and Invertor.

The first one is of conventional design. Invertor is used for NOT gate of logic circuit. Since input voltage to 741 OP-AMP is limited to half of supply voltage, the output of level-detector is not fully given to the invertor, but a potentiometer is used in between. Thus invertor here is an inverting amplifier, with a gain of about ten. The output will be at -12 volt in rest-state, and when the muscle is contracted, the output voltage is changed to +12 volt.

Logic-gates .

The essential of logic-gates are shown in fig-14 where the truth-table is also given. The actual circuit is shown in fig-20. The diode-logic is used for AND gates. The basic arrangement of AND gate and the driver stage is shown separately in fig-24. The input voltages may be +12V or -12 volt, the point Z is kept at +9 volt. When both the diodes are reverse-biased, the potential of point Q will rise to +9 volt, but if any of the two points x and y are at-12 Volt, the potential of Q will also be at the same level.

When the AND gate is 'on', potential of Q is at +9 volt, and output point P will also be at +9 volt. The driver stage is required, because output current will be of the order of 100 mA.

We have used power-transistor (2N3055) in the motor-circuit. It's specifications are :

S/NPN, 115 watts (with heat sink).

f_{α} = 800 Kc/s. (minimum value).

V_{cbo} = + 100 volt = Max. collector to base voltage, emitter open.

h_{fe} = 20-70 (at I_c = 4 Amp).

It's metallic body serves as collector terminal and the two pins are emitter and base.

Since the collector current is between 2-3 amps, the base current was between 80-100 mA (measured).

Hence the output terminal of EMG channel should be able to supply that much of current. The channel itself cannot supply that much of load, hence the driver stage is necessiated.

The voltage between collector and emitter of SC 212 remains between 1-2 Volt and its rating is 300 mW. Hence even if it supplies 100 mA, it is loaded upto 200 mW.

RESULTS AND PERFORMANCE

At first the electronic circuit was tested by three voltmeters. This is required to train the muscles to avoid erratic signal.

By the logic gates three states 11, 10 and 01 are seperated. The corresponding output terminals are marked WF, WE and WF respectively. Three voltmeters are connected at these three terminals, i.e. between ground and WF, ground and WE etc.

No connection is provided in the begining to the electro-mechanical hand. Only it was desired that with the WF of normal subject, the corresponding voltmeter only should show the deflection, and so as for the other movements.

At first test was carried on the normal hand of the author. One set of electrodes are provided on the extensor carpi radialis longus muscle. These two are active electrodes and made of silver. On the extension of the same muscle the ground electrode, which is made of brass, is put at an approximate distance of 4-5 cm. Both brass and silver has been used as ground electrode and no noticeable difference in performance was observed. The other two active electrodes, made of silver, are put on the flexor carpi ulnaris. The distance between the active electrodes are about 4 centimeters. We have tried to increase differential potential by increasing the distance between two active electrodes. In that case the two electrodes are put to anatomically different sites, hence by this the common mode potential is increased. We have found that about 4 cm is the optimum distance.

Again it was found that the above-mentioned position of ground electrode gives minimum common-mode potential to both the channels and consequently one ground electrode is sufficient for both the channels.

Now with FF, the first voltmeter produced deflection, with WE the second one. With WF, initially, both first and third voltmeter produced deflection. Thereby, the exercise on muscles requires little training and concentration. With little care, which is self-learned during the process, the author was able to produce deflections in three voltmeters by three respective motions of hand. For these three movements, training required was of few minutes only which is very encouraging.

Afterwards, the connections are provided between FF, WE, WF of control circuit to the required bases of transistors of fig-19. The prosthetic hand followed the movements of natural hand, time-delay was not measured, but it is of the order in milliseconds. In fact, the motor should be connected to the shaft of magnetic clutches through gearing. Speed requires reducing with the corresponding increase in torque. It is mentioned that only one-fourth revolution of the magnetic-clutches (or even less than that) are sufficient to complete a motion. If the speed of the motor is 1400 r.p.m., we can even reduce it upto 10:1 or even more, the r.p.m. of shaft will be about 140, and time for one-fourth revolution is approximately 0.1 sec. or 100 millisec. It is even possible to reduce the speed upto 20-30 times with the consequent rise in torque. By this method griping force upto several Kg can be created by a small micro-motor.

In the proto-type prosthetic hand, fabricated by us, the motor was directly connected to the shaft of the magnetic-clutches and not via gearing. Since the reduction of speed mainly effects the delay (between the movement of natural hand and prosthetic hand), the later was not measured in our case. Delay in the electronic circuity is negligible, one more important factor of delay is in the exponential rise of motor current due to inductive effect of winding of magnetic clutch.

If due to erratic firing of muscles, more than one control-terminal provide voltage to the bases of power transistors, two situations may arise. When two bases are fired, the two transistors will be simultaneously 'on'. The current through the motor will remain same like that of single firing. It is mentioned previously that this current is approximately battery voltage divided by motor resistance and limited to 2-2.5 amp for 12 volt battery, because under that condition the whole of the supply voltage is dropped across collector resistance. In the case of dual firing, this collector current will be divided through two transistors and two magnetic clutches.

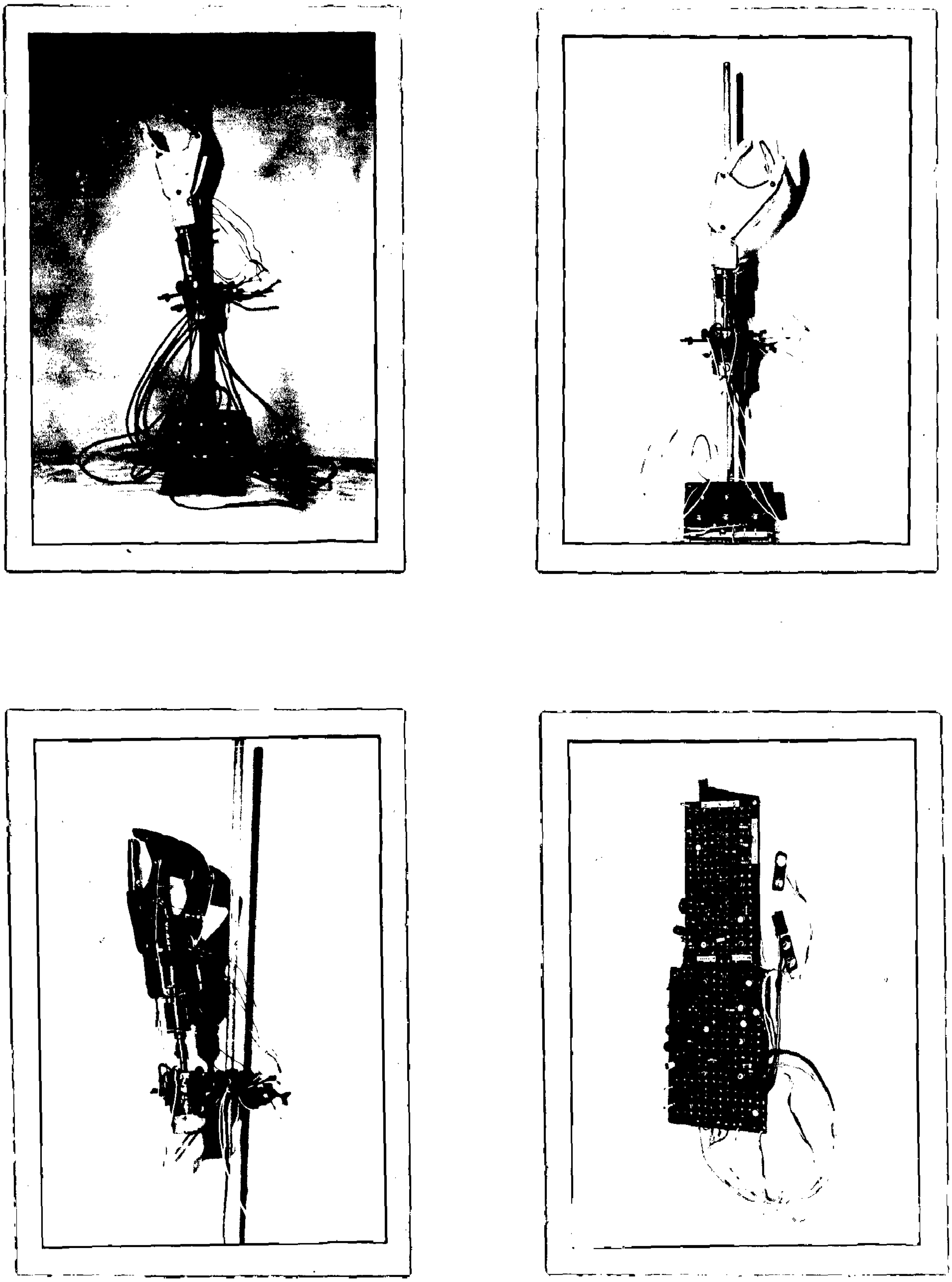


FIG-25.

Insufficient current will flow through the magnetic clutch and the part-A will not attract part-B. In the case of tripple firing, case will be still the same, i.e. no magnetic clutch will operate.

If any two of the group S_1-S_2-S_3 or S_4-S_5-S_6 are simultaneously fired, current will flow through one field-winding of motor. In that case motor will freely run, since no clutch is activated. But if the two fired transistors belong to two different groups (like S_1-S_4, S_2-S_6 etc), equal current will flow through both the field-windings resulting cancellation of flux, hence the motor won't run. In the case of tripple firing like S_1+S_2+S_5, unequal current will flow through both the windings and resultant flux will be small causing faster-running of the motor.

In short, if two or more bases of power-transistors are simultaneously fired dueto erratic signal, firstly no magnetic clutch will operate and secondly the motor may freely run or may not run at all. By all means, these are safe and no damage or overheating will occur in any part.

Thus the reliability or safety of the overall arrangement is quite high.

Several positions of the prosthetic hand are photographed and are shown in fig-25.

OTHER APPLICATIONS OF EMG-CONTROL

Positive phantom is not essential, and let us consider if the phantom of the patient is negative?

Training of the muscles unrelated with the function is probably the only means in that case. Of-course, the muscles related with the function can also be used, e.g. the same pair of muscles used in this work can be trained to provide FF, WE and WF as usual. Or the biceps-triceps pair can be trained.

If the amputation starts from the shoulder region, phantom sensation is of no use in that case. The muscles like trapezius, scapulla, etc. are to be trained in that case.

Use in Orthotics

It is the stimulation of paralysed muscle by myoelectric control. Say, the below-elbow portion of a patient has been paralysed. The hand can operate if electric stimulation is provided to the appropirate muscles. Here also EMG signals are picked-up, processed and is used to control the stimulation to the paralysed muscle. This work is now being pursued in United States, Yugoslavia, and Italy and shows considerable promise. Primary difficulties lie not with the mode of control, but rather in the development of efficient means of stimulating the paralysed muscle. A major improvement in stimulating technique would almost assuredly lead to wide use of myoelectrically controlled electric functional stimulation.

Military and Industrial Application

The research on myoelectric control of aids to the physically handicapped has been accompanied by a variety of programs intended to apply myo-electric control to military and industrial applications. Proposals have ranged from machine-tool control and control of remote manipulators to a control of powerful ' exoskeletons ' or a large number of powerful robots by a single operator. Between these extremities lie suggestions for ' Servo-restraint ' or 'Servo-boost ' systems to aid pilots of high-speed aircraft during severe turbulence or high accelerations.

Most of these systems are intended for use by normal individuals, highly skilled in specific tasks, but hopefully without extensive training in the operation of a myoelectric control system. Since they are part of operators 'job', rather than a permanent addition to his person, the use of surgically implanted electronic equipment would seem inappropriate. However, most proposals will require a large number of separate myoelectric signals. If a robot is to mimic its operator faithfully, it is apparent that the output of most of the operator's skeletal muscles must be monitored. Crosstalks among outputs of various muscles make this impracticable with surface electrodes unless some sort of pattern-recognition computers are employed. Pattern-recognition, in turn, adds greatly to the complexity of the system and may require extensive set-up procedures for each operator. Further, the reliability requirements for the military applications proposed are very severe. It is hard to reconcile these requirements with proposals involving dozens of surface electrodes, applied with considerable haste and subjected to violent movement.

CONCLUSIONS

There are few drawbacks of myoelectric control system from the layman's point of view :-

(i) The movements of fingers or wrist or any other type looks like that of cartoon-picture and may also look ridiculus. Since graded stimulation with appropriate feedback is still a far hope, the quick grotesque robot-like movements may not be liked by the patient. Of-course, by reducing the speed of the motor a resonably slower movement can be provided but that will increase the 'delay' as explained earlier. However a compromise of speed is required here.

(ii) Myo-electric control systems are all externally powered. The power-package is indeed a problem. The rechargable nickel-cadmium cells are the most appropriate to use here. The amp-hour capacity is to be fixed by calculating the power-consumption of motor and of electronic circuitry. The capacity should be such that eight-to-ten hours in the day time continuous work is possible. The batteries will be charged during night when the patient will sleep. Even with this the power-pack with the processing circuitry may weigh near 1 Kg and this the patient must carry. A specially devised waist-belt is proposed here.

(iii) The overall weight of the prosthetic hand may be about 500 - 600 gms with our device for six motions. This weight is however not much and the patient can carry it conveniently. The weight of the lost hand may be more than that but the artificial appendage seems to be a burden where as the natural hand does not. However, with improving technology further reduction in weight is possible.

(iv) The approximate cost of the device described in this work will range between Rs.2500/- to Rs. 3000/- including the battery (if the motor is replaced in our device, 500 mA-hour rechargeable batteries can be employed). For the average people of our country this seems to be rather costly.

Apart from all these factors, the manufacturing of myoelectrically-controlled artificial hand is reasonable for its novelty - i.e. control of hand by the direct 'will' of the patient. The satisfaction obtained thereafter over-shadows all other drawbacks.

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