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MYO-ELECTRIC CONTROL

OF

PROSTHESES FOR MID-FOREARM AMPUTEES

AND

FOR ORTHOTIC HAND SPLINTS FOR QUADRUPLEGICS

A DISSERTATION PRESENTED TO

THE DEPARTMENT OF ELECTRICAL ENGINEERING

OF THE UNIVERSITY OF THE WITWATERSRAND

IN FULFILLMENT OF THE

REQUIREMENTS FOR THE DEGREE

OF MASTER OF SCIENCE IN ENGINEERING

BY

COLIN ROCH

(SEPTEMBER 1968.)

I hereby declare that the dissertation entitled "Myo-electric Control of Prostheses for Mid-Forearm Amputees and for Orthotic Hand Splints for Quadruplegics" is my own work and has not been submitted to any other University for a degree.

Colin Ruch.

27th September, 1968.

TO

THE NATIONAL COUNCIL FOR

THE CARE OF CRIPPLES

IN SOUTH AFRICA

PREFACE.

The following quotation written by Dr D.C. Simpson, M.B.E. B.Sc., Ph.D., one of the foremost prosthesis research workers of today, forms the basis for the developmental work presented in this thesis.

> "When some part of the normal bodily equipment is missing an attempt is often made to bring the person back into line with his fellows by providing him with a prosthesis to take the place of the absent component. The success or failure of this attempt will depend on the nature of the part and how closely the replacement matches it both in its place in the function of the whole body and its cosmetic appearance'

At all times Dr Simpson's ideals were borne in mind in an endeavor to develop a myo-electrically controlled hand prosthesis for mid-forearm amputees.

C. RUCH
SEPTEMBER, 1968.

ACKNOWLEDGEMENTS.

In May and June of 1966, the author was sent to England and Europe to visit the major prosthetic research centres to gain first hand knowledge of progress in the field of prosthesis control, particularly of myo-electric control, applied to prostheses designed for mid-forearm amputees. Three types of myo-electric hand prostheses were seen abroad. They were:-

a) The Russian Type which was the first ever myo-electric prosthesis developed,

b) the Italian type, which is very similar in its control system to the Russian system, and,

c) the Bottomley myo-electric hand prosthesis, which utilises the most sophisticated type of control yet developed for hand prostheses.

Much time was spent with the technicians, engineers and clinicians involved with the development and fitting of the various types of myo-electric prostheses and much was learned from them. On returning to South Africa, the author felt that he now could draw on his newly gained knowledge and experience to develop a type of prosthesis which would, as far as practicable, embody the most salient features of the types of prostheses seen.

The author would like to express his deep appreciation to the National Council for the Care of Cripples in South Africa, for their financial assistance, without which the above tour could not have been undertaken, and without whose continuous interest this project could not have been brought to fruition; to Dr. Morris Milner of the Department of Electrical Engineering of the University of the Witwatersrand, for initially stimulating the author's interest in the new science

of Bio-engineering, and for his continuous and able supervision of the thesis project; to Professor G.T. du Toit, Head of the Department of Orthopaedics of the University of Pretoria, and the H.F. Verwoerd Hospital for making facilities available through the University and Hospital for the research programme, and for his continuous patronage and encouragement when problems seemed insurmountable.

The author wishes to thank the Department of Electrical Engineering at both the University of the Witwatersrand and the University of Pretoria, for the loan of equipment used in the research programme; also, Mr van Rensburg and the staff of the Electrical Engineering Workshop Witwatersrand University, for help in constructing experimental prostheses.

Thanks are due to Mr Ball at the Government Limb Fitting Centre at Auckland Park Johannesburg, and his staff both in Johannesburg and at the Orthopaedic Workshop at the H.F. Verwoerd Hospital Pretoria, for their assistance in fitting the experimental prostheses and ortheses for those patients without whose confidence and perseverance this project could not have been brought to a successful conclusion; and to the authors colleagues in Pretoria, and fellow engineers, appreciation is expressed for their constructive critisms, comments and advice.

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CHAPTER 1.

1.1 Statement of the Problems.

A prosthesis system for mid-forearm amputees, over which the amputee has control of the speed of operation of the terminal device and its force output, was the main aim of this project. The control was to be effected by utilising the remaining muscles in the amputation stump. The myo-potential occurring in the control muscles when they are volitionally contracted was to supply the actual control signal. The ultimate conclusion of the project was to result in the development of a practical myo-electric hand prosthesis which could be produced locally for the mid-forearm amputee.

The following criteria were to be borne in mind in the system development:-

1.1.1 The system developed must be functionally and cosmetically acceptable to the patient. The terminal device was to have a grip strong enough to enable the wearer to obtain sufficient work output from the hand. The prosthesis had to look like a normal hand, and the noise level of the prosthesis motor and the associated gearing mechanisms, should be kept to a minimum.

1.1.2 The patient must be able to afford to pay for the hand prosthesis. That is, the prosthesis and its control system should be able to be manufactured at a realistic price. This was to be accomplished by economic circuit design and the use of economy components. In no way however, should the function or the reliability of the system be impaired by using inferior components.

1.1.3 The patient should at all times be able to set up the electronic system with a minimum of bother, and should be able to make slight adjustments to the

sensitivity of the system if this should become necessary. It must be possible to make all required adjustments without the aid of qualified technicians.

1.1.4 The maintenance required for the prosthesis must be kept to a minimum, and intervals between servicing of the system should be as long as possible, say, six or nine months at a stretch. Day-to-day running costs must also be kept low.

1.1.5 The size and bulk of the electronic pack and the batteries required to power both the prehension motor and the control system, must be kept to a minimum. Where possible, the electronic system was to be designed to fit into the prosthesis socket.

1.2 Myo-electric Control of Powered Hand Orthoses.

At the Orthopaedic Department of the H.F. Verwoerd Hospital where developments relating to section 1.1 were carried out, 14 - 16 quadriplegic patients, i.e. patients who are paralysed in both upper and lower extremities, are admitted per year. It was felt that any study into myo-electric control should include an investigation into the possibility of using myo-electric control for hand splints for these quadruplegic patients. The handsplint could be controlled by utilising any remaining muscle in the arms or shoulders of the patients.

After preliminary investigations, it was decided that myo-electric control could indeed benifit the quadruplegic patient as far as control of powered orthoses are concerned. (An orthesis is an appliance made to aid an existing, but malfunctioning body part, whereas a prosthesis replaces a missing body part.) A number of control systems were developed and are presented in Chapter 6. At the outset, it is stressed that the systems presented are but first attempts at the myo-electric control of powered orthoses and that there are

a great number of problems, both technical and psychological, which must be overcome before the systems can be considered to be perfect.

It is felt however, that myo-electric control has an important role to play in orthoses control, perhaps an even more important role than in the control of powered prostheses. This point of view is encouraged by American Space research workers viewing myo-electric control of powered arm and hand splints as a possible method of overcoming astronaut immobilisation due to the excessive gravitational forces experienced during space acceleration or deceleration.

CHAPTER 2.

ACTIVE PROSTHESES.

2.0 Brief Historical Background.

Amputations resulting mainly from injuries sustained through war have been practised for hundreds of years. Soldiers needed amputations to remove shattered limbs to prevent gangrene and sepsis. The use of prostheses, designed to take the place of these lost limbs, are consequently as old as amputation surgery itself.

The large number of amputations carried out in World War I, (12,000 patients alone were treated at the Limb Fitting Centre in Roehampton, London), resulted in the development of new types of artificial limbs and in the improvement of limb fitting techniques.[1] Particular attention at that time was paid to the design of artificial legs, and it was only towards the end of World War II, that prosthetists (limb-fitters) began developing adequate artificial hands.

In todays highly mechanised society, amputations annually are on the increase despite stringent industrial safety requirements. Further impetus to the study of prosthetics was given by the unfortunate occurrence of the thalidomide children. Modern technological advances are employed in attempts to design newer and more functional artificial limbs.

Hand prostheses that are functional, i.e. have articulated thumb or fingers, are termed 'active prostheses' while prostheses that fulfil merely cosmetic purposes are termed 'passive prostheses'.

2.1 Basic Requirements of a Prosthesis.

When providing an amputee with a prosthesis a number of points must be considered to ensure return to the patient

of as much function of the missing body part/s as possible. The patient should be able to learn how to use his new limb quickly, and with a minimum of restriction of natural movement patterns. The prosthesis should at no time become a burden and/or a physical hindrance, and it should cause a minimum of discomfort physically and psychologically. A hand prosthesis should be functional in all positions of the arm. For example, the amputee should be able to operate the prosthesis with the elbow fully extended equally as well as with the elbow fully flexed; he should be able too, to control the prosthesis with his arm behind his back. It is extremely important for the amputee to have complete control over the prosthesis as far as its positioning, speed of operation, and force exerted by it, are concerned.

2.2 Upper Extremity Prostheses.

The three types of upper limb prostheses in present use are:-

1) Mechanically operated - using body movement for operation and control.

2) Pneumatically controlled and powered - using compressed CO_2 gas as the motivating power.

3) Electronically controlled - electrically powered, using stored battery power.

2.2.1 Mechanical Prostheses.

In the case of mechanical prostheses,[2] the hand or other terminal device, such as a split-hook, is attached to a socket, which is fitted over the stump of the amputee. A snap-catch incorporated in the socket permits interchanging of terminal devices, hand or split-hook, to suit the cosmetic or functional requirements of the wearer. This mechanism, also permits passive rotation

of the terminal device and hence, it may be fixed in any desired radial position. In the case of the hand prosthesis, the thumb, and the index and middle fingers are usually articulated at their base, and, in the rest position, are in a state of opponence. When wishing to open the hand, the wearer must exert a pull on a cable. The distance through which the cable travels determines how far the hand opens. The socket is held in place by means of a harness which passes across the shoulders of the amputee, and is looped round the contralateral axilla, (fig. 2.1.).

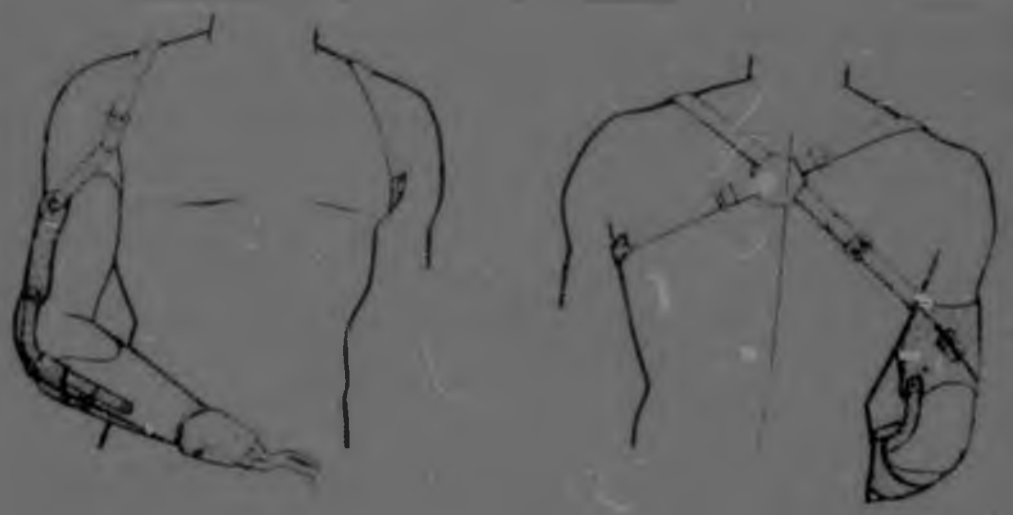


Figure 2.1. Harness Associated with the Mechanical type Prosthesis.

A fixation point on the harness is provided for mounting a cord, which is connected to the cable controlling the hand operation. A movement, such as rounding the shoulders, causes the cord to retract from the terminal device thus opening it. On relaxing, a spring-return housed in the terminal device causes it to close. Because of the large muscle groups in the shoulders, quite considerable force (up to 15lbs.) is available and

is transmitted to the prosthesis.

2.2.2 Pneumatic Prostheses.

Pneumatically operated prostheses[2,3] have reached a high degree of development. The gas used for powering the motor, usually a piston and cylinder type, is almost invariably carbon dioxide stored in containers at pressures at about 750lbs./sq. inch. A reduction valve on the gas container permits a working pressure of approximately 80lbs./sq. inch. Miniature valves are used to direct the gas flow. These valves may be operated in a number of ways. A harness similar to that described in 2.2.1 may be used. The valve forms part of the harness, and varying degrees of rounding the shoulders cause the valve to allow gas to flow to the motor to exhaust the gas, or to restrict gas flow in either direction to maintain a fixed prosthesis position. Similar valves may be operated by actual movement of the stump, or by muscle bulge occurring through voluntary muscle contraction of any suitable and available muscle near the site of amputation.

2.2.3 Electronic Prostheses.

Myo-electric controlled, electrically operated prostheses are just now in the developmental stage.[3,4,5] When a muscle is contracted voluntarily or otherwise, it is found that there is definite electrical activity in the muscle associated with the contraction. Because of the conducting properties of the biological materials surrounding the muscle, minute electrical ccurrents may be picked up on the skin surface in the immediate vicinity of the active muscle. This is known as a myo-electric signal. When a mid-forearm amputee calls to mind his 'phantom hand' and attempts to open it, myo-electric signals are present in the remaining hand extensor muscles. Similarly, when the amputee attempts to close his 'hand' myo-electric signals are found in

the hand flexors.

In the myo-electric prosthesis, the myo-electric signals are picked up by surface electrodes housed in the socket fitting over the stump. They are then amplified, rectified and short-time integrated and used to control hand opening and closing. There is fairly direct proportionality between the myo-electric signal amplitude and the tension exerted in the muscle. This factor might be used in providing a proportional control over the speed and force of the prosthesis operation. (The first myo-electric hand,[7] developed in the U.S.S.R. (cf. Chapter 4.1.), did not make use of proportionality in its control system and a purely 'on-off' system was used. The hand either opened or closed with constant speed, and the motor characteristic and capability determined the force exerted by the prosthesis. Subsequent and more sophisticated prostheses have made use of the proportionality factor (cf. Chapters 4.3., 4.4. and 5.).)

2.3. Prosthesis to patient feedback.

2.3.1 Mechanical Prostheses.

The standard mechanical prosthesis has two main feedback paths. The first is visual feedback where the amputee can, in most cases and assuming that he is not blind, see how the prosthesis is functioning. He can see the effect of the prosthesis on the object being handled. The second feedback path is a sensory one. The extent to which the shoulders are rounded, for example, and how much effort is required to maintain the prosthesis in a desired position, is directly related to the prosthesis action. The velocity and force of the prosthesis is thus directly under the amputee's control, and extremely high degrees of accuracy in velocity, positioning and force output are attainable. To a lesser degree, feedback is also provided by the reaction of the socket against the stump of the amputation.

2.3.2 Pneumatic Prostheses.

With the pneumatic prosthesis, external power is used and consequently there can be no simple body sensory feedback as that present when direct body power is used (as above). Besides the visual feedback an auditory signal is present. This is due to the hissing sound made by the gas as it passes through the valves. Prosthesis operation is extremely rapid, and in most cases accurate positioning of the hand is impossible using standard 'open-shut' or 'on-off' type valves. Force feedback is also impossible with these valves.

Two newly developed valves however, now permit accurate positioning[8] and force feedback.[9] These are:-

a) a position-servo valve which has a tiny control lever, the position of which is synchronised with the degree of extension or flexion of the hand presthesis, and,

b) a force feedback valve which increases or decreases the force being delivered by the prosthesis as the force on the control lever is increased or decreased.

2.3.3 Electronic Prosthesis.

Electrically controlled prostheses permit the most versatility as far as feedback paths are concerned. The first myo-electric and (the Russian type) is however, extremely limited in this respect due to its simple control system. The myo-electric signals merely switch the motor on or off. Auditory and vibrational feedback exist and both types of feedback are provided by the motor drive. As the motor load changes, a change in motor pitch may be heard. Slight vibrational effects are set up as the motor and associated gearbox turn. (Visual feedback is assumed.)

More advanced myo-electric systems[4,5,6] utilising the proportional relationship between the myo-electric signal amplitude and the volitional mechanical muscle activity, have been developed. The velocity of action of the prosthesis has been made proportional to the myo-electric signal amplitude. Should the wearer require faster action of the prosthesis, he merely increases the effort in the controlling muscles.

In another control system (Chapter 4.3.) force feedback has also been provided. Semi-conductor strain gauges, connected in a bridge type arrangement, are mounted on the mechanical linkages opening and closing the hand, and detect the compression and tensile forces in the linkages. The out-of-balance bridge signals brought about by these forces are fed back to a comparator which compares it with the controlling myo-electric signal. If the myo-electric signal is the greater, the driver motor is commanded to increase the force it is delivering. If the myo-electric signal is less than the strain gauge signal, the motor will cause the hand to relax its grip.

Slip sensitive transducers have been incorporated in yet another type electronic hand.[10] When an object held by the prosthesis begins to slip from its grasp (due perhaps to the weight of the object), the slipping is immediately detected by the transducers and the prehension motor is commanded to increase its grip until no further slipping is detected. Feedback here, is limited to the electronics of the hand and is not actually fed back to the amputee.

Pressure sensitive devices may be incorporated in the fingertips and palm of the hand, and the pressure signal may be used to modulate some type of pinching or vibrating mechanism[11] attached to, say, the earlobe or strapped around the arm. The greater the pressure detected, the harder the ear or arm will be pinched. Similarly, an audible tone may be frequency or amplitude modulated by the changing detected pressure. These possibilities

need further experimental investigation.

The table below summarises the nature of feedback tracks available from the different types of prostheses.

MECHANICAL	PNEUMATIC	ELECTRONIC
Visual	Visual	Visual
Position	Position	Position
Force	Force	Force
Velocity	Velocity	Velocity
	Auditory	Auditory
		Vibrational
		Slip

2.4 Additional Remarks on Active Prostheses.

2.4.1 Mechanical Prostheses.

Since the operation of the prosthesis is controlled by the rounding of the patient's shoulders, an unnatural pattern of movement has to be learned in order to operate the prosthesis. For this operation to become normal to the patient, considerable time and practice is required. It sometimes may take many months for an amputee to become proficient in the use of his prosthesis. The shoulder harness which must be worn is always uncomfortable and frequently amputees have discarded the prosthesis completely because of this factor. The harness is unsightly too, and may be a source of embarrassment to the amputee.

Control of the mechanical type prosthesis is not possible

in all positions of the arm. For example, the patient is unable to open the prosthesis when it is held above the head or behind the back - the mechanical positioning of the cable and harness will not permit such action.

A most important feature of the mechanical prosthesis is that the amputee has complete awareness of prosthesis position, speed and force being exerted. The extent to which the shoulders are rounded and how much effort is required to maintain the prosthesis in a desired position, is directly related to the prosthesis action. It is to be stressed that the positioning, velocity and force of the prosthesis are directly under the patient's control and very accurate controls are attainable.

2.4.2 Pneumatic Prostheses.

If the control valves supplying and exhausting gas from the prosthesis are mounted on a shoulder harness, and operated by the patient rounding his shoulder to varying degrees, then the same problems as are associated with the mechanical type prosthesis are present in the pneumatic system. Again patient retraining must take place, and the patient must endure the discomforts and embarrassments of a harness. Added to this, a gas supply cylinder must be carried by the patient.

Prosthesis action is however very rapid and a very substantial force output can be obtained from the terminal device. Specially designed valves do permit accurate position control and force control.

If the valves are so positioned as to operate off the 'muscle bulge' that takes place when a muscle is contracted, and if the control muscles used are the wrist extensor and flexor muscles, then not very much patient retraining is required. This method of control however, is felt to be somewhat unreliable unless the patient has extremely powerful muscles and good control over them.

The supply of gas for powering the prosthesis may present a problem. Special arrangements would have to be made for refilling the CO_2 supply bottles. This might necessitate the patient having to buy a number of bottles in order to have some available while others are being filled. In the light of the experience gained in the laboratory, (Appendix F) it is felt that this kind of powered prosthesis would prove to be uneconomical.

2.4.3 Myo-electric Prostheses and the Reasons for its choice as an Optimum Control System.

The best way to provide control of an artificial hand is to 'intercept' in some way, the message originating in the brain as it travels along its route via the spinal cord and nerves, to the muscles controlling hand and wrist operation. Because of a number of practical and technical difficulties it is not possible to pick up the control signal in the brain or nervous system in a simple way. The electrical activity accompanying muscular contraction however, and the ease with which this electrical signal can be monitored, provide a convenient transducter for 'intercepting' volitional instructions to the hand. These signals can of course only be monitored if the wrist and hand control muscles, or part thereof, are still present after the amputation. The myo-electric control of artificial limbs has therefore particular significance in systems designed for mid-forearm amputees.

It does not take more than a few minutes, and in most cases not more than a few seconds for an amputee to learn how to operate a myo-electrically controlled prosthesis. In fact, he does not have to learn anything at all. He merely has to call to mind his 'phantom hand' remember how he opened and closed it, and then attempt to actually open and close his 'hand' - the prosthesis control system intercepts the brain message manifestation in the muscle and acts accordingly. The above factor saves much time

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and eliminates much frustration for the patient providing him with a simple control system. This is the most important reason for adopting this kind of prosthesis control.

The harness and straps associated with the mechanical prosthesis and its cable system can be completely dispensed with, and the patient no longer has to put up with the discomfort caused by the harness. Similarly with the pneumatic prostheses in conjunction with myo-electric control, the valves and their mountings and harness may be discarded.

Unlike the mechanical prosthesis the myo-electrically controlled prosthesis can be operated in any position of elbow or shoulder orientation. Little difficulty is experienced in providing a power source for the myo-electric system - nickel cadmium rechargeable cells are easily available. Simple electronic techniques only are required to recharge the batteries.

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CHAPTER 3.

GENERAL INTRODUCTION TO MYO-ELECTRIC THEORY.

3.1 The Origin of the Myo-electric Signal.

The existence of 'electricity resulting from muscular contraction' was described by Galvani as early as 1786. This 'electricity', referred to as a myo-electric signal, is detectable using suitable amplifiers across any muscle undergoing contraction.

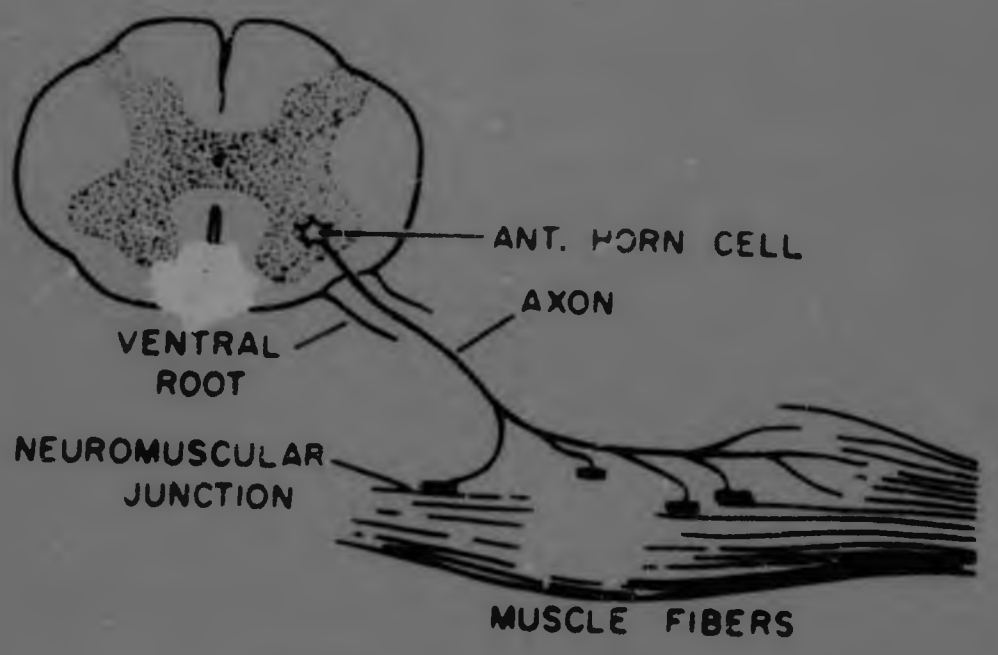


Fig. 3.1. Diagramatic Representation of a Motor Unit.

When we move our limbs, electrical impulses travel along the spinal cord to the nerve cells known as the anterior horn cells. Each anterior horn cell sends a nerve fibre to the muscle where it divides into branches, and may control a large number of muscle fibres. The nerve cell and the muscle fibres it controls are called a 'motor unit'. A muscle contains many such motor units. (figure 3.1.)

According to the membrane theory of nerve conduction[12] the nerve fibre is surrounded by a semi-permeable cell membrane. In the resting state of the cell, the external fluid to the nerve is relatively rich in sodium and chloride ions and relatively poor in potassium ions. In contrast, the cytoplasm or internal fluid of the cell, is rich in potassium, but poor in sodium and chloride ions.

The cell membrane is virtually impermeable to sodium but is relatively permeable to potassium and chloride. As a consequence, there is a difference of electrical potential across the membrane of the cell of the order of 50 - 100 millivolts. The membrane outside is positively charged relative to the cell inside, and the magnitude of the potential is related to the ratio of the potassium and sodium concentration on the inside and outside of the cell membrane. (Sodium ions are, in fact, constantly entering the cell from the surrounding tissue fluid, but are rapidly expelled against their concentration gradient by an ill-understood mechanism which has been termed 'active-ionic transportation'. Whatever the precise mechanism may be, there is much evidence to suggest that this process requires a constant expenditure of chemical energy. This energy is provided by the degradation of organic food materials, particularly carbohydrate, ultimately to carbon dioxide and water, and the energy produced is stored in the form of energy rich chemical bonds.)

When the nerve fibre is activated (by the anterior horn cell), the active region becomes more permeable to the ions and de-polarisation of the nerve fibre occurs. The outside of the nerve becomes more negative with respect to the inside as a result of ion migration across the membrane. A 'de-polarisation' front then travels down the motor nerve at a speed of approximately 40-50 meters/second. After the front has passed, the nerve becomes re-polarised within a few milliseconds.

When the impulse impinges on the myo-neural junctions,

the muscle fibres undergo a similar de-polarisation phenomenon. This depolarisation and subsequent re-polarisation is accompanied by a momentary shortening contraction of the energised muscle fibre. A pair of electrodes attached to a single muscle fibre will detect a bi-phasic signal of the form shown in figure 3.2. as the impulse is propogated along the muscle.

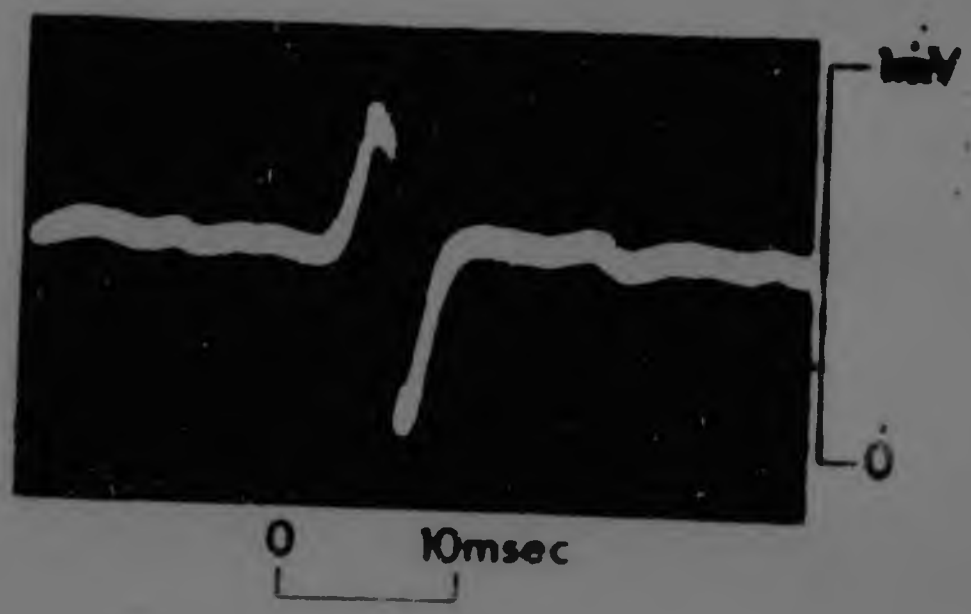


Fig. 3.2. Bi-phasic Action Potential Measured Extra-Cellularly. (Micro-electrodes may be used for inter-cellular measurements).

Generally, when electrodes are placed over a muscle group undergoing contraction, a complex waveform resulting from the summation of a number of biphasic myo-electric signals occuring simultaneously, is picked up (fig.3.3.).



Fig. 3.3. Composite Myo-Electric Signal Monitored using surface electrodes.

3.2 The Frequency Spectrum of the Myo-Electric Potential.

When bi-polar skin electrodes are employed to monitor muscle action potentials, it is found that the wave-form contains frequency components between the range 25c.p.s. to 1,000c.p.s. The frequency content varies in different muscle groups and is dependent on the number of muscle fibres comprising the motor unit.

A number of workers[2,3] have undertaken detailed studies to determine the e.m.g. (electro-myograph) frequency components, and it was not considered necessary or beneficial to verify their findings. Results published by Hirsch, Kaiser and Peterson,(1964), which include the findings of most other research workers, are presented in fig. 3.4. The frequency spectrum of the myo-potentials derived from M.biceps brachii is illustrated. There is maximum activity in the range about 50c.p.s. At higher band-pass frequencies, there is a strong reduction of amplitude.

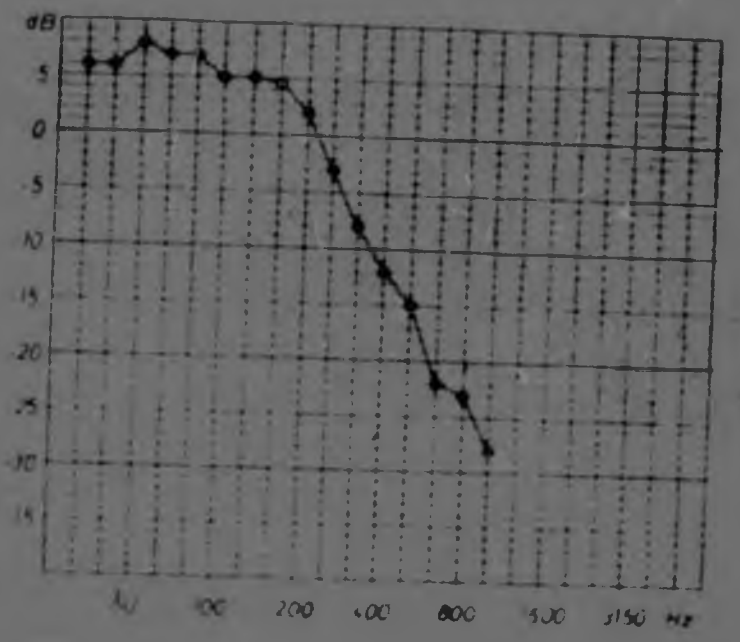


Fig. 3.4. Relative Frequency content Vs Frequency derived from M.biceps brachii (filters of $^1/_3$octave used).

3.3. Factors Determining the Selection of the Optimum Frequency Band-width for Prosthesis Control.

The great reduction of signal content at high frequencies, might indicate that the range about 50c.p.s. would be most suitable for use as the major control signal frequencies. The following factors must however, be considered at this point;

a) Power-line noise, i.e. 50/60c.p.s. interference is a maximum in this region.

b) Noise generated by movement of the electrodes against the skin as the muscles are contracted, has predominantly low frequency components which would also fall within this frequency range.

c) There is greater proportionality between the rectified and averaged signal derived from the higher frequency

components of the myo-electric potential than from the lower.[2]

d) Figure 3.5. published by Kaiser et al[2] (1964), shows in logarithmic scale, the output from four different filters, each having a bandwidth of one octave. The tests were carried out on a person who was asked to keep his muscle contraction constant for 10 seconds. The uppermost curve shows the activity at 50c.p.s., curve 2, at 200c.p.s. Curves 3 and 4 which partly coincide, show the activity at 400 and 800 c.p.s. respectively. (Actually the 800c.p.s. level lies 11dB lower than the 400c.p.s. level.) The distance between horizontal lines corresponds to a change in level of about 40% while the abscica gradations represent a time interval of one milli-second. The peak to peak deviations in the 50c.p.s. filter correspond to a ±20% voltage change; at 200c.p.s., a ±12% change and in the 400c.p.s. and 800c.p.s. filter ranges, respectively changes of ±7% and ±4%. The higher frequency components of the myo-electric signal output are thus 'smoother' than the lower frequency components.

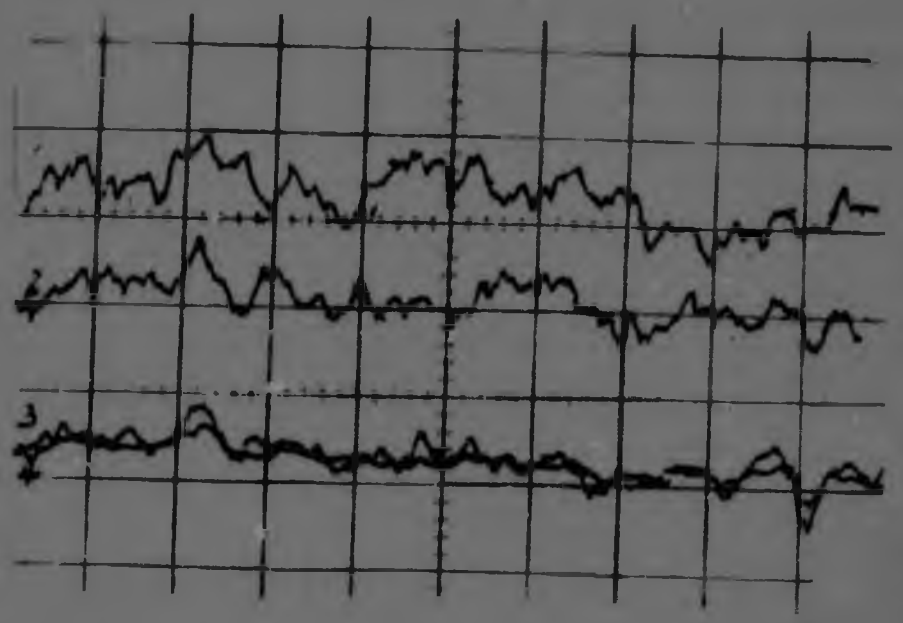


Fig. 3.5.

The above criterea indicate that filters with a pass-band covering the range 300 - 1000c.p.s. would reduce the susceptibility of interference, and the spurious signal fluctuations more prevalent in the lower frequency range, increase the linearity between the muscle tension and the myo-electric output and provide an overall smoother output.

3.4. Pick-up Electrodes for Myo-electric Signals.

A number of factors must be considered when choosing the pick-up electrodes for the myo-electric signals. Three types of electrodes may be utilised when monitoring myo-electric signals. They are:-

a) surface electrodes,
b) percutaneous electrodes,
c) totally implanted transmitter type electrodes.

3.4.1. Surface Electrodes.

The types of electrodes used in the commercially available Russian and Italian myo-electric hand prostheses, (as described in Chapter 4.1, 4.2) are made from metal discs a few millimeters in diameter, and are placed on the skin surface immediately over the muscles to be monitored. The Italian electrodes are made from gold plated brass, and are concave inwards so that an electrode jelly may be interposed between the metal and the skin to reduce skin-electrode impedance. The electrodes of the Russian myo-electric hand prosthesis are probably made of stainless steel and are mounted in a spring-loaded housing in the prosthesis socket to ensure good electrode-skin contact. The Bottomley myo-electric hand prosthesis (Chapter 4.3.) utilises silver electrodes, shaped concavely inwards for the same reason as the Italian type electrodes.

The distance between the electrodes varies according to the type of prosthesis. The distance between disc centres

of the Italian prosthesis is 21mm., and of the Russian prosthesis, 15.5mm. It is the author's experience that the choice of the distance should not be aribtrarily fixed for all patients, but should be chosen to suit each patient individually to scan as much of the muscle being monitored as possible. (Determination of the optimum situation of electrodes would provide a useful research project).

A problem of concern when using surface electrodes is the polarisation of these electrodes. Potential differences appear across the skin-electrode interface, and influence myo-electric signal measurements as far as magnitude and phase is concerned.[4] The polarising potential also contributes to a deterioration of the common mode rejection properties of the myo-electric signal amplifier unless the geometry of the electrode pair, and the electrode-skin interface, are physically perfectly matched. Radeford et al[4] express the opinion that proper electrode selection is vital in a myo-electric system that is not merely an 'on-off' type. In the more sophisticated systems, electrodes should be used in conjunction with suitable electrode jelly to lower skin-electrode impedence and the electrodes should have as large a surface area as possible for the same reason.

Various metals and metal alloys have been suggested for use as surface electrodes to minimise or prevent polarisation. For example, chloridised silver discs have a low polarisation potential as have sintered pulverised silver and silver-chloride electrodes (Beckman type 350050 Bio-potential skin electrode).

An important consideration in choosing the electrode type to be used is the long term effect on the skin. The use of a wet electrode, that is, an electrode used in conjunction with an electrode paste, will obviously enhance the possibility of electrode-skin reaction.

3.4.2. Percutaneous Electrodes.

Basmajian (1962)[5] and Scott (1965)[6] have described electrodes used for myo-electric studies, that are inserted percutaneously, by means of a cannula into the muscle. Leads are left protruding through the skin and the myo-potentials are fed to the amplifier via these leads. The problem of electrode-skin polarisation does not occur with these types of electrodes. It is claimed that these electrodes may be inserted painlessly. The signals obtained are of extremely good quality and even single muscle fibre activity may be monitored in this way. The possibility of multi-channel myo-electric control of prosthesis using percutaneous electrodes is facilitated since muscle groups which may be anatomically close to each other can be electrically separated. The obvious disadvantage of such types of electrodes are the risk of infection and the likelihood of wire breakage.

3.4.3. Totally Implanted Electrodes.

The problems associated with implanted electrodes are so numerous and complex, that many chapters would be required for a comprehensive discussion on all the relevant factors involved in their design. Only a brief description will be presented here.

Bottomley and Riley (1966)[7] have suggested using totally implanted electrodes for monitoring myo-electric signals to be processed by a prosthesis control system. The problem of powering the transmitter is overcome by using the bio-electric signal itself as the power source. The implanted electrode envisaged by Bottomley could, when finally developed, be inserted by means of a 'large size injection' obviating the necessity for the patient to undergo an operation to insert the transmitter type electrode.

The transmitter itself is an extremely simple device.

The active modulating component consists of a varactor diode. The myo-electric signal is impressed across the varactor diode, thus changing its capacitances. The effect of this capacitance change is to change the resonant frequency of the main oscillator circuit. A frequency modulated signal is thus transmitted and picked up by a coil situated in the socket about the stump of the amputation. These electrodes are however, still in the developmental stage and are not available for general use.

3.4.4. Selection of Electrode Type.

For simplicity, as far as manufacture, durability and ease of application, and cost, the surface type electrode was chosen. Stainless steel electrodes are used and the myo-electric amplifier input impedance is sufficiently high so that the effect of polarisation is not considerable. The diameter of the stainless steel electrode discs is $^{1}/_{2}$" and it is not necessary to use any electrode paste with the electrodes. The limited experience gained thus far with these electrodes shows that no skin irritation takes place.

3.5. Requirements of a Myo-electric Amplifier.

3.5.1. Voltage and Current Gain.

The amplitude of the myo-electric signal voltage about the wrist extensor and flexor muscles, that may be detected using surface electrodes is normally of the order of a few hundred microvolts, and may even attain a maximum of 1 or 2 millivolts when the muscle is undergoing strong contractions. The first requirement of a myo-electric amplifier, is then clearly, that it should possess a high gain - high enough to provide an output which can be processed to provide an adequate control signal for the prosthesis prehension motor drive. A minimum voltage gain of 10,000 will be adequate in most cases since the ultimate drive requirements are in the order of 4 volts.

Depending on what type of control system is envisaged for the prehension motor, adequate current amplification must be provided.

3.5.2. Impedance Considerations.

To ensure good matching between amplifier and the 'generator', i.e. the muscle which is considered to behave as an active element surrounded by layers of various biological materials, it is necessary to have some knowledge of the equivalent impedance levels of the generator and the surrounding tissue. Also, the impedance variation at the skin surface interface must be known. An equivalent low frequency circuit of the generator after Bottomley (1965) [8] may be presented as follows (figure 3.6.).

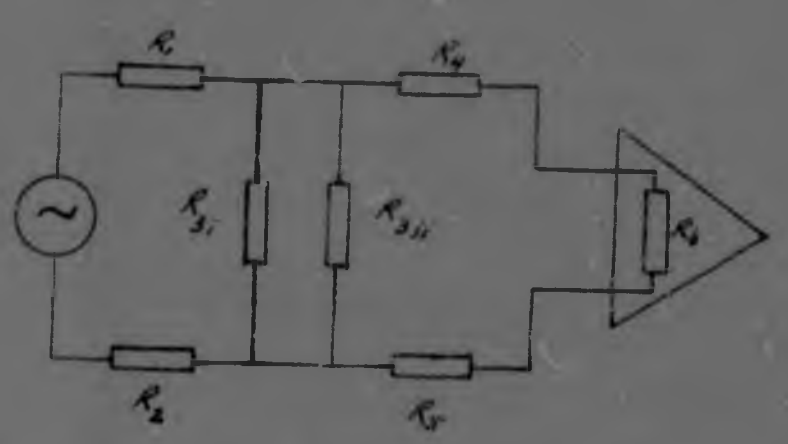


Fig. 3.6. Equivalent Circuit (after Bottomley).

R_1 and R_2 represents the series resistance of the muscle, sub-cutaneous tissue and skin. R_{3i} represents the parallel value of the muscle and sub-cutaneous tissue. R_{3ii} represents the skin resistance between the two pickup electrodes. R_4 and R_5 represent the electrode-

skin resistances. R_6 is the amplifier input resistance.

R_1 and R_2 and R_{3i}, may be assumed to be low and remain approximately constant, while R_{3ii} is known to vary between 5kn - 200kn. Considering a dry area of skin, R_{3ii} will be high; removing the uppermost layer of skin (by rubbing with an abrasive) reduces the resistance to the low value indicated above. Slight sweating of the skin will also reduce the value of R_{3ii}.

R_5 and R_6 will vary as the pressure between the electrode and skin is altered. The pressure change may occur due to the change in muscle diameter as it is contracted (muscle bulge). Sweating occurring at the electrode contact site will also influence R_5 and R_6. The changes in resistance here are considered to be of second order magnitude and will not influence the choice of amplifier input resistance to any marked degree.

From the foregoing it is obvious that the input impedance of the myo-electric amplifier must be high - in the order of hundreds of kilo-ohms. - if loading of the signal source is to be avoided. Input impedances of 1 - 10 meg-ohms are common in myo-electric amplifiers.

3.5.3. Common mode rejection.

Because of the high output impedance of the muscle generator, and its low level of signal output, the problem of A.C. mains interference is of concern. This problem may be overcome by using a balanced input preamplifier stage consisting either of a transformer coupled of differential amplifier type circuit configuration. A common mode rejection ratio of 5,000 to 1 is usually sufficient to exclude interference signals. In the presence of strong A.C. fields, however, it was found that even a common mode rejection ratio of 10,000 to 1 is inadequate to totally reject 50c.p.s. mains interference. In the design of the myo-electric hand prosthesis mains

interference was totally rejected by including a "twin-T'filter having maximum rejection at 50 c.p.s. (Chapter 5.3.1.).)

3.5.4. Frequency Bandwidth.

The frequency components of the myo-electric signal must be considered in amplifier design. The amplifier must be able to amplify all the major information carrying frequencies in the muscle output signal. (Chapter 3.2. describes the frequency component structure of the myo-electric wave-form.) Briefly, it has been established that the e.m.g. signal has components which lie in the range 25 - 1,000 c.p.s. In choosing the appropriate bandwidth of the amplifier, factors such as artefacts, caused by skin-electrode movements, power line interference frequency, and other factors as expanded in Chapter 3.3. must be considered.

3.5.5. Summary.

The necessary requirements of an amplifier for monitoring myo-electric signals using surface electrodes may be summarised as follows:-

1) High input impedance - greater than 1meg-ohm.

2) Balanced input stage.

3) Good common mode rejection ratio - greater than 5,000:1.

4) High gain - greater than 10,000.

5) Suitable frequency response.

6) Suitable filters for eliminating 50/60c.p.s. interference and/or motor noise interference.

In portable systems, the size of the amplifier and its power consumption must also be considered.

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CHAPTER 4.

SOME CONTROL SYSTEMS FOR MYO-ELECTRIC HAND PROSTHESES.

4.1. The Russian Myo-electric Control Prosthesis.

In the U.S.S.R., the Central Institute for Prosthesis and Artificial Limb-making, began work in 1957, on new methods of prosthesis control using external (as opposed to internal - i.e. 'body power'), power sources. Since the first experimental work involving electro-myo-graphy in 1925, by Adrian, it has been known that action currents can be picked up using a pair of electrodes connected to a Galvanometer. It was decided for a number of reasons (cf. Chapter 2.4.3.), to control the prosthesis bio-electrically, and a paper describing the first myo-electric hand prosthesis[1] was presented at the First International Federation of Automatic Control in Moscow in 1960.

4.1.1. Electronic Control System.

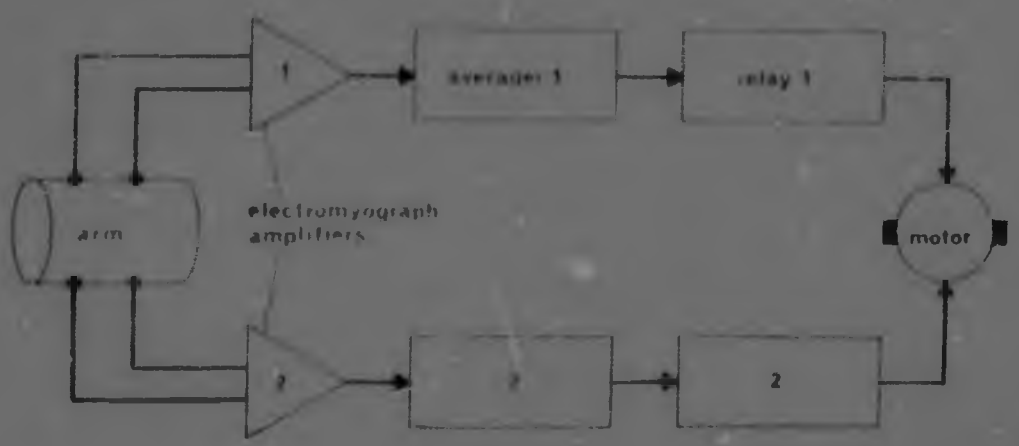


Fig. 4.1. Block diagram of Russian Myo-electric Hand Prosthesis.

Two sets of surface electrodes are attached to the wrist extensors and flexors of a mid-for-arm amputee and these pick up the myo-electric signals set up in the muscles when they are volitionally activated (fig.4.1.). These are fed to amplifiers having a high input impedence and a fixed gain of some thousands. The myo-potentials are then rectified and short-time integrated and when the amplified and averaged myo-potential reaches a preset level one of the two relays - depending on which muscle group, the wrist extensor or flexor, has been activated - closes. The relays connect the motor to the power supply for either clockwise or counterclockwise rotation. The amplifier output is therefore used as an 'on-off' signal to control the motor which drives the hand.

No electronic details such as circuitry or component values have been published but from a visual inspection which the author was permitted to make, the following remarks may be made:-

The myo-electric amplifiers each have a transformer input to provide high common-mode rejection. This is followed by a three stage R.C.-coupled amplifier for voltage amplification. The only mechanical components in the amplifier package are two relays which are employed in the motor control circuitry. These two relays are so connected that they cannot both be pulled in at the same time. It is assumed that two medium power transistors are used for current amplification to drive each of these relays.

The distributors of the Russian hand in the U.K. have made some alterations to the system and improved it to some extent. All the Russian transistors have been replaced by low noise silicon transistors and the relays have been replaced by switching transistors.

4.1.2. Mechanical Description of Hand.

The hand itself is a plastic shell with a reversible D.C. electric motor, gear train, and screw-jack which operate the thumb and fingers, mounted on it.(2) The fingers, which articulate as a single unit, are also made of plastic and are not divided into phalanges thus, movement is only at the metacarpal-phalangial joints. The drive causes the index and middle fingers, and the thumb to move away from, or toward each other, giving a grasp of the three-jaw-chuck type. The hand is covered with a cosmetic glove whcih affords it an extremely life-like appearance.

The published mechanical specifications of the hand are:-

1) Weight of prosthesis = 800 grams

2) " " power-pack = 320 "

3) " " amplifier = 120 "

4) Dimensions of amplifier = 82 x 46 x 21mm.

5) " " power-pack = 100 x 63 x 37mm.

It is possible to hold objects which have a diameter of up to 50 to 60mm., and a weight of up to 1 Kg. The hand and all the joints are strong enough to carry objects of up to 6Kg weight.

4.1.4. The Prosthesis and Stump housing.

The hand prosthesis is mounted onto a forearm made of laminated plastic and is tailored to fit snugly about a plaster mould made of the amputation stump to be fitted. Specially designed rings in the socket allow the hand to be rotated in the long axis of the forearm passively, thereby enabling the hand to be fixed in any desired radial

position. The amplifier, hand motor and electrodes are inter-connected by means of flexible wire, via miniature plugs and sockets.

The electrodes used are made out of stainless steel and generally no electrode paste is used. (A nickel-silver type of electrode was used by the U.S. distributors; but this was found to be toxic when electrode paste was used and when the skin under the electrodes perspired.) The electrodes are mounted in a plastic housing in the socket and are spring-loaded to ensure good electrode-skin contact.

In the earliest models of the Russian prosthesis leather straps were used to keep the socket firmly about the stump. In a new type of fitting the socket is moulded about the olecranon and is closely fitted about the distal end of the humerous. The socket is designed to fit snugly so that it requires some manipulation to put it on and to remove it. Nearly full elbow extension and flexion is possible using this technique, known as the Munster technique, and the patient experiences very little restriction of movement.

The obvious advantage of this method of attaching the prosthesis to the stump is that no uncomfortable harness or straps have to be worn. (In cases where the stump is too short, this technique is not successful.)

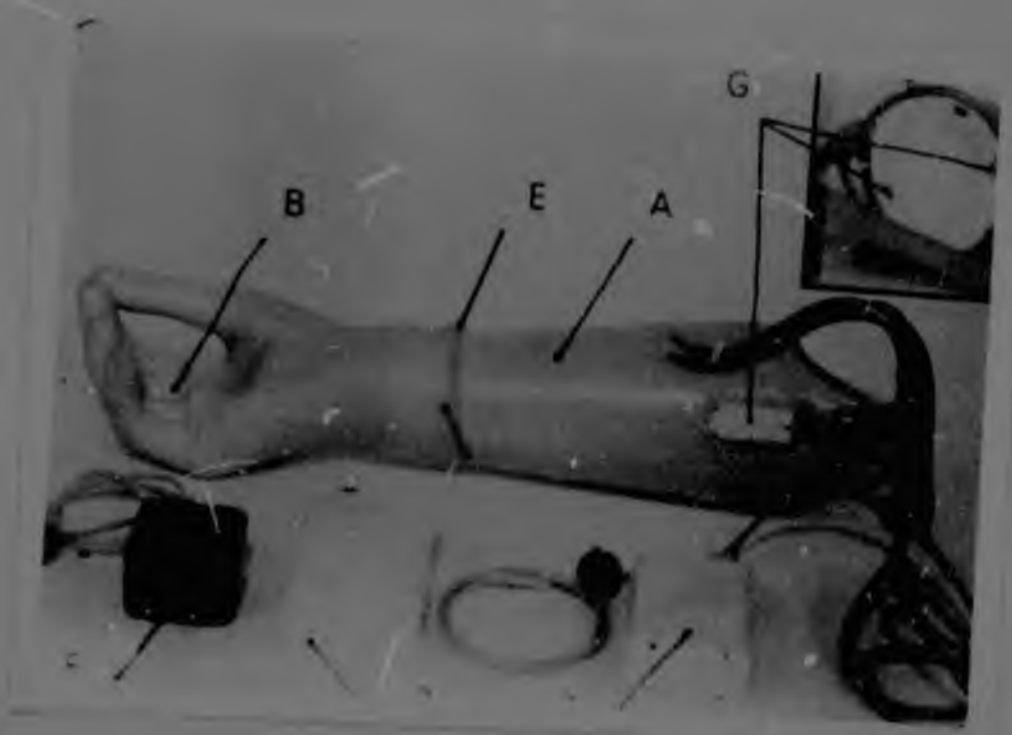


Fig.4.2. The Russian Myo-electric Hand Prosthesis.

4.1.4. Drawbacks of the Russian Myo-electric Prosthesis.

There are a few undesirable features of the electronic control system, which should be mentioned. These are:-

1) The gain of the myo-electric amplifiers is fixed. This causes great difficulty in detecting the correct sites for placing the pick-up electrodes and does not allow for the problem of cross-talk between muscles.

2) There is no overload protection associated with the motor. This gives rise to uneconomic operation as far as battery power is concerned, and the possibility of damaging the motor due to over-heating.

3) The system employed is purely an "on-off" type. There is consequently no control over the speed of operation of the prosthesis (now fixed by the rotor characteristic), nor over the torque output of the motor. The force of the grasp of the hand can thus only be modulated by timing by the patient.

4.1.5. Conclusion.

Some overseas workers have condemned the Russian Myo-electric Hand prosthesis as being extremely 'unenlightened'. The electronic techniques are claimed to be inadequate and the components used unreliable, and suggestions have been made that the electronic system may simply be replaced by two micro-switches which are operated by 'muscle bulge' occurring when the control muscle is contracted.

It is felt by others and it is the author's opinion too, that the Russians are to be congratulated for being the first to actually construct and fit a myo-electrically controlled prosthesis. Shortly after having fitted a small number of amputees, the prosthesis was exported to other countries so that it could be assessed clinically and at home, by as many patients as possible. At no time was it ever claimed that this prosthesis and associated control system, was the ultimate in prosthesis perfection. Hence, the Russian contribution and the impetus it gave to the study of myo-electric prostheses, should not be underestimated.

4.2. The Italian Myo-electric Hand Prosthesis.

The principle of operation of the Italian Myo-electric hand prosthesis[3] is much the same as that of the hand prosthesis developed in the U.S.S.R. Myo-electric potentials are picked up from the extensor and flexor muscle groups in the remaining stump of an amputee, and are operated on, to close the relays controlling the prehension control motor. There are however, a few decided advantages that the Italian hand has over the Russian hand.

4.2.1. Advantages over the Russian Myo-electric Hand Prosthesis.

The advantages of the Italian Myo-electric prosthesis are:-

1) An extremely powerful miniature D.C.motor and associated gearbox power the hand, resulting in a gripping force of approximately 15 Kg. (34.5lbs). This enables the amputee to carry out tasks that require normal hand strength and allows him to handle anything from fragile light objects to heavy ones.

2) Two gain attenuating potentiometers, one for each e.m.g. channel are provided, so that the correct levels for triggering the motor relays can be set up easily. (In the Russian hand, there is no such facility, and extreme difficulty can be experienced in finding the correct sites for situating the pickup electrodes. This was a difficulty encountered when the Russian hand was set up for demonstration in South Africa.)

3) Overloading and stall protection is provided for the motor. This ensures that in the stall condition the motor does not become overheated and that battery power is conserved.

The prosthesis is produced at the Laboratories of I.N.A.I.L. the Italian National Rehabilitation Organisation at Vigorso di Budrio, near Bologna. At present, approximately one hundred of these artificial hand prostheses are in use, and it is hoped to assess the performance and worth of the device from the experience of the wearers.

4.3. The Bottorley Myo-electric Hand Prosthesis.

The West Hendon myo-electric hand prosthesis[4,5,6] developed by Bottorley, Nightingale and Styles, and commonly referred to as the Bottorley Hand, is the most sophisticated device of its kind. A number of significant improvements over the Russian and Italian control systems have been made. These render the Bottorley Hand far superior in its operation to any hand thus far developed.

4.3.1. Proportional Control.

It has long been known that there is direct relationship between the magnitude of the e.m.g. (myo-electric) signal appearing across a particular muscle group and the force exerted by these muscles. (cf. Chapter 3.)

A proportional type control was consequently developed by Bottomley et al, to replace the 'on-off' operation of the previous discussed prostheses. The performence of the hand is directly proportional to the e.m.g. signal amplitude (magnitude) picked up by the electrodes. Thus for example, in order to obtain a stronger grip, the wearer must exert more effort in the remaining wrist extensor and flexor muscles.

The velocity with which the hand opens or closes is also proportional to the e.m.g. signal. Measures of both the force being exerted by the prosthesis, and the speed of operation of the prosthesis, are fed back to the system enabling the wearer to have better control over the state of the system.

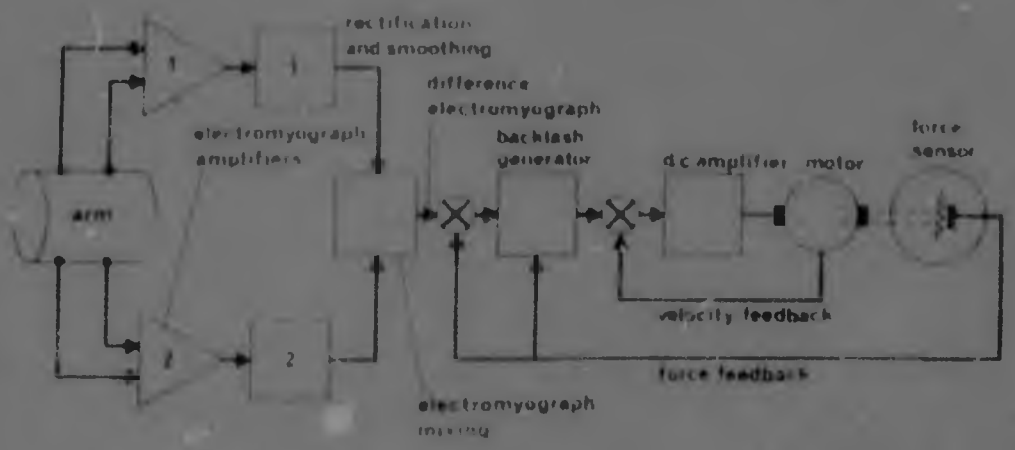


Fig.4.3. Block diagram of the Bottomley Myo-electric Hand Prosthesis.

4.3.2. Force and Velocity Feedback.

Force feedback is provided by means of semi-conductor strain gauges located on the lead screw-shaft which open and close the hand. Slight bending of the shaft is caused by the transmission of the force. This causes a change in electrical resistance of the strain gauges. Two strain gauges are used, one in tension and the other in compression, and form opposite arms of a bridge. The bridge output signal is then fed back and compared to the myo-electric signal amplitude. If the bridge signal is greater than the e.m.g. signal, the prosthesis is commanded to lessen the intensity of its grip. If the bridge signal is less than the e.m.g. signal the prosthesis is commanded to increase the grip intensity. When equilibrium is reached the force of the prosthesis grip is directly related to the e.m.g. signal amplitude. No pressure sensitive devices are located on the hand itself, as it is felt that only direct pressure on the sensitive parts would result in a reliable signal being obtained.

The velocity of the free moving prosthesis is directly proportional to the output from the D.C. differential amplifier. The motor is connected in a bridge type arrangement and as the speed varies, the variation in back-e.m.f. so produced is compared to the e.m.g. signal. The difference between these two signals is then used to increase or decrease the speed of the prosthesis action.

4.3.3. Awareness of Prosthesis Position.

Without actually looking at the prosthesis, the amputee would not normally be aware of its position. To remedy this, a light spring is incorporated in the hand and the strain gauge bridge is adjusted so that its output is zero when the fingers of the prosthesis are in a position similar to that of a normal relaxed hand. Consequently, when the amputee relaxes his grip on an object the hand

will return to a neutral 'relaxed' position, and not remain in the position in which the amputee ceased his volitional control.

4.3.4. Backlash Generator.

Bottomley[6] has found that it is generally not possible for an amputee to maintain a constant level of e.m.g. signal amplitude for even a short period of time. The signal fluctuates a certain amount (c.f. Chapter 3.) resulting in a jerky movement of the prosthesis when an attempt is made to maintain it in a desired state. To eliminate this jerky effect an 'autogenic backlash generator' processes the D.C. differential amplifier signal, before it is passed on to the motor speed control and force control systems.

The 'backlash generator' incorporates a certain 'dead' zone in which fluctuations about any level must exceed a predetermined amount before its output changes. The spurious variations in the e.m.g. signal when an attempt is made to maintain a fixed muscle output level, are therefore eliminated. (Appendix F contains a full description of the operation of the circuit). (Alternatively, this undesirable effect may be overcome by increasing the RC time constant of the rectifying circuit. The drawback of such a measure however, is that the response time of the system as a whole is adversely increased.)

4.3.5. Overload Protection.

Two micro-switches are mounted in the hand to prevent the motor from stalling when the lead-screw has either opened or closed the hand to its maximum limit. The micro-switches also operate when the motor is overloaded in any position of the hand. This feature conserves battery power and prevents damage to the motor due to overheating.

4.3.6. Pick-up Electrodes.

Silver electrodes are used to pick-up the e.m.g. signals. Since the input impedance to the e.m.g. amplifiers is extremely high no electrode paste is needed to reduce electrode-skin resistance. Work is being done however, to develop electrodes that are implanted in the muscle and that will transmit the e.m.g. signal to a receiver in the hand control system. This will result in more accurate information of the muscle activity.
(cf. Chapter 3.4.2.).

4.3.7. Conclusions.

The high-lighting feature of the Dotcomley hand is obviously its proportional control. Finer overall function of the hand can be obtained as well as operation, which is more natural than the purely on/off operation of the Russian and Italian hands.

Although the force feedback system is a most elaborate and ingenious innovation, in fact, this feature is considered to be a possible drawback of the system. In order to maintain a grip of a reasonable force, the amputee must maintain the degree of e.m.g. signal output demanded by the system. After a few minutes the controlling muscle will tire, resulting in a diminished signal amplitude and consequently a relaxation of the prosthesis grip. The amputee must therefore switch off the control system should he wish to maintain his grip for more than say 30 seconds without discomfort.

Because of the complexity of the system the electronic unit is rather large and together with the large power pack required, the amputee has quite a load to carry about on his person. At present, the Royal Air Force at Farnborough, is investigating the possibility of producing the electronic unit in thick/thin film form. Until favourable results from Farnborough are forthcoming, this

prosthesis will remain an experimental project limited to small scale use by a few selected amputees.

4.4. Myo-electrically Controlled Position-Servo System.

Hirsch, Kaiser and Petersen[7] have developed a myo-electric control system which differs from those previously described. They use the myo-potential as the command signal in a position-servo system. Only one muscle group is used as the control source as opposed to the previously described systems, which use a two-channel control.

4.4.1. System Operation.

The myo-electric signals are derived via bi-polar skin electrodes and are fed to a differential input-stage amplifier having an input impedance of $\pm$300 K.ohm. A D.C. analogue signal is obtained which is directly related to the myo-potentials being picked up.

The analogue signal is then led to a bridge circuit which compares this signal to a signal generated as a result of the angular position of a potentiometer. The potentiometer is linked to the prehension mechanism of the terminal device, be it hand or hook type prosthesis (see fig.4. .1.) The bridge output signal is then amplified, and fed to the control circuit which then modifies the prosthesis position in such a manner that in equilibrium, the bridge output voltage is zero. Any change in volitional myo-electric signal output is accompanied by a corresponding change in prosthesis position. When no signal is present, the prosthesis position corresponds to that of maintaining the fingers in the extended position. (Fig.4.4.).



Fig.4.4. Block diagram of Myo-electric Position - Servo Control System.

The static force when the prosthesis is grasping an object is within certain limits proportional to the difference between the actual position of the prosthesis, and the position it would have assumed at that myo-electric signal input, had there been no load.

4.4.2. Motor Drive - An optimum Electro-mechanical System?

An interesting feature of this particular myo-electric prosthesis is its motor system. Two D.C. motors are employed to provide the required prosthesis operation. Motor No.1 is used to provide quick movements, and is incapable of providing much torque. Motor No.2 produces the relatively large torque associated with slow movements, for example, when the prosthesis is gripping or squeezing an object. During purely static conditions i.e. when the prosthesis is maintaining a grip and is fixed about an object, Motor No.2 need not be activated as Motor No.1 keeps the prosthesis locked in position with a minimum of

current. A real relaxation is not possible until there is a change in direction in the motor system, that is, until the control signal is volitionally relaxed.

By using this type of motor arrangement the authors claim that there is cotinuous electro-mechanical function, both during dynamic and static conditions. The cumulative effect of Motor No.1 and Motor No.2 is claimed to be similar to using a series-wound motor and the graph below (fig.4.5.) illustrates the relationship between the velocity and force of the motor pair. It is seen that the force, or torque outrut increases as the speed of the motor pair decreases.

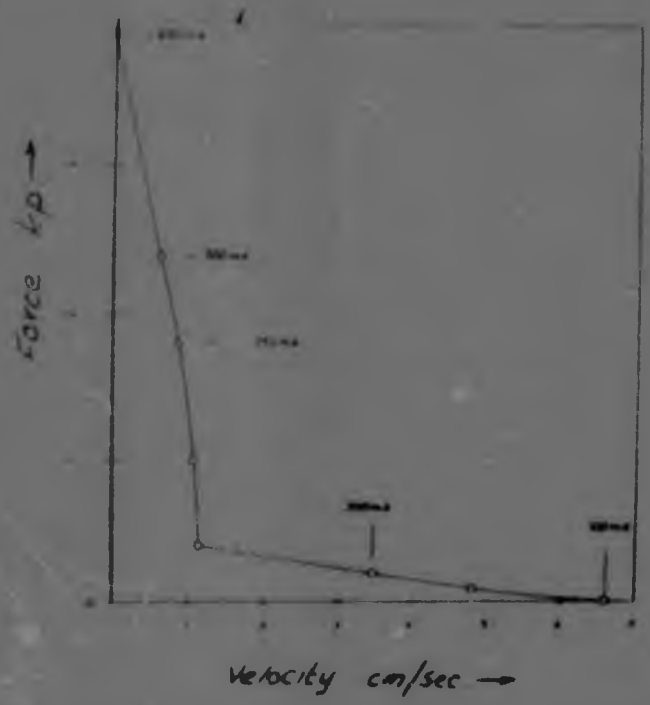


Fig.4.5. Force-Velocity at maximum activation of prosthesis motor system. (After Hirsch, Kaiser and Petersen (7).)

4.4.3. Conclusion.

It should be noted that this system requires a constant expenditure of battery power to the motor in any position of finger flexion, excepting in the fully extended position.

Also, it is necessary for the amputee to put out a continuous myo-potential to maintain any desired position to the prosthesis other than that of the fingers in the fully extended position. Cosmetically, a prosthesis with the fingers fully extended in the normal state is unsatisfactory. This, plus the above mentioned factors are severe drawbacks to the system.

4.5. A Three State Myo-electric System.

4.5.1. Need for the Development of the System.

In the case of the above-elbow amputee, the wrist extensor and flexor muscles are not available as control sites for a hand prosthesis. Control sites other than the above mentioned must be found; and their selection is dependent on the following criteria:-

a) The new control site must have some "psychological" association with the amputated arm. That is, it is undesirable intuitively to use a foot muscle to control an arm - training of the control muscle would be extremely difficult because of the complete dissociation of hand and foot movement.

b) The control must be effected with a minimum of effort and a minimum of unnatural and/or grotesque movement, and,

c) The control site must be used for prosthesis control only, to prevent and minimise inadvertant operation of the prosthesis. The controlling muscle must not be one required for normal remaining functional movement so that its use does not restrict the amputee in any way.

Dorcas and Scott[8] used the remaining Biceps brachii muscle in the stump of a young above-elbow amputee as the control site. This selection fulfills the above requirements because the biceps had no functional role in

body movement aside from that of controlling the prosthesis. Further, retraining of this control muscle would not be difficult because of the past mental connection with arm and therefore hand movement. In this case the biceps were the only muscles available for prosthesis control. The prosthesis used was an electrically powered split-hook type and required two signals for its operation - that of opening the hand and closing it.

This single muscle group i.e. the biceps, was therefore required to provide both signals. A method of achieving this dual control by one muscle is to elicit different levels of volitional myo-electric activity in the control muscle and trigger electronic circuitry to open and close the hand at different preset levels. Three separate states of the control circuitry are readily available, namely:-

1) No volitional activity - corresponding to no activity.

2) Normal muscle activity - a 'prosthesis close' signal.

3) Greater than normal activity - corresponding to a 'prosthesis open' signal.

4.5.2. System Operation.

The volitional myo-electric signal is picked up via surface electrodes to a differential amplifier input stage. Beckman Type 350059 Bio-potential Skin Electrodes were those used. The signal is further amplified using two common-emitter stages for voltage amplification and a common-collector stage is used to obtain a low output impedance, i.e. current amplification. It is then rectified and short time integrated using a diode clamp stage, and resistor and capacitor in parallel, respectively. The output from the rectifier stage is then fed into a modified Schmitt Trigger. Two relays act as the collector loads in the two legs of the Schmitt Trigger. (See accompanying

diagram). When the myo-electric signal input is zero, then both relays are de-energised. When the myo-electric signal strength is 250μV 'peak to peak', enough base current flows into T_1 to switch it on. The current now flowing through T_1 is channelled via T_2 closing Relay 1. A larger signal, approximately 300μV, triggers the Schmidt Trigger and causes Relay 2, to become energised and Relay 1, to become de-energised. (The current now flows through T_3.) The patient may select which relay he wishes to become energised, that is, the channels of his choice, merely by modulating his volitional myo-electric output.

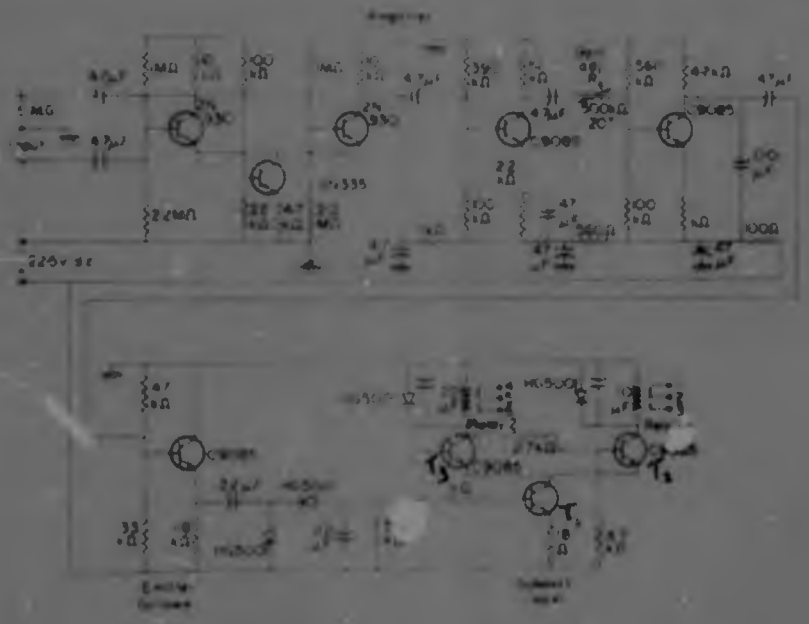


Fig.4.6. Schematic diagram of 3-state Myo-electric Control System (Dorcas and Scott,(1966).

4.5.3. Case Study.

A patient who was fitted with the above system was not a primary amputee, that is, he had the use of a mechanical prosthesis before being fitted with this myo-electric control type. And, although he had good control over his new prosthesis, after a short time he reverted to his mechanical prosthesis. The myo-electric system was not

completely discarded but was then adapted to provide voluntary operation of a wrist rotator mechanism in both directions for the mechanical prosthesis. In the particular case mentioned by Dorcas and Scott, the chief remaining problem was occasional involuntary operation of the control system due to biceps activity accompanying shoulder flexion. The authors expressed the opinion that this problem could be overcome in time by training.

4.5.4. Conclusion.

The addition of the Three State Myo-electric System to achieve wrist rotation or pronation and supernation has resulted in a more functional prosthesis system than could be obtained with manual control alone. This hybrid type control is unique in its concept and it is surprising that other hybrid systems employing both mechanical and myo-electric control sources have not been developed. More details concerning this aspect as related to the work of the author, are presented in Chapter 7.

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CHAPTER 5.

5.1. Design Procedure and Requirements of the Myo-electric System.

5.1.1. Initial Studies.

A preliminary investigation involved the design of a suitable pre-amplifier to monitor myo-electric potentials. A pre-amplifier suggested by Fryer and Deboo (1965)[1] was modified to this end. Suitable power stages were added to this pre-amplifier and attempts were made to use the myo-electric signal as the command signal in a single channel 'on-off' motor control system. At this stage, difficulties were experienced due to 50 c.p.s. mains interference, skin-electrode movement noise, and the electrical interference generated by the motor. These difficulties were overcome however, as described later in this chapter.

Next, a two channel 'on-off' control system was developed. Problems of 'cross-talk' between the myo-electric amplifiers necessitated re-design of the power stages and thereafter a two channel myo-electric system, not very different in its operation from the Italian and Russian systems (Chapter 4), was built. It was not considered necessary to test the system clinically, as the design of the 'on-off' system was viewed as an exercise merely to enable the author to familiarise himself with the problems associated with myo-electric control.

The next stage involved assessment of the operation of the systems described in Chapter 4, and a decision on the operating characteristics of the prosthesis to be developed.

5.1.2. Design Criteria of the Myo-electric System.

It was quite apparent that a simple 'on-off' system

similar to the Russian myo-electric control system, was inadequate. The factor of proportionality between the myo-electric signal output and the muscle tension, (Chapter 3.3.), is valuable control information and should not be neglected in syster design.

This proportionality factor can be made use of to provide single-channel control of a position-servo as described in Chapter 4.4., or it can be used to provide control of a dual channel control syster as in the Bottomley Hand described in detail in Chapter 4.3.

Since the myo-electric prosthesis to be designed was intended for mid-forearm amputees who have wrist extensor and wrist flexor muscles still remaining after hand amputation, (normally the case), it was considered wasteful not to use a dual-channel type of control employing both available wrist muscles.

After due consideration it was decided to attempt the development of a system in which the amputee had control over both the speed of operation of the prosthesis, and the force exerted by it. Both speed and torque output have been made proportional to the controlling muscle tension i.e. to the rectified, and smoothed myo-potential.

The question of velocity feedback and force feedback was then considered. Complex electronic circuitry would be required to accomplish this. Also, the bulk of the electronic pack would become rather large unless special assembly techniques could be used. It was also felt that the extra advantages gained with velocity and force feedback were not important enough to warrant the added complexity in development and construction, and associated bulk and expense.

5.1.3. Summary.

The final requirements of the electronic control system

may be summarized as follows:-

a) The system to be designed must have two separate control channels. The control signals are to be derived from the remaining wrist extensors and flexors.

b) The system is to utilise the proportionality phenomenon between volitional muscle tension and myo-electric signal output.

c) The speed of operation of the prosthesis motor and its torque output are to be related to the muscle tension.

d) The size of the electronic pack is to be kept to a minimum. Attempts are to be made to make the electronic unit small enough to be built into the prosthesis socket itself, and so obviate the necessity for the amputee to carry a separate bulky container for the electronics.

e) Cost is to be kept to a minimum. This is to be considered a major factor in the development. However, the reliability of the system should not suffer because of the use of inferior components; economical design must be used to keep costs down.

5.2. Description of System and its Operation.

The operation of the Myo-electric Hand Prosthesis is described with the aid of the diagrammatic representation of fig.5.1.

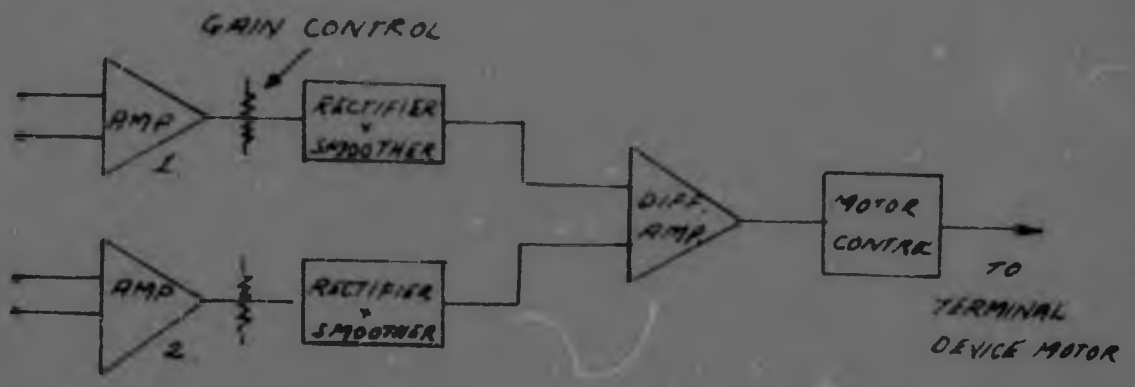


Fig.5.1. Block diagram of the myo-electric Prosthesis System.

The myo-electric potentials generated by volitional contractions in the remaining wrist extensors and flexors of the amputee, are picked up by the myo-electric amplifiers. The amplifiers each have a differential amplifier input stage, are completely transistorised, and conform to the requirements of a myo-electric amplifier as set out in Chapters 3.3. and 3.5. The gains of the amplifiers are continuously variable between 0 and 50,000.

After amplification, rectification and smoothing takes place, a D.C. signal related to the tension in the controlling muscle is available. The signal is then passed to a unity gain differential amplifier. The output from this amplifier provides a signal which is either negative or positive, depending on which muscle group has been activated. The output signal therefore, contains information as to which muscle group, the wrist extensors or flexors are electrically more active, and their degree of activity, i.e. the signal output is

proportional to the difference in tension in the two controlling muscle groups.

The motor control circuitry is proportional in its operation to the differential amplifier output. This allows the patient to have good control over the speed of operation of the prosthesis and over the force exerted by it.

5.3.1. The Myo-electric Amplifier.

The myo-electric amplifier used in the myo-electric hand system consists essentially of three stages - a differential amplifier pre-amplifier and two DC-coupled stages. The differential pre-amplifier stage has emitter follower inputs and a constant current source in the long tail pair of the differential amplifier to ensure good common mode rejection (fig.5.2.). The type of transistors used are 2N3711 silicon planar transistors which have suitable h_{FE} and low enough noise figures at those signal levels monitored from the controlling muscles. It was not considered necessary to use matched transistors or dual transistors for the pre-amplifier. This would have entailed extra expense as far as components are concerned and any imbalance introduced by unmatched transistors, resulting in poorer 50 c.p.s. rejection capabilities, can quite easily be filtered out in later stages. Any D.C. drift due to temperature differences in the differential amplifier transistor pairs have no effect as A.C. signals are of significance only.

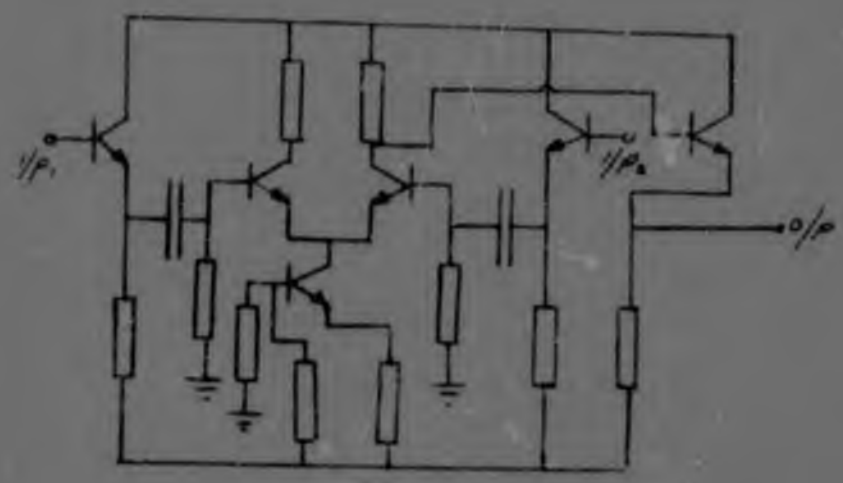


Fig.5.2. Differential Input Stage.

It was found that when the amplifier was required to monitor myo-electric potentials in the vicinity of strong electric fields, such as those generated by mains transformers, the common mode rejection of the amplifier was not good enough, and a simple method of further rejecting 50 c.p.s. interference was employed. The differential amplifier output is fed to a twin-T network designed such that its maximum rejection is at 50 c.p.s. (fig.5.3.). The components are so chosen that a maximum slope rejection characteristic is obtained. (Appendix B).

For optimum operation the twin-T network must be fed from a low impedance source and must be loaded by a high impedance source. To ensure that this is the case, emitter followers are used to provide the required impedance levels. The frequency response of the amplifier with and without the 50 c.p.s. filter, are illustrated in Appendix C.

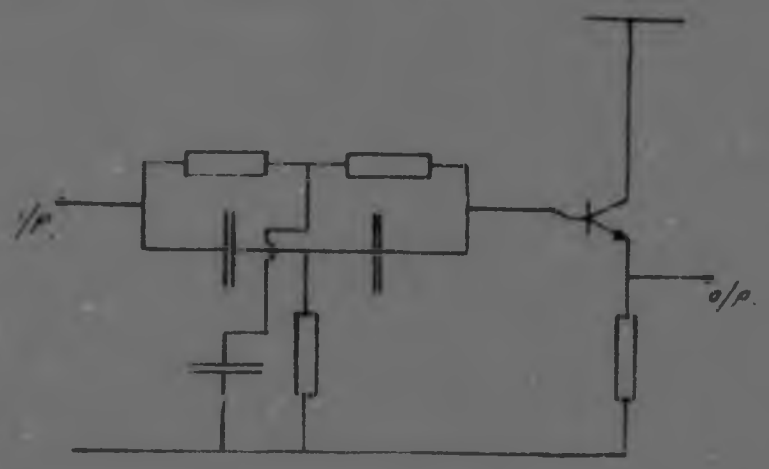


Fig.5.3. Twin-T Rejection Filter.

The output from the twin-T emitter follower is coupled via a 0.1 μf capacitor to the first of the two R-C coupled voltage amplifier stages (fig.5.4.); a high input impedance is thus presented to all low frequency component signals. This limits spurious signals resulting from unintentional electrode-skin movements. It is also known that the maximum signal information content as regards the state of the muscle activity, is contained in the higher frequency components of the e.m.g. signal. (Chapter 3.3.)

To increase stability of the system and cut down its upper frequency response, a 1kpf capacitor is connected in parallel with the feedback resistor. Maximum voltage amplification is obtained in this stage in an attempt to economise on components and space.

The signal is then fed into the second common emitter amplifier stage via an R.C. circuit containing a potentiometer. The gain of the amplifier is continuously variable via this potentiometer from 0 - 50,000. The

final stage amplifies the incoming signal by 25x, and because of the emitter resistor, has a high input impedance and does not load the previous stage appreciably at any setting of the gain potentiometer. The signal is now rectified and smoothed to obtain the D.C. analogue of the muscle activity.

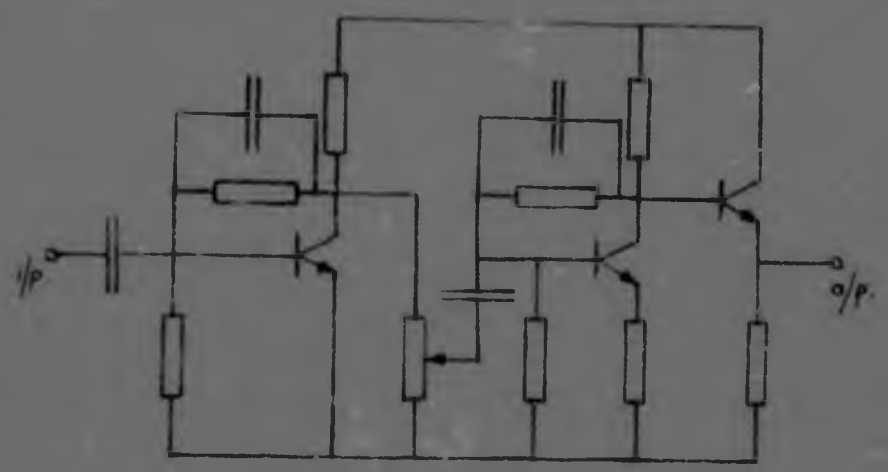


Fig.5.4. Common Emitter Amplifier Stage.

5.3.2. Derivation of the D.C. Analogue Potential of the Electro-myo-graph.

The output from the e.m.g. amplifier consists of an A.C. signal of voltage amplitude of $\pm$ 5.5V maximum. Before passing the signal through a rectifying and smoothing circuit the power output of the final voltage amplifier stage must be increased to avoid loading of this amplifier; an emitter follower stage is used for this purpose.

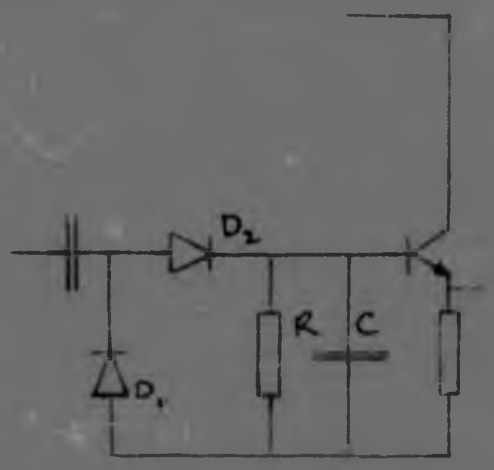


Fig.5.5. Rectifier and Smoother Stage.

The rectifier stage is shown in fig.5.5. The diode D_1, is inserted to ensure a stable zero-signal point. Smoothing, averaging or short-time integrating is obtained by connecting the capacitor C in parallel with R. It is considered unnecessary to full-wave rectify the myo-electric signal as adequate information is available in the half-wave rectified signal (via D_2).

The choice of the time-constant RC, is extremely important.[7] Some workers have found that a choice of component values such that the product RC = 100m.sec. is best. This allows for reasonable smoothing of the wave-form and acceptable response time of the system. A choice of R = 10 K-ohm and C = 10μf (RC = 0.1 secs.) was made. Some patients however considered the prosthesis action somewhat jerky and they had slight difficulty in maintaining a desired prosthesis speed of action. To overcome this factor a 'backlash generator' (Appendix A) as employed in the Bottomley Myo- lectric Hand Prosthesis has been incorporated where required.

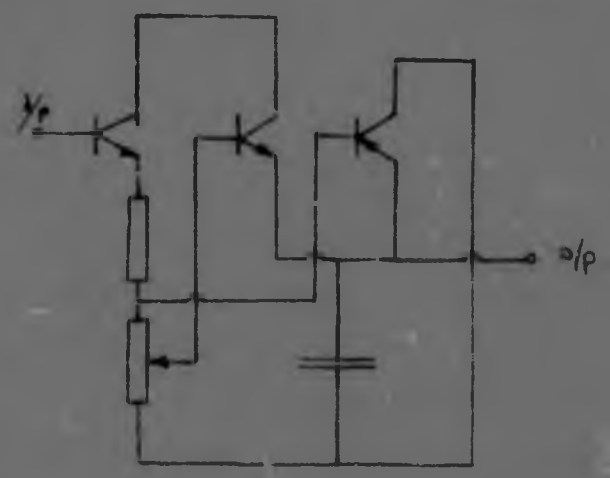


Fig. 5.6. Backlash Generator.

Where the 'backlash generator' circuit (fig.5.6.) is used the time constant (RC) of the smoother may be reduced to about 0.022 secs., (R = 2.2 K-ohms, C = 10 f.) as most of the smoother action would now be derived from the 'backlash generator'. Following the 'backlash generator', a D.C. amplifier provides a potential that varies from - 6 volts at zero muscle activity to + 6 volts at maximum muscle action.

In practice, it has been found that with a little training the amputee can quite easily learn to maintain his myo-electric output at a level which is reasonably constant. Also, the combined inertia of the motor and the prehension mechanism helps to 'smooth out' the results of the irregular current bursts proportional to the jerky averaged myo-potential. The extra advantage gained by including a 'backlash generator' is generally not great enough to warrant its inclusion in the myo-electric system. Only where a patient has abnormally spurious or weak signals due to physiological deficiencies is the 'backlash

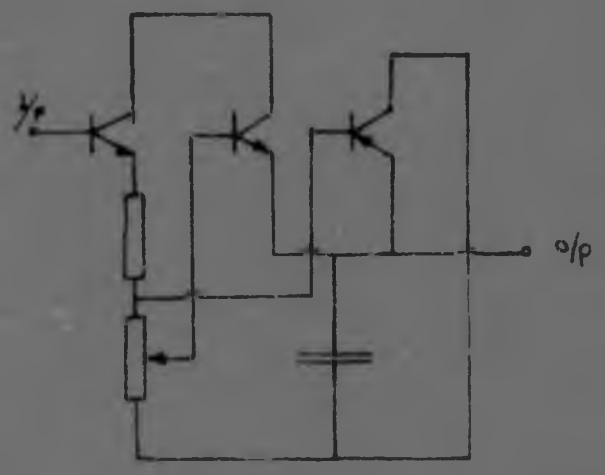


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generator' of significance to system performance. (Circuitry utilising the 'backlash generator' is described in Chapter 6 where myo-electric systems involving quadruplegic patients is discussed.)

5.3.3. Power Supply Separation and Difference Amplifier.

In the early experimental stages it was soon discovered that all endeavours to achieve suppression of the terminal device motor noise, would not enable it to operate from the same power source as the sensitive myo-electric pick-up amplifiers, without causing intolerable interference in the control system. It was therefore decided that the motor should have its own separate power supply.

The use of two power supplies facilitated the design of the motor output stage as follows. As indicated in the figure below (Fig.5.7) the output from the e.m.g. amplifiers are connected respectively to the single-ended input of the motor control circuitry and its earth terminal. When the output connected to the motor control circuitry input goes positive, the motor is directed to turn in one direction and when the other output goes positive, it is as if the first output has dropped negatively and the motor is directed to rotate in the reverse direction. The difference signal between the averaged D.C. signal resulting from wrist extension and flexion is thus simply extracted.

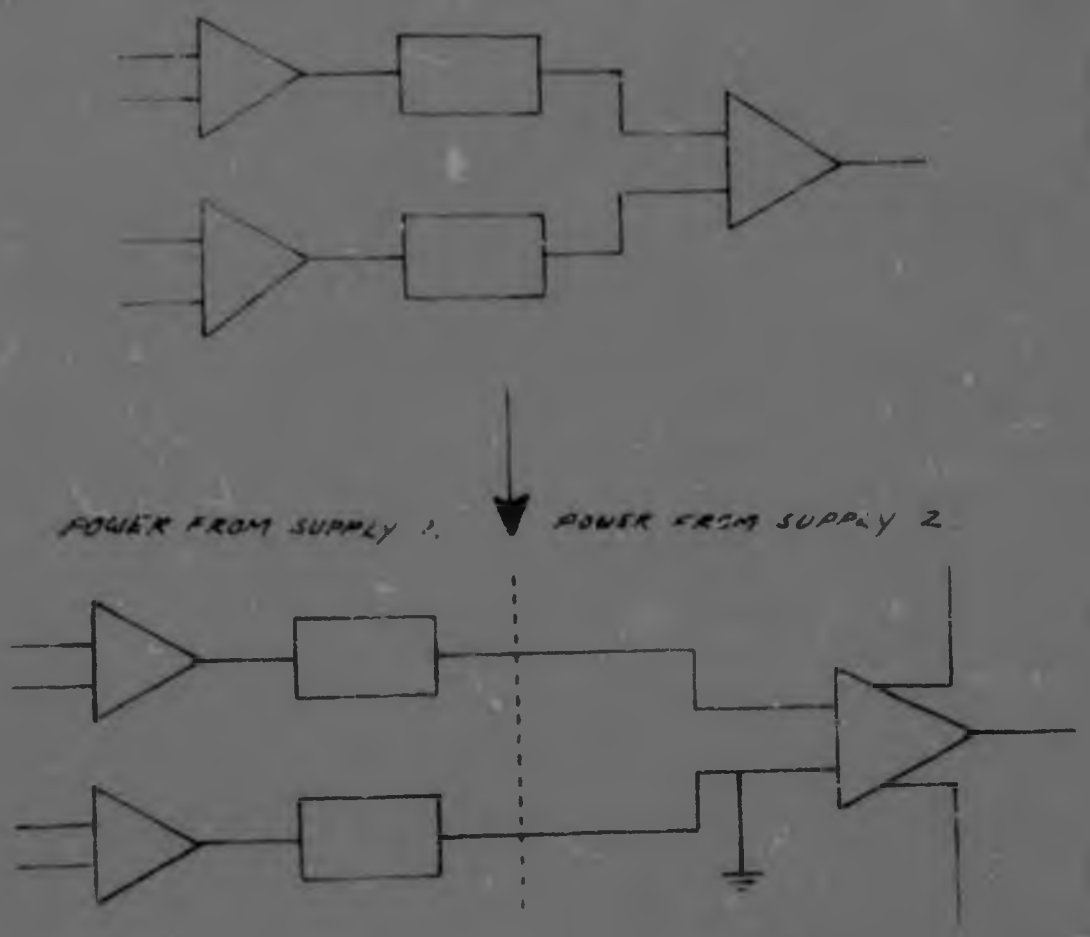


Fig.5.7. D.C. Differential Amplifier System.

5.3.4. Motor Control Circuitry.

Unlike most other myo-electric systems in current use (Chapter 4.1., 4.2.), motor speed and torque output are directly under patient control. This is effected by making the motor circuit performance dependent on the D.C. averaged myo-electric signal amplitude. Ideally, the motor is required to provide negligible output torque and speed control is essential until the patient just touches the object he is about to grip; at this point the speed should drop to virtually zero and torque output of the motor should now be under patient control, thereby providing a variable force with which the object can be gripped. A diagramatic presentation of this requirement is provided in fig.5/8.

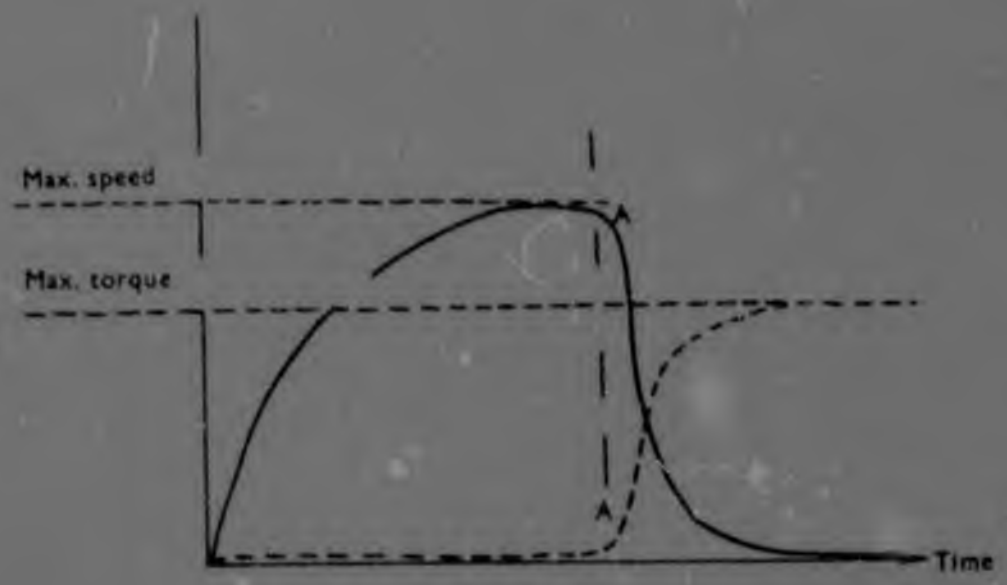


Fig.5.3. Idealised Speed-torque Characteristic.

The type of motor A.E.C. CO ..E used to power the prosthesis prehension mechanism is a D.C. Type motor, having a permanent magnet field. The choice of such a motor is dictated by economic considerations and because speed and torque output in both forward and reverse directions is electronically easy to achieve. There is no necessity to provide pulsing circuits or generate an A.C. power source as would be required if a stepping, or synchronous motor would be used.

The speed of the prosthesis, i.e. the speed of the motor and the torque of the motor, are related to the armature voltage and current as follows:-

$$e = \text{back e.m.f.} = -N\dot{\Phi}/dt$$
$$= V - IR$$

where V = armature voltage,
I = armature current and
R = armature resistance

The torque output is proportional to the armature current.
i.e. torque.e. = K.I.

Clearly then, the drive circuitry must have constant current and voltage control characteristics to meet the idealised requirements of fig.5.8. In practice this is not simply realisable. The circuit employed in the prosthesis control systems does however embody a high output impedance stage and consequently provides a controlled constant current source and thereby good torque control. At low currents that is, at low torque, the output voltage is also proportional to the input signal providing proportional speed control. E.m.g. signal output versus motor speed and motor torque relationships are presented in Appendix D.

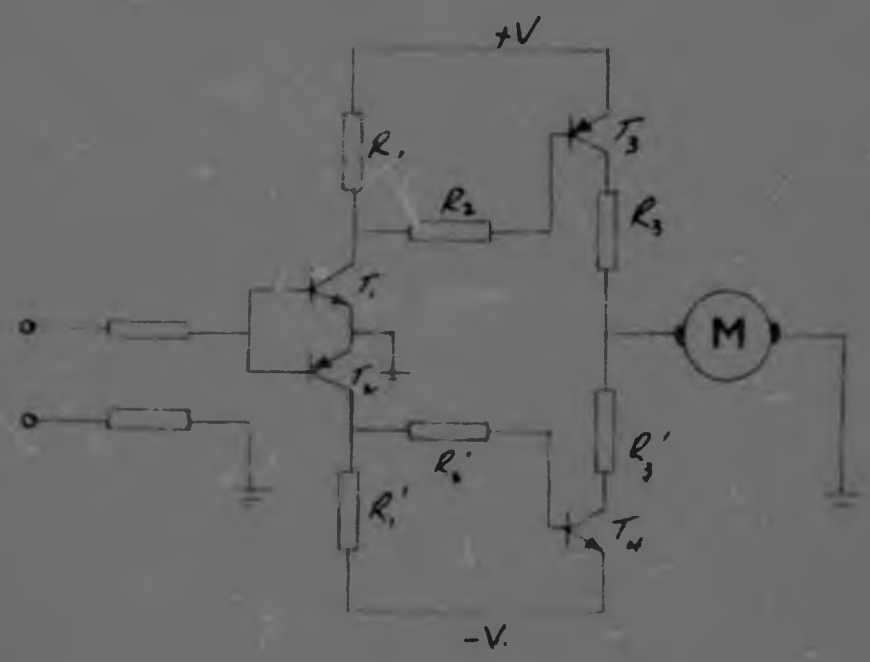


Fig.5.9. Motor Control Circuitry.

The limitations of the circuit in fig.5.9. above are illustrated when the prosthesis wearer wishes to initiate slow opening or closing of the terminal device. A large amount of current flow in the motor control circuit is required to overcome the initial inertia of the motor and its gear-train. The patient must therefore generate an e.m.g. signal consistent with the current requirements of the system. As soon as the motor overcomes the inertia of the system it will not require so much current and the motor will thus tend to

speed up to provide a back e.m.f. to reduce current flow. The patient under these circumstances does not have velocity control. This limitation however, is not as serious as it might appear as the patient soon adapts to the system and learns to compensate adequately for its drawbacks.

Since their outputs are taken from the collector leads, the input and output stages of the motor control circuitry are both high output impedance sources, i.e. constant current sources. Two complementary pairs of transistors are used - both sets are low cost economy plastic encapsulated transistors.

P_1 and P_1, are used merely to stabilise the operating points of T_3 and T_4 thus ensuring that they do not at any stage switch themselves on. The function of P_2 and P_2, is two-fold;

a) to provide base current to the output transistors and also

b) together with P_3 and P_3, to limit the main armature current flow, i.e. they serve as an automatic safety current limiter protecting both output transistor and armature winding.

5.3.5. <u>The Prehension Mechanism</u>.

Initially, an attempt was made to make the prehension mechanism from an Otto Bock standard functional hand prosthesis normally powered by body power (Chapter 2.2.1.) The spring-return mechanism in the hand was removed and a simple nut and screw assembly fitted into the rear of the hand. A number of motors were used in attempts to provide adequate speed and grip for the hand. The only easily available motor that could be used was a Richard 6v motor with accompanying gearbox. Two hands were adapted for electrical power operation.

The adapted prostheses were discarded after testing for the following reasons:-

1) The gearbox was not strong enough and although neoprene gears are used in its construction, these were stripped of their teeth when the patient attempted to grip with some force.

2) The gearbox was provided with a number of different speeds. When the velocity of the prosthesis was adequate, the torque output was not, and vice versa. This indicated that the motor itself was underpowered.

3) The efficiency of the motor was extremely poor.

4) The motor life was low - the brushes burned out after a short time.

5) The size of the motor-cum-gearbox was such that it could not be mounted in the palm of the prosthesis itself and thus had to be mounted externally in the rear of the prosthesis. This would result in the patient receiving an artificial arm that was somewhat longer than his amputated arm. Depending on the length of the patients remaining stump the increase in length could be anything up to 3" longer than normal. While having a longer arm has some decided advantages, it is cosmetically quite unacceptable and inconvenient.

After much consideration it was decided to use a commercially available prosthesis as the terminal device for the myo-electric controlled system. To this end the Otto Bock electrical hand type 8E8. was imported together with a cosmetic glove. This hand looks extremely life-like and is much more psychologically acceptable to patients than the electrical hand adapted from the mechanical prosthesis. The speed and torque output of the prosthesis is good resulting in overall efficient prosthesis operation.

The type of motor used is a permanent magnet D.C. motor type A.F.G. G015. It is electrically much less noisy when compared to other motors used in experimental prostheses and no special noise suppressing circuitry was required. Appendix D shows the voltage versus speed characteristic, and illustrates the torque versus current characteristic of the motor.

The motor is geared down approximately 50x by a train of three gears. The final reduction gear is so fabricated as to form part of the finger and thumb mechanism. A spring loaded mechanism prevents the motor and its associated gears from locking in either of the extreme open or extreme closed positions. The gear-train is so constructed that the fingers can be closed by applying an external force to the fingers and thumb. The reverse is not true and the hand can only be opened by a volitional command by the patient. The spring mechanism described above is utilised to return the hand to a neutral or half-closed position if the hand has been opened to its maximum. The springs chosen are such that the hand is returned to a neutral position only and is not fully closed. (It is envisaged that once suitable machine equipment and technical assistance are available, adaptations of these hands will be produced locally).

The mechanical section of the hand is covered by an outer hand moulded in soft rubber. A cosmetic glove fitting over the outer hand is available in a number of different skin colours to match that of the patient's skin.

5.3.6. Method of attachment of socket plus terminal device to the stump of the amputation.

One of the advantages of the myo-electrically controlled prosthesis is that the need for shoulder harnesses, to open and close the prosthesis, falls away. The same harness is used partly to keep the prosthesis firmly and stably attached to the amputation stump.

The first myo-electric hand prosthesis, i.e. the Russian prosthesis, was provided with a number of leather straps which were used to anchor the socket conveniently above the elbow. In the newer myo-electric prostheses systems however, an entirely new concept in attaching the socket to the stump is used. This is known as the Münster technique, and is now almost universally used.

A plaster cast is made of the stump and elbow joint and a plastic moulding is then made from the cast. The cast is so shaped that it fits snugly about the top of the olecranon and hooks round the bony structure of the humerus.

An extremely stable fitting can thus be obtained and forces which exceed normal hand requirements only will dislodge the socket from the stump. Similar fitting techniques are used to attach the prostheses in the myo-electric system being described.

5.3.7. Choice of Battery Supply.

For reasons explained in Chapter 5.3.3. it was decided to provide separate power supplies for the amplifier circuits and motor driving stages. It would not be economical to use commercially available primary cells for this purpose, and it was therefore decided to use rechargeable cells. The following table compares the various characteristic properties of nickel-cadmium, silver-zinc rechargeable batteries and the primary Léchlanché cell.

	RECHARGABLE NICKEL CADMIUM	RECHARGABLE SILVER ZINC	DRY LECLANCHE
Mass Kg.	.313	.173	.141
Volume cm^3	103	133	80
Energy stored kJ.	22	33	55
Energy density kJ/kg.	70	190	390
Max. usage before replacement or recharge. (days)	1	1	2
Replacements/year.	1	2	183
Cost/battery R.	R18	R14.6	R .70
Cost/year.	R18	R29.2	R128.00

Table 5.1. (c.f. Reference 2.)

Nickel-cadmium batteries have the following advantages over the silver-zinc batteries:-

1) Nickel-cadmium batteries have a longer life than silver-zinc batteries.

2) Extreme caution must be exorcised when recharging silver-zinc batteries. The life of these batteries is substantially shortened by even slight overcharging or over discharging. This is generally not the case with nickel-cadmium cells.

3) Silver-zinc cells are not entirely sealed and if inadvertantly turned upside-down, the electrolyte, a very caustic solution, may spill out, causing damage to the patient as well as to his clothing. Nickel-cadmium cells are completely sealed.

4) As illustrated in table 5.1, the operating cost per annum of nickel-cadmium cells is less than that of silver-zinc.

Although the energy density, energy stored and mass characteristics of the silver-zinc battery are better than the nickel-cadmium cell, nickel-cadmium cells were chosen to power the prostheses and orthoses because of factors 1 - 4 above and because of their availability. Two brands of batteries were used, SAFT and DEAC types. The SAFT cells are less expensive and more easily available than the DEAC, and where the increased weight can easily be accommodated, SAFT cells are used.

5.3.8. Power Requirements of the Myo-electric System.

The current taken by both amplifiers when the hand is at rest is 12mA. When in operation i.e. opening or closing, a maximum of 17 m Amps, depending on the input signal strength, is drawn.

If we assume that the maximum time/day that the hand will be working at maximum output is one full continuous hour, (the worst possible condition) then the total number of mAmp hours required will be:-

(12 - 1) 12 + 17 mAmp-hours.
i.e. 132 + 17 = 149mAmp-hours.

Choosing a battery of 180 mAmp-hours will adequately suit the above requirement. Supply voltages of + and - 6 volts relative to earth were found to be convenient and were therefore chosen.

The current taken by the motor drive stage when no driver signal is present, is zero mAmps. since current is only drawn when the motor is commanded to turn. The amount of current drawn by the motor is of course directly related to the torque output required by the patient.

On no load, that is, when the torque output is used solely to open or close the hand against no external load, the current drain is low, approximately 35 mAmps. Under stall conditions a maximum of 325 mAmps can flow. In practice, it has been found that a battery having a capacity of 100 mAmp-hours is adequate for the control section responsible for opening the hand. Since more power is required when the hand is used to close about an object or to exert a force on an object to grip it firmly than when opening, it is necessary to provide the electronics responsible for controlling hand closing with a battery having a higher current capacity. To this end SAFT cells of 180 mAmp-hour capacity or DEAC 225 mAmp-hour capacity cells are used. In most cases however, the cells used in the power-pack are all 180 or all 225 mAmp-hour capacity for the sake of uniformity and to facilitate battery charging.

The weight of the battery pack is 517 grams for the SAFT cells and 412 grams for the DEAC cells. The batteries are carried in a specially constructed plastic container moulded in such a way that it fits snugly round the waist. An ordinary leather belt is used to keep the battery firmly in place. (Fig.5.10.).



Fig.5.10. Photograph of Battery Case.

One of the patients fitted with a myo-electric hand prosthesis was a young child, and it was felt that the weight of the battery pack was rather excessive for her. Two battery packs containing batteries of 100 mAmp-hour capacity each were made for her. If, during the day, the batteries ran down, the second battery pack could be used. The average weight of these battery packs was 295 grams. DEAC rechargable cells were used in preference to the SAFT batteries because of their lesser weight.

The nickel-cadmium cells were found to be most reliable as far as charge maintenance and current output are concerned. Not enough data is yet available from our patients as to the life of the batteries. From manufacturers data however, approximately one years life can be expected if the batteries are charged correctly, that is, if the charging current is not more than 10 - 15% of the rated mAmp-hour capacity of the cell. (To date only 5% of all batteries used have become defective and it is suspected that these were overcharged at an excessive current.).

Power is supplied to the electronic control pack and the motor via a six-core cable. Two of the wires in the cable are screened and these are used to relay the + 6 and - 6 volt supplies to the myo-electric amplifiers; the outside screen is at zero potential with respect to these two supplies and is regarded as the system earth. Schematically, the batteries are connected as illustrated in diagram 5.11 below.

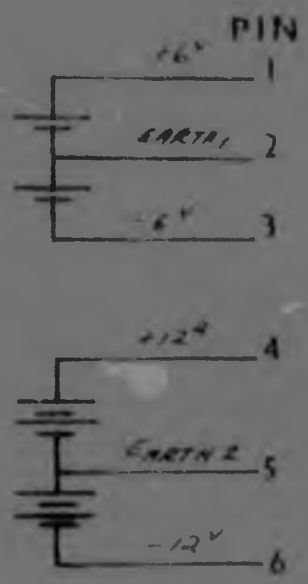


Fig.5.11. Battery Connections.

5.3.9. The Battery Charger.

For optimum recharging of the nickel-cadmium cells used to power both the electronic pack and prehension motor, it is recommended by the manufacturer that the cells be charged from a constant current source. The charging rate is specified at a current of 10 - 15% of the mAmp-hour capacity of the battery under charge. In this particular case it means therefore, that a charging rate in the region of ± 20 mAmps is required. In order to charge the batteries from a state of complete discharge to their maximum capacity, a charging time of 12 - 14 hours is recommended. (At a charging rate of 10% of the mAmp-hour capacity.) Assuming that only 8 hours charging time is available, then a correspondingly higher charging current must be employed to recharge the cells completely.

For the 180mAmp-hour cell, the charging current would then be:

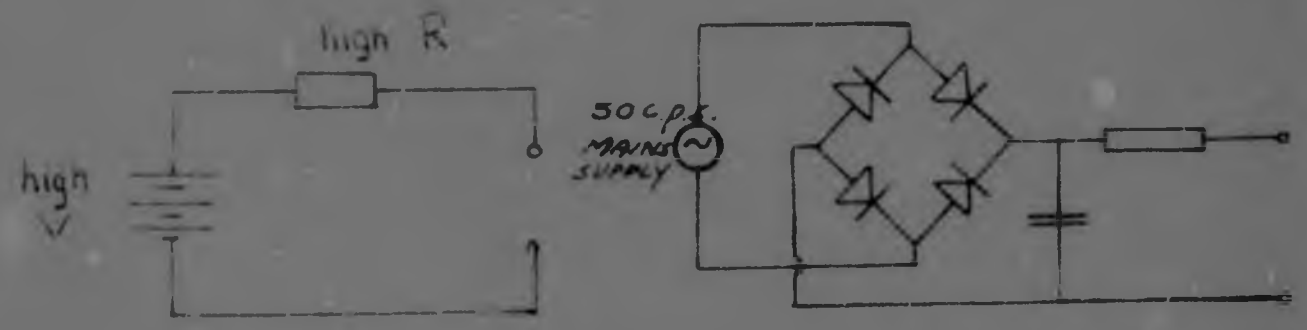
$$\frac{10}{100} \times 180 \times \frac{12}{8} = 27 \text{ m}\Omega\text{mps.}$$

$$= 15\% \text{ of } 180 \text{ m}\Omega\text{mp-hours.}$$

Requirements of both manufacturer and overnight charging time, are therefore fully met.

A constant current supply implies a high output impedance source. The simplest method of obtaining such a constant current source is to provide a high driving voltage in series with a large resistor as shown in fig.5.12. A large battery voltage is required to ensure adequate current flow in the circuit.

The mains supply can readily be used to provide a high enough driver voltage and a circuit of the type illustrated in fig.5.13. may be used as an efficient battery charger.



Figs. 5.12 and 5.13. Simple Battery Charger.

The rectified output voltage Vout is.

$$Vout = 1.4 \times 230^{V}$$
$$= 310v.$$

Charging current = Current through R = 27 mAmps.

The voltage across the batteries to be charged if they are series connected, is made up of 6 cells of 6 volts each giving 36 volts. The volt-drop across the resistor R is then $310 - 36^{V} = 274$volts.

The value of R is then $\frac{Vout - 36}{I\ charging}$

$$= \frac{310 - 36}{27 \times 10^{-3}}$$

$$= \frac{274}{27} \times 10^{3}$$

$$= 10\ K\text{-ohm}.$$

Compared to the effective internal resistance of the batteries while being charged, R = 10 K-ohm. will provide a most adequate constant current source.

The power dissipation in R $= I^2R$

$$= 27^2 \times 10^{-6} \times 10^{4}$$
$$= 27^2 \times 10^{-2}$$
$$= 7.29\ watts.$$

To ensure good heat dissipation rather a large resistor (physically) and heat sink must be used. A reduction in power dissipation is easily accomplished by reducing the driver voltage supply and consequently the value of R. Half-wave rectification of the mains voltage will provide the required lower supply voltage.

The main drawback of the type of battery charger described above is the possibility of the user getting a serious

electrical shock if he inadvertently touches the output terminals when no load is applied to the charger. The total voltage of Vout = 310 Volts, will be present at the output when no current is being drawn through P.

An alternate circuit that may be used is illustrated below in fig.5.

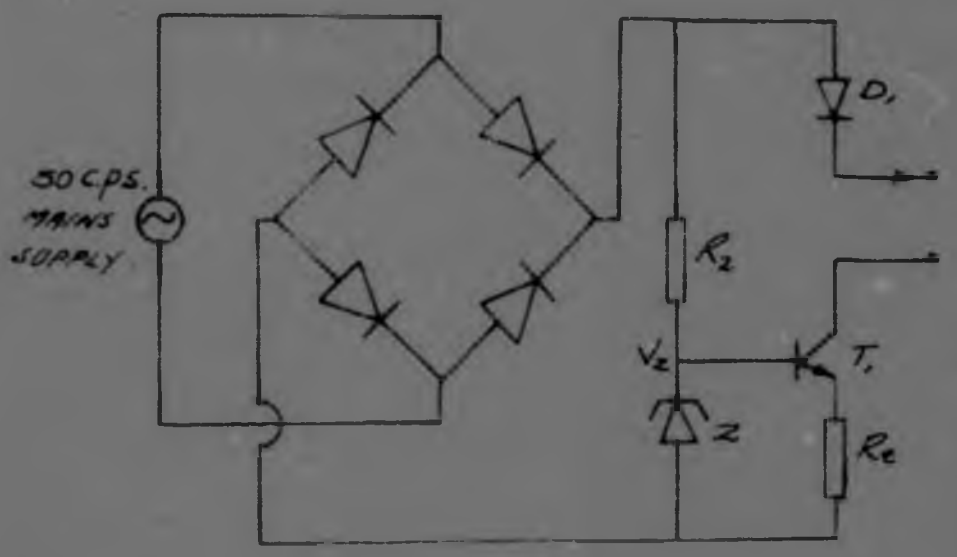


Fig.5.14. Transistorised Battery Charger.

T_1 is used to provide the constant current source, the constant current being given by:

$$I_{cc} = \frac{V_z - v_{be}}{Re}$$

where Vz = the zener diode voltage
and V_{be} = base emitter voltage of T_1

The zener voltage is selected depending on the transformer secondary output and the base current of T_1 for the required value constant current. Silicon transistors were used for T_1, resulting in a V_{be} of 0.6volts.

In fig.5.14 the diode D_1 is inserted to prevent any discharging of the batteries under charge should the supply voltage be switched off.

The transformer secondary output voltage must be high enough to allow for the voltage drop due to the voltage of the batteries being charged (36^V) and to allow for the volt drop across R_e and for V_{cc}. An output of 35A.C. for the secondary was chosen.

The danger of electrical shocks from this circuit is reduced considerably. The cost however is increased over the first circuit. Both circuits have been found to work well and if the proper precautions are taken in selecting the coupling connectors between charger and batteries, it is just a matter of preference as to which charger is used.

The second circuit has been used for the battery chargers being used. The following table indicates what components were chosen for the chargers used to charge battery packs of 100, 180 and 225mAmp-hour capacities.

Cell capacity mAmp-hours	100	180	225
Charging rate mAmps for 8hrs.	15	27	33
V_{Zener}	5.6	5.6	5.6
R_e ohms.	330	180	150
D_1	IN4002	IN4002	IN4002
T_1 3 watt.	MJE 570	MJE 570	MJE570
R_2 ohms.	680	680	680
Zener Type watt.	MZ10007	MZ10007	MZ10007

TABLE 5.2.

5.3.10. Housing of Electronic Pack and Battery.

One of the objects of the systems designed, was to develop an electronic pack that would be small enough to be built into the socket of the prosthesis. To this end, an attempt was made to mount the components on a piece of veroboard, having a 0.1" x 0.1" matrix, which would be small enough to fit into the available space in the socket. The space available in the socket is dependent on the length of the stump of the amputation and the space taken up by the hand prosthesis housing inside the socket. The space required by the hand prosthesis housing is a maximum of $^{3}/_{4}$".

When hand amputations are carried out, the surgeon in most cases, will amputate the hand in such a way that enough space is available for the fitting of a conventional prosthesis. Also, to ensure good vascularisation of the stump, the amputation is normally carried out approximately four inches above the wrist joint in an adult male. Allowing for the $^{3}/_{4}$" required by the prosthesis housing as much as 3" is sometimes available for the electronic pack. This space is more than adequate for housing the electronic system. Fig.5.15. shows the electronic unit mounted on a prosthesis housing. The gain control potentiometers are also housed in the socket itself. This results in increased patient acceptance of the prosthesis as he does not feel over-burdened with additional pieces of hardware. Also, the setting up of the prosthesis is facilitated: the patient need only make one connection, that of plugging in the battery, to be ready to use the arm.

If the stump of the amputation is too long for the electronic unit to be mounted axially in the prosthesis socket, then the electrodes may be mounted on a round piece of veroboard mounted parallel to the stump end. A printed circuit board, specially designed for this purpose

is shown in fig.5.15. If the amputation has been carried out through the wrist joint, it is recommended that the patient undergo further amputation of the stump if he is at all interested in wearing a prosthesis of any kind.



Fig.5.15. Electronic Unit Mounted on Prosthesis Housing.

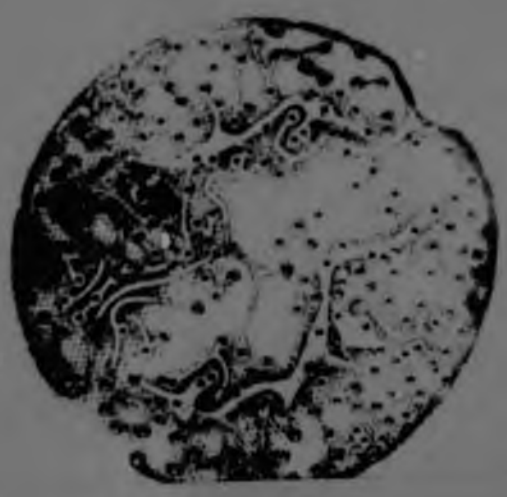
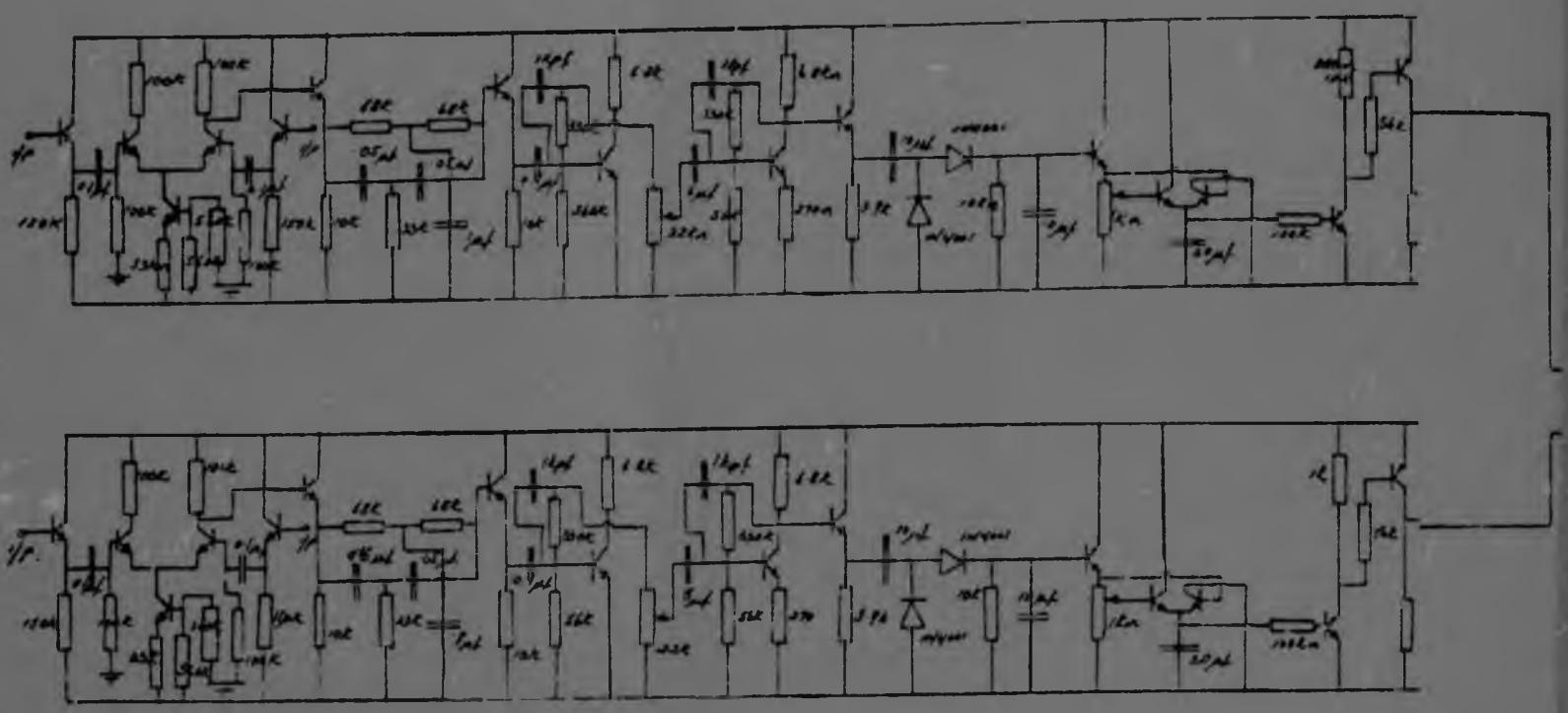
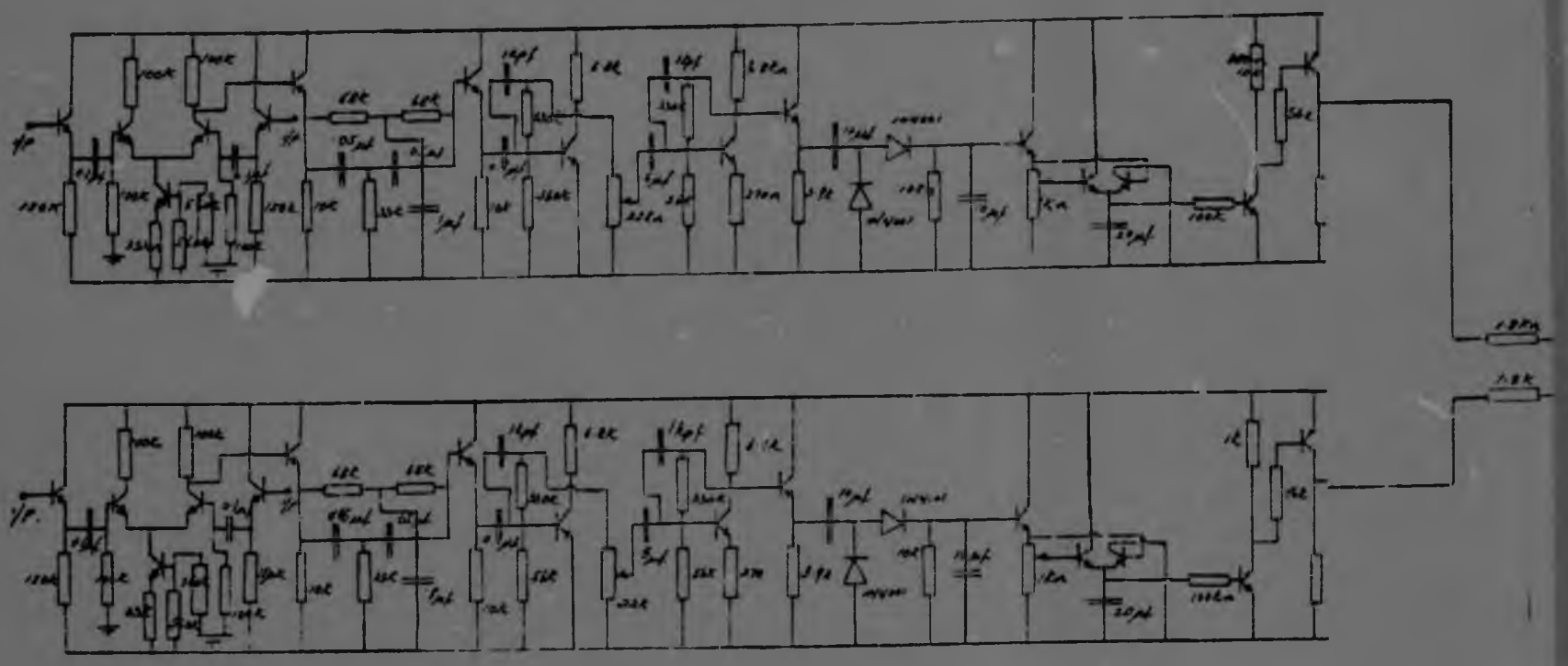
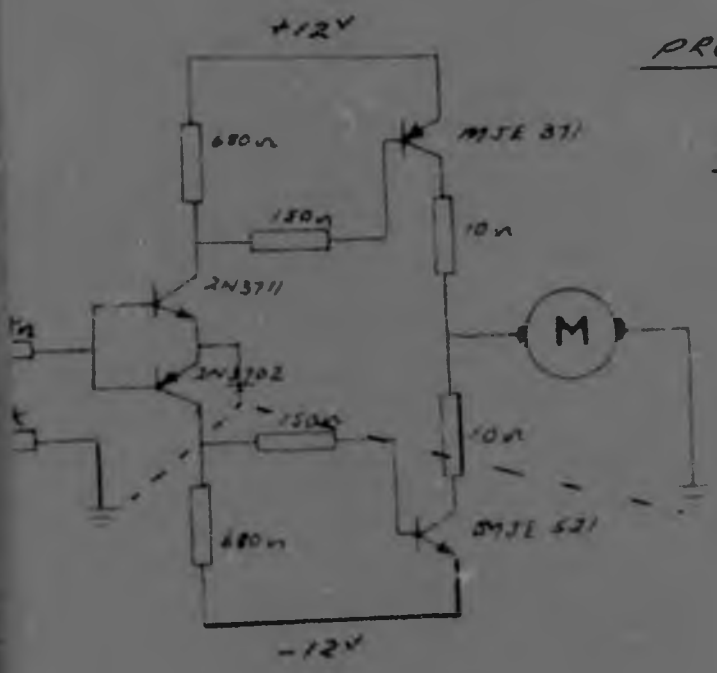


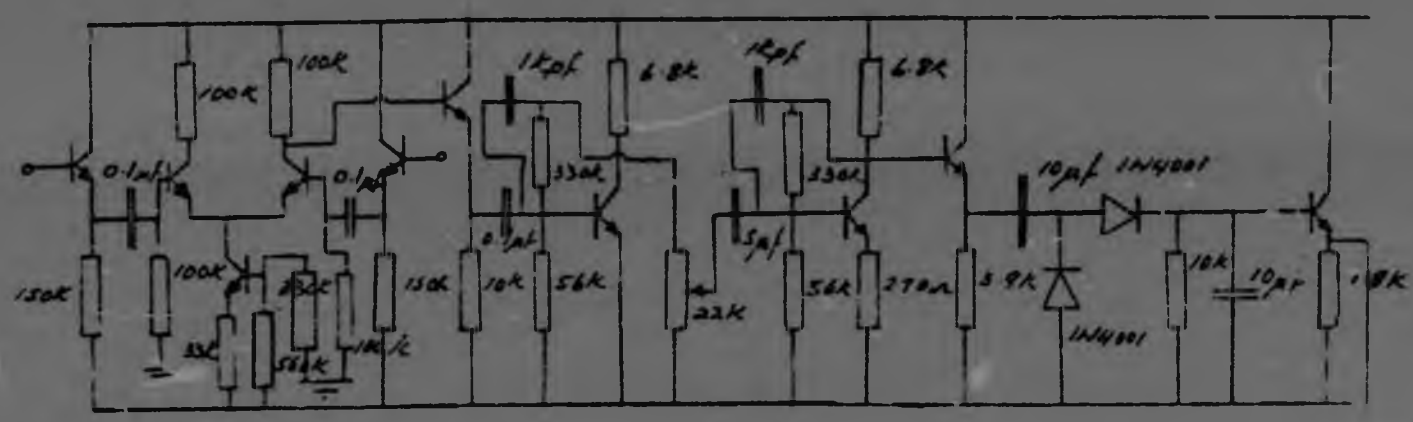
Fig. 5.16. Printed circuit board.

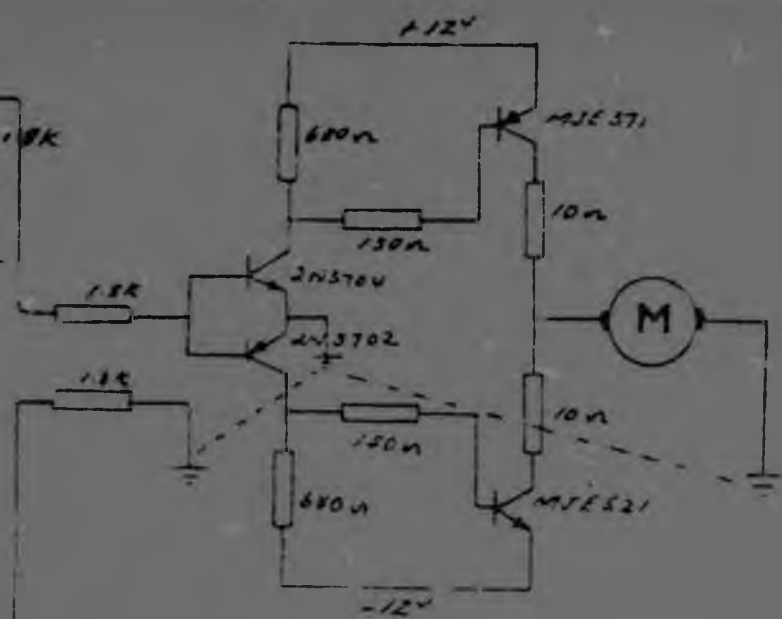


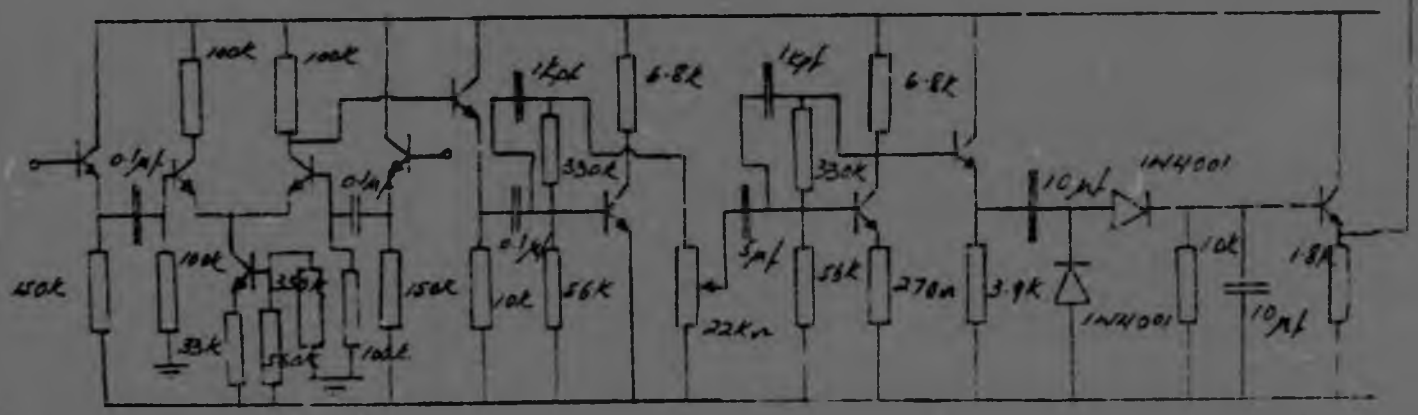


CIRCUIT DIAGRAM OF HAND PROSTHESIS INCLUDING TWIN-T FILTER




100k
100k
1kpf
6.8k
18pf
6.8k
0.1μf
0.1μf
330k
330k
0.1μf
5pf
10μf
1N4001
150k
100k
33k
56k
150k
10k
56k
22k
56k
270n
3.9k
1N4001
10k
10μf
1.8k


+12v
680n
MJE 371
10n
130n
2N5704
1.8k
2N3702
M
1.8k
10n
150n
680n
MJE521
-12v


100k
100k
1kpf
6.8k
1kpf
6.8k
0.1μf
0.1μf
330k
330k
0.1μf
5pf
10μf
1N4001
150k
100k
33k
56k
150k
10k
56k
22kn
56k
270n
3.9k
1N4001
10k
10μf
1.8k

CIRCUIT DIAGRAM OF HAND

PROSTHESIS (EXCLUDING TWIN-T

FILTER

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CHAPTER 6.

Myo-electric Control of Orthotic Devices with Particular Reference to Hand Orthotics for Quadruplegics.

6.1. Introduction.

At the Orthopaedic Section of the H.F. Verwoerd Hospital there are on the average fourteen paraplegic and/or quadruplegic patients admitted per year. In the case of the quadruplegic patients all volitional control of the hands is lost and generally the only arm muscles unaffected by paralysis are the biceps (which are usually strong), brachio radialis and extensor carpi radialis longus. (Spinal lesions at C_5/C_6 segmental level) The latter generally retain about 50% or less of the normal strength. Consequently, patients are unable to do even the menial everyday living activities such as feeding themselves, writing or attending to their toilet.

Attempts to provide myo-electric control of orthotic hand splints for quadruplegic patients were described by Vodovnik et al[1] (fig.6.1.). Myo-electric signals from the left trapezius muscle were used to modulate a faradic type stimulator positioned over the right extensor digitorum muscle. (A faradic stimulator provides stimulating pulses of duration of less than 1 msec.). The fingers of the hand therefore extended in sympathy with the volitional trapezius muscle contraction, and the patient was reported to be able to attain various degrees of hand extension at will. A splint was constructed to stabilize wrist movement, and a spring-loaded mechanism was provided to return the fingers to the normally flexed position when contractions in the controlling muscle ceased.



Fig.6.1. Orthotic Control System of Vodovnik et al.

Preliminary tests, carried out on our patients to ascertain whether a splint of the type described above would be suitable, proved unsuccessful because of a complaint of severe pain. When the arplitude of the stimulating pulses approached a value that provided useful hand-power output, pain became unbearable. The muscle-stimulation method was therefore abandoned and externally powered splints controlled by signals derived from the patient's remaining muscles were accordingly considered.

The work on orthotic devices and the systems developed below are logical extensions of the work carried out in developing and producing a myo-electric hand prosthesis. These orthotic systems have been used at the H.F. Verwoerd

Hospital in Pretoria with varying degrees of success and to the best of the author's knowledge, there is no published literature available on this aspect of the rotorisation of paralysed upper extremities.

6.2. The Need for the Development of Sophisticated Control Systems for Paralysed Upper Extremities.

Hand splints or exo-skeletons are constructed for each quadruplegic and some form of external power, electrical, pneumatic or other is provided to open and close the splint.[2,3,4,5] A pincer grip, (thumb, and index and middle finger), is consequently available to the quadruplegic. Providing control for such a rotorised splint has been attempted using the standard prosthesis control methods.

6.2.1. Powered Orthesis.

In the case of the pneumatic power system, valves incorporated in a shoulder harness are used to supply or exhaust CO_2 gas to a "McKibben muscle[5] (fig.6.2.) or a piston and cylinder system.



Fig.6.2. McKibben Muscle Powered Orthesis.

Alternatively, miniature valves may be located at any convenient control site and may be operated by pushing the shoulder backward or forward or by a shrugging movement. Electric micro-switches may be utilised in a corresponding fashion when electrically powered systems are used.

The above methods of control are usually difficult for the patient who is invariably confined to a wheelchair and must rely, in most cases, on gravity to maintain his posture in the wheelchair. Balance is maintained through control of the remaining muscles in the patient's shoulders and chest. Very often one arm is hooked about the wheelchair-frame to prevent the patient from falling forward. It is clear then, that the standard prosthesis control sites and movements are not at all suitable for orthotic hand splints for quadruplegics.

6.2.2. Choice of Control Muscle - Myo-electric Control?

Since the shoulders have been excluded as a control site, the only other sites are the remaining biceps, brachi radialis and extensor carpi radialis longus. The biceps may be discarded immediately, because the likelihood of involuntary triggering of the control mechanism is greatest since this muscle is employed when lifting objects as a synergist, and when putting them down as an antagonist. Brachi radialis and/or extensor carpi radialis longus, must then be used as the controlling muscles.

Two methods may be used for activating the control system when muscles are used as the control media. They are:-

a) Muscle bulge or change in muscle bulk, - that is, the circumferential change of the muscle as it shortens on being volitionally contracted, and,

b) Volitional myo-potential generation.

The latter method, is the only practical method, since the muscles are considerably weaker than normal muscles, making an accurate and reliable detection of the change in muscle bulk extremely difficult.

6.3. Factors Influencing Syster Design.

A) As mentioned above the control muscles are usually very weak. This means that the myo-electric amplifier syster must be very sensitive indeed.

B) Because of anatomical proximity, it is difficult to separate the myo-potential of the brachii radialus muscle group and extensor carpi radialus using surface electrodes. The myo-potential generated by both these muscle groups will be picked up with any bending of the elbow in certain positions or extending of the wrist. The signals so generated may trigger the control circuitry. A 'dead zone' must therefore be included in the circuitry allowing reasonable elbow flexion and wrist extension without activating the control circuitry.

C) The wrist joint presents a problem in that is is unstable. There are no wrist flexors to stabilise the hand and the patient has to depend solely on gravity to obtain any measure of wrist flexion. This is overcome by stabilising the wrist so that it is fixed externally by metal supports in a functional position. (Although the elbow is also unstable because of the absence of the triceps, this is relatively unimportant here as gravity is used to extend the elbow. The mass of the arm itself provides adequate stability.)

D) Only one control signal exists - that generated by the combined action of the wrist extensor and part elbow flexors, i.e. extensor carpi radialis and brachi radialis respectively. This control site must be capable of relaying three reliable commands to the control circuitry. These are:-

1.) "orientate fingers toward thumb to form pincer grip."
2.) "open hand", that is come out of pincer grip position, and extend fingers, and,
3.) "stop in any intermediate desired position".

These system criterea were considered in the design of the three systems described below.

6.4. Control Systems for Orthotic Hand Splints.

6.4.1. Operation of System 1.

Because of the patients' low work tolerance and general weak condition, it was decided to develope an orthotic control system which would permit control with minimum exertion. The following system, described with the aid of diagrammatic representations of fig.6.3. and 6.4., was designed and built.



Fig.6.3. Block diagram of System 1.

The myo-electric amplifier inputs are connected to the remaining wrist extensor muscles (extensor carpi radialis longus and brachii radialis) in order to monitor the volitional muscle activity when the patient attempts to extend his wrist. In fact, the wrist is not permitted to go into extension as the orthotic splint is so constructed[6] as to stabilise the wrist in a fixed position. When the rectified and smoothed volitional muscle contraction exceeds a predetermined level, the Schmitt Trigger is triggered and the beginning of a pulse is initiated. As soon as the activity ceases the trigger switches the pulse off.

The pulse is then passed through a shaping circuit to obtain an extremely fast rising and fast falling pulse, switching between the voltages zero and +6 volts. The pulse train thus formed by the patient contracting and then releasing his muscle is passed to a counter consisting of two 'T' bistable multivibrators in series. (These form a "divide by 4" circuit.)

The outputs of the 'flip-flops' are then gated to provide an output pulse on two different lines corresponding to counts '1' and '3' of the 'flip-flop' cycle.

The outputs of the decoding gates are then fed to a differential amplifier and its output determines whether the orthotic motor will be stationary, or rotating in such a way as to cause the fingers of the device to open or close.

6.4.2. Details of Electronic Circuitry of System 1.

The myo-electric amplifier, rectifier, and smoother.

This section is exactly analogous to that used in the myo-electric hand prosthesis (see Chapter 5.3.)

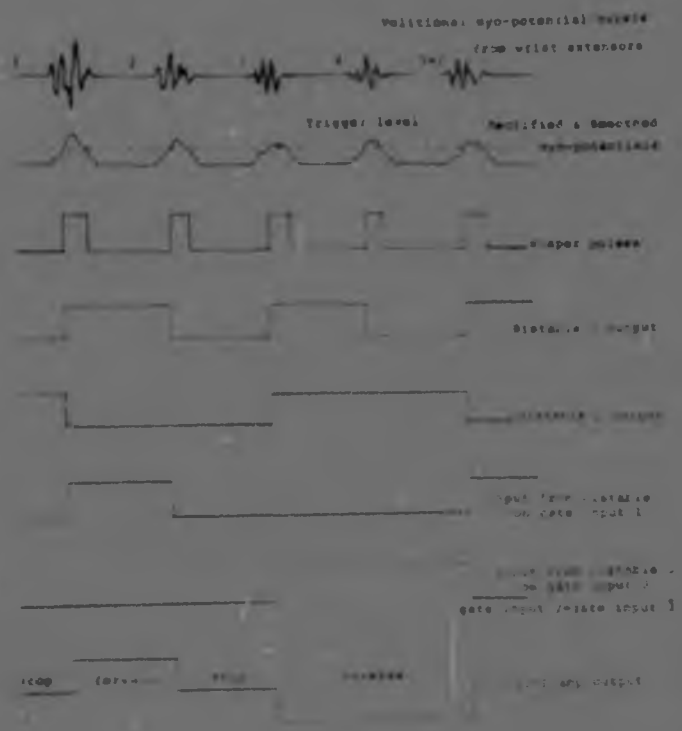


Fig.6.4. Waveforms Associated with System 1.

count number	f.f.1.	f.f.2.	line 1 (forward)	line 2 (reverse)
1	1	0	1	0
2	0	0	0	0
3	1	1	0	1
4	0	1	0	0

Fig. 6.5. 'Flip-Flop' Output Table.

The Schmitt Trigger and Shaper (fig.6.6.)

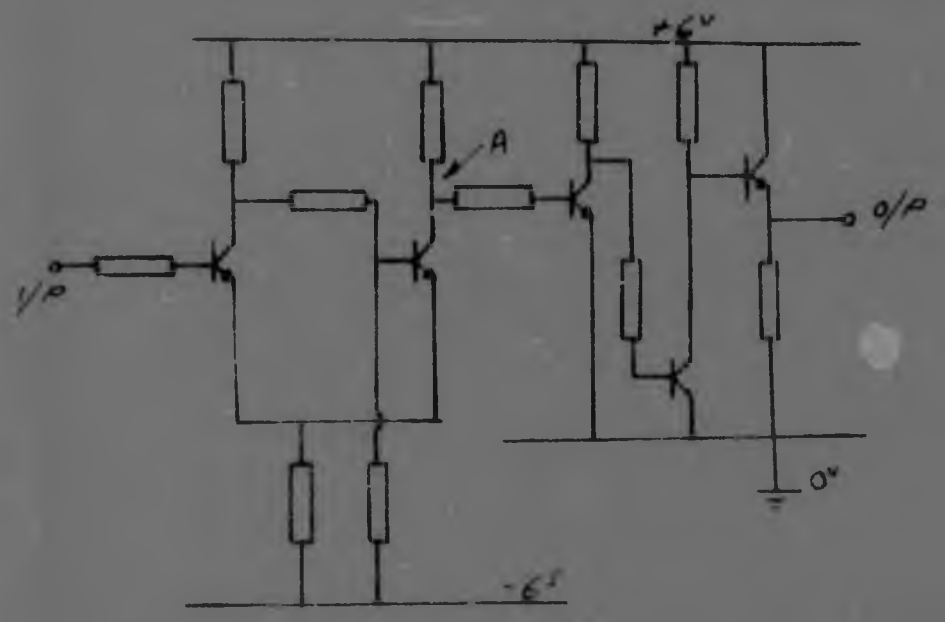


Fig.6.6. Schmitt Trigger and Shaper Circuit.

When the signal from the amplifier is sufficient to trigger the Schmitt Trigger, point A switches from just less than zero volts to +6 volts. This is fed to an inverter circuit whose emitter is tied to the earth (zero volts) rail. This means that when the inverter saturates, the output voltage cannot go below zero volts and when the Schmitt Trigger output is low, the inverter output is +6 volts. The signal is then fed to a second inverter so that the emitter follower output is in phase with that of the Schmitt Trigger.

The inverters are in fact derived from two - 2 input gates of an integrated circuit, Motorola Type MCP 724P. This integrated circuit contains four 2-input gates and the remaining two gates are used in a future stage.

The emitter follower output impedance (fig.6.6.) is low and because of the two preceeding inverters the output pulse has an extremely fast rise time.

Counter circuitry.

The counter circuit consists simply of two integrated circuit J.K. 'flip-flops', Motorola Type MC 790P. The J and K inputs are tied to earth and the clock pulse input is used as the T-input to the 'flip-flops'. For correct operation, these 'flip-flops' require the fast rise time generated by the shaper circuit described above. The diagramatic connections are shown in fig.6.7.

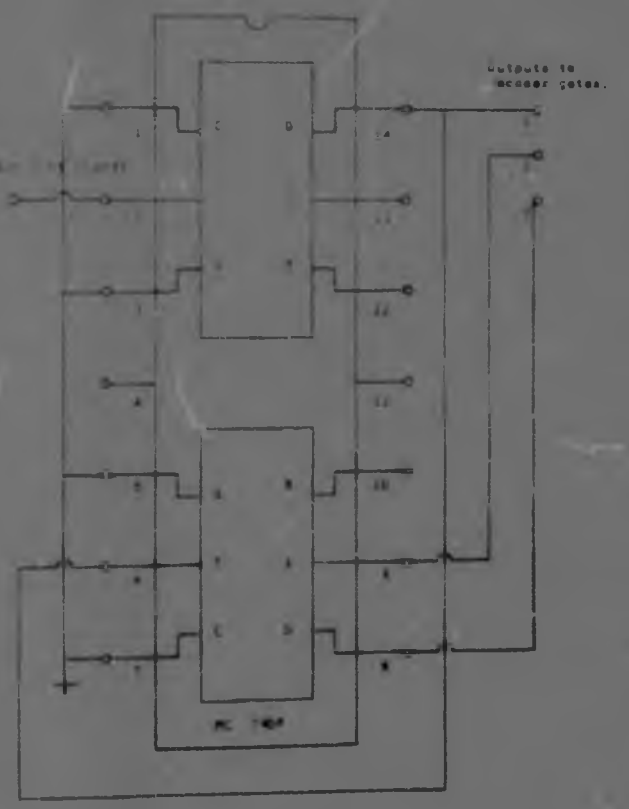


Fig.6.7. Counter Connections.

The outputs of the 'flip-flops' are then connected to the two remaining 2-input gates of the MC 724P integrated circuit as follows. (fig.6.8.)

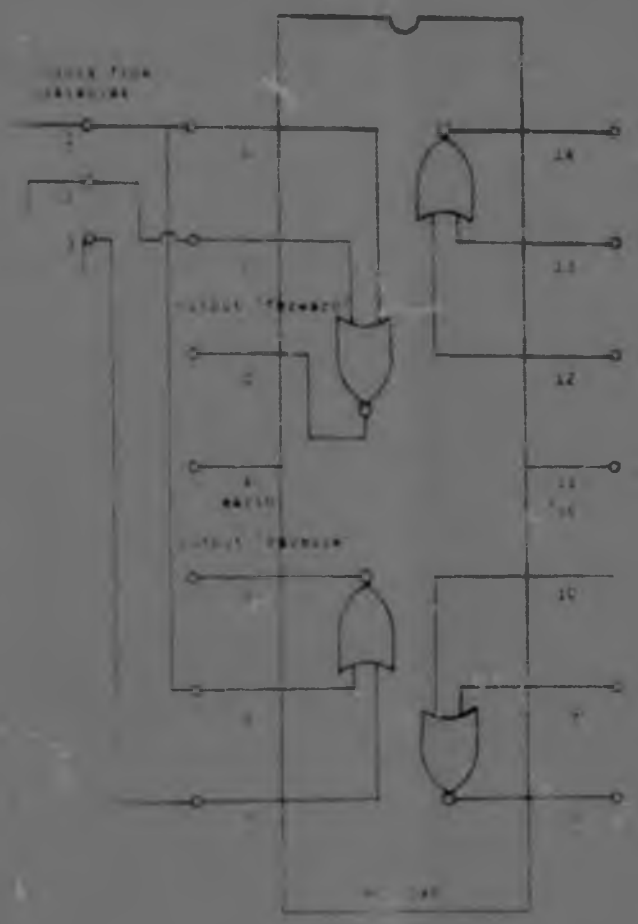


Fig.6.8. Counter Decoder Connections Fig.6.8.

Orthesis Motor Control Circuitry.

As in the section on the motor control circuitry of the prosthetic hand (Chapter 5.3.3.) the differential amplifier to which the outputs of the gates are fed, are respectively, the earth connection of the motor battery supply, and motor control circuitry input. The motor control circuitry is similar to that described in Chapter 5.3.4.

At a later stage it was decided that it was unnecessary to employ the above circuitry as the motor control only operates in an on/off mode. (There is no velocity or force control possible with this system.) Also, two

sets of batteries are required for the above circuit (a positive and negative supply relative to earth). Two relays were therefore substituted for the motor control. These are energised respectively on counts '1' and '3' of the counter cycle. This system requires only one battery which constitutes a saving in this respect. The relays are interlocking to prevent accidental shorting of the motor battery terminals and are connected (as shown in fig.6.9.) such that when relay '1' is pulled in the motor is driven in one direction, and when relay '2' is pulled in, the motor is driven in the reverse direction.

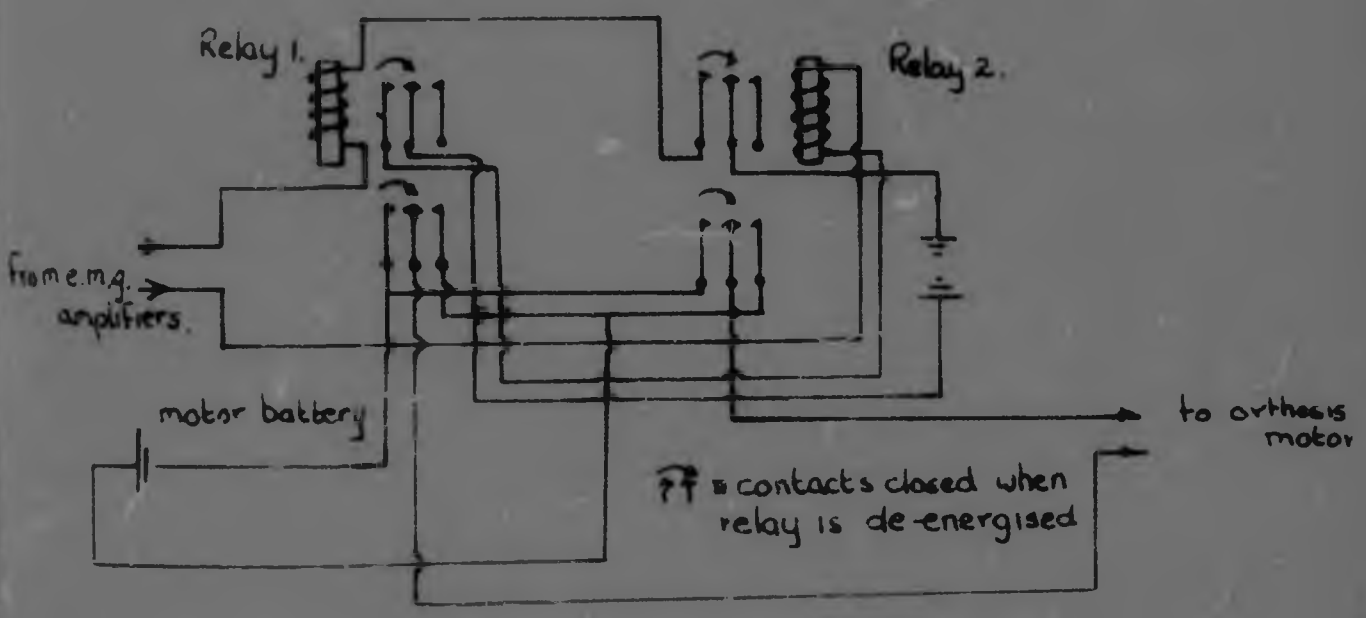


Fig. 6.9. Relay-Motor interconnections.

6.4.3. Operation of the System.

The most serious drawback of this system and the reason for its relegation to a system to be tried only as a last resort, is illustrated with the aid of the following example. Let us assume that the fingers of the orthotic device are closing, and the patient stops the movement just short of the object to be grasped, or that the patient has stopped the movement before he has exerted sufficient grip. Should he now wish to close his hand

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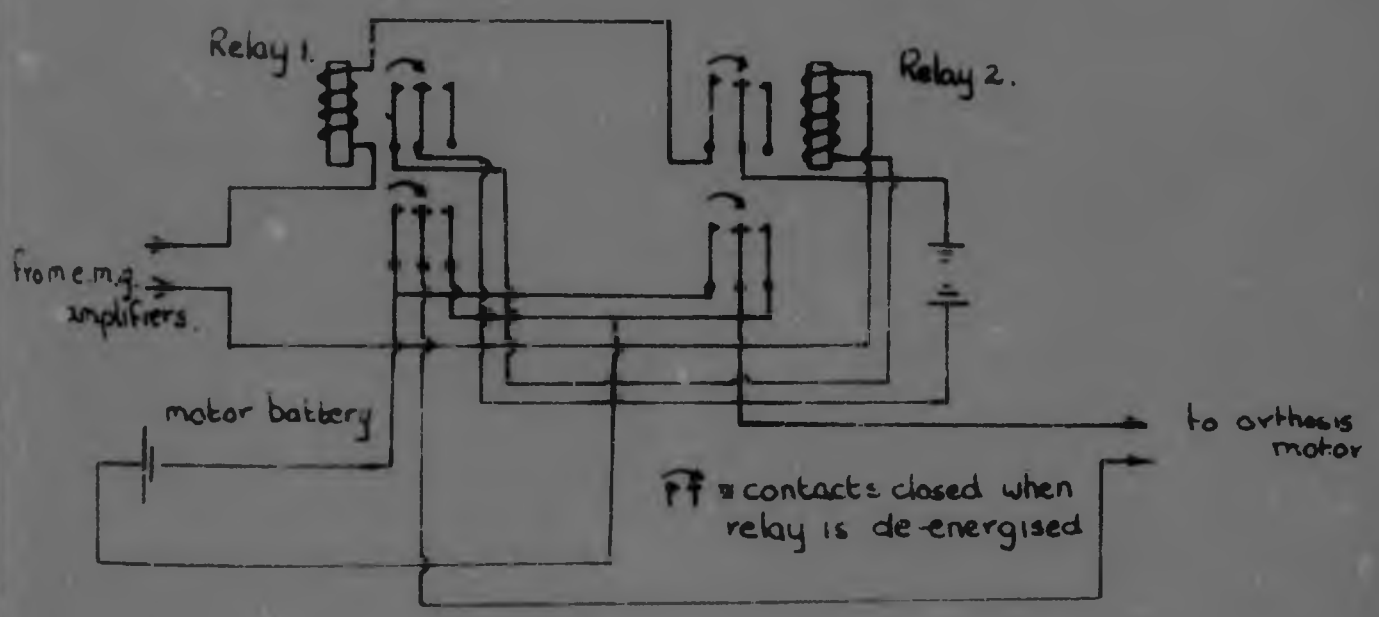


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further, he has to trigger the counter three times, i.e. provide three myo-generated impulses. The first pulse, to get to count '1'; the fingers will then begin to open. Immediately thereafter he must deliver the second pulse to stop the opening movement, and finally he must deliver the third pulse - to start the hand closing again. Besides being wasteful as far as time is concerned the delay in selecting a required movement is foreign to our normal movement selection patterns, and the sequential movement selection is thus difficult for a patient to learn.

Also, because of the extreme sensitivity of the e.m.g. pickup amplifiers the patient might inadvertently trigger the electronics and initiate an undesired orthesis movement. This results in eventual frustration for the patient and he is then in a continual state of tension and anxiety while using the apparatus. (An on/off switch within easy reach of the patient has been provided so that he can cut out the motor battery circuit at will.)

6.4.4. Housing of Electronic Pack and Batteries.

Since the patient is confined to a wheelchair, we are not limited by the size or weight of the batteries powering the amplifier, associated electronics and motor.

The batteries, etc. can conveniently be mounted on some section of the wheelchair. Three 6 volt motor-cycle batteries are used each having a capacity of 11 a.m.r. hours, to power the electronic control and to provide power for the orthesis motor.

It is recommended that the patient have the batteries charged about once a month, and a commercially available battery charger is issued to the patient for this purpose. Photograph 6.10. shows a typical housing built to enclose all the required electronics and necessary battery power and illustrates how mounting is effected on the rear of the wheelchair.

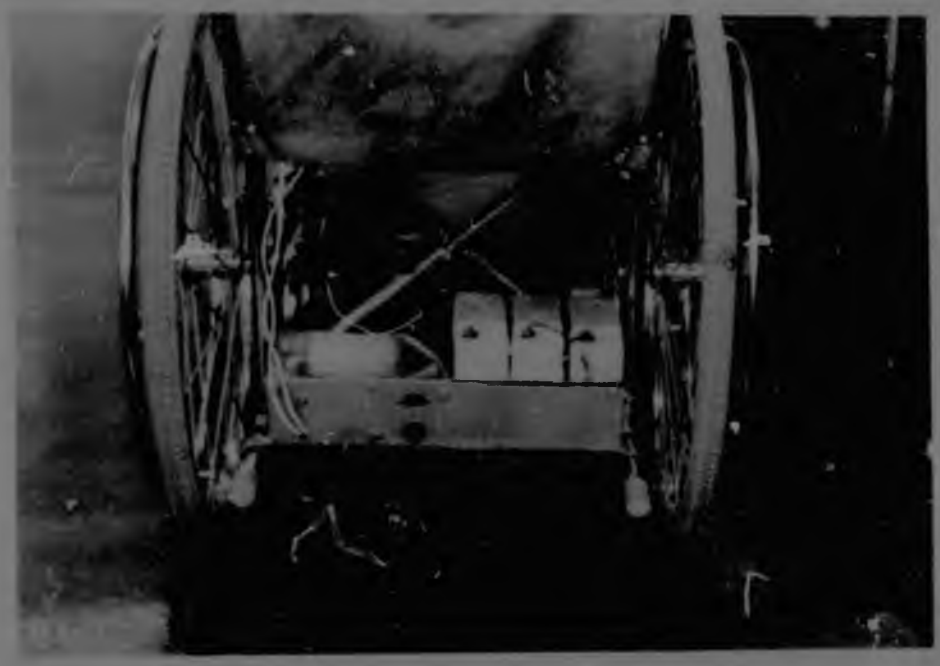


Fig.6.10. Mounting of Electronic Pack on Rear of Wheelchair.

6.4.5. Operation of System 2.

Electrodes are situated over the remaining wrist extensor muscles and brachii radialis. The signals appearing at the electrodes when these muscles are volitionally contracted are fed to a myo-electric amplifier, and after amplification, are rectified and smoothed to obtain a D.C. analogue voltage of the muscle activity. The output signal is then processed according to the system illustrated in block diagram form in fig.6.11.

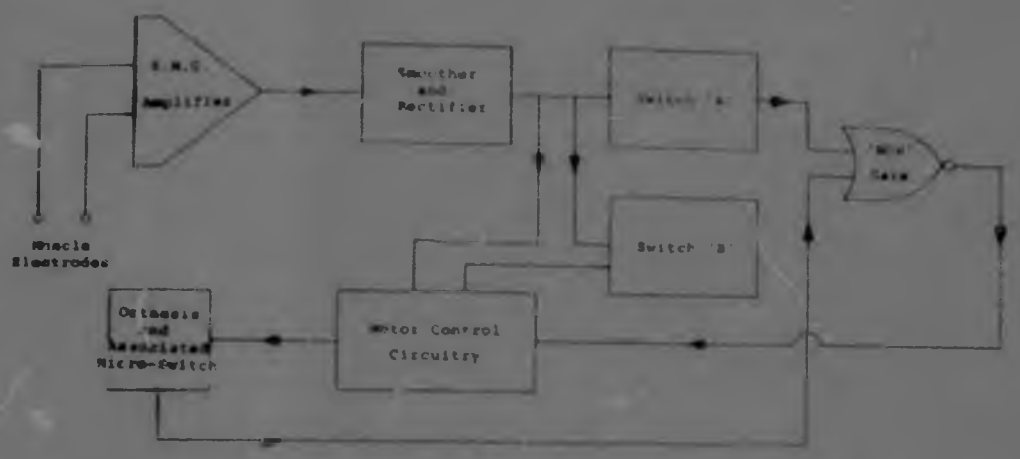


Fig.6.11. Block diagram of system 2.

When the myo-electric activity reaches a predetermined level the switch 'A' is activated. The output from the switch 'A' is fed to a dual input 'nor' gate. The other input to the 'nor' gate is derived from a micro-switch which is mounted on the hand splint. The micro-switch is so situated that it is triggered only when the orthesis is opened to its maximum limit. When neither of the two inputs are present, the orthesis motor is commanded to open, that is, $\bar{A}$ or micro switch $\equiv$ 'orthesis open'.

When either of the 'nor' inputs are present, the circuitry interprets the gate output signal as an instruction to maintain the motor in a stationary position. Should the volitional muscle activity exceed a second preset level greater than the D.C. level required to trigger switch 'A' a second switch 'B' is triggered. The orthesis motor will then be commanded to rotate in such a direction as to close the splint. The performance of the motor is directly related to the volitional myo-potential output above the D.C. level required to trigger switch 'B'. Consequently,

the patient has control over the speed and force of the orthesis in the closing direction, in much the same way as the amputee has control over the closing action of the myo-electric hand prosthesis described in Chapter 5.

When the patient requires the position of his splint and therefore his hand, to remain in a certain desired fixed position, he merely reduces the level of the myo-electric activity in the controlling muscles to that required to trigger switch 'A' only, and not 'B', and thereby provide an input to the 'nor' gate. As explained above, this results in the motor stopping.

When the patient requires to open his hand or release his grasp about an object, he relaxes the controlling muscle completely. The switch 'A' is now turned off and neither of the two inputs are present at the 'nor' gate. The output of the 'nor' gate now commands the control circuitry to open the hand and the splint opens to its maximum limit. At this point the micro-switch is triggered, providing one input to the 'nor' gate. The reversing action of the motor is now inhibited, and the orthesis is inactive until a volitional signal is received from the control muscle.

6.4.6. Mechanical Construction of the Hand Splint of System 2.

The mechanical construction of the splint is similar to that described in the previous chapter, with the exception of only one feature, - namely, the inclusion of a micro-switch on the splint frame. (As illustrated in photograph 6.12.) The micro-switch is triggered when the orthesis is nearly fully open. Provision has been made to allow the motor to slow down after the micro-switch has been triggered, by triggering the micro-switch a few shaft revolutions before the orthesis is completely open.

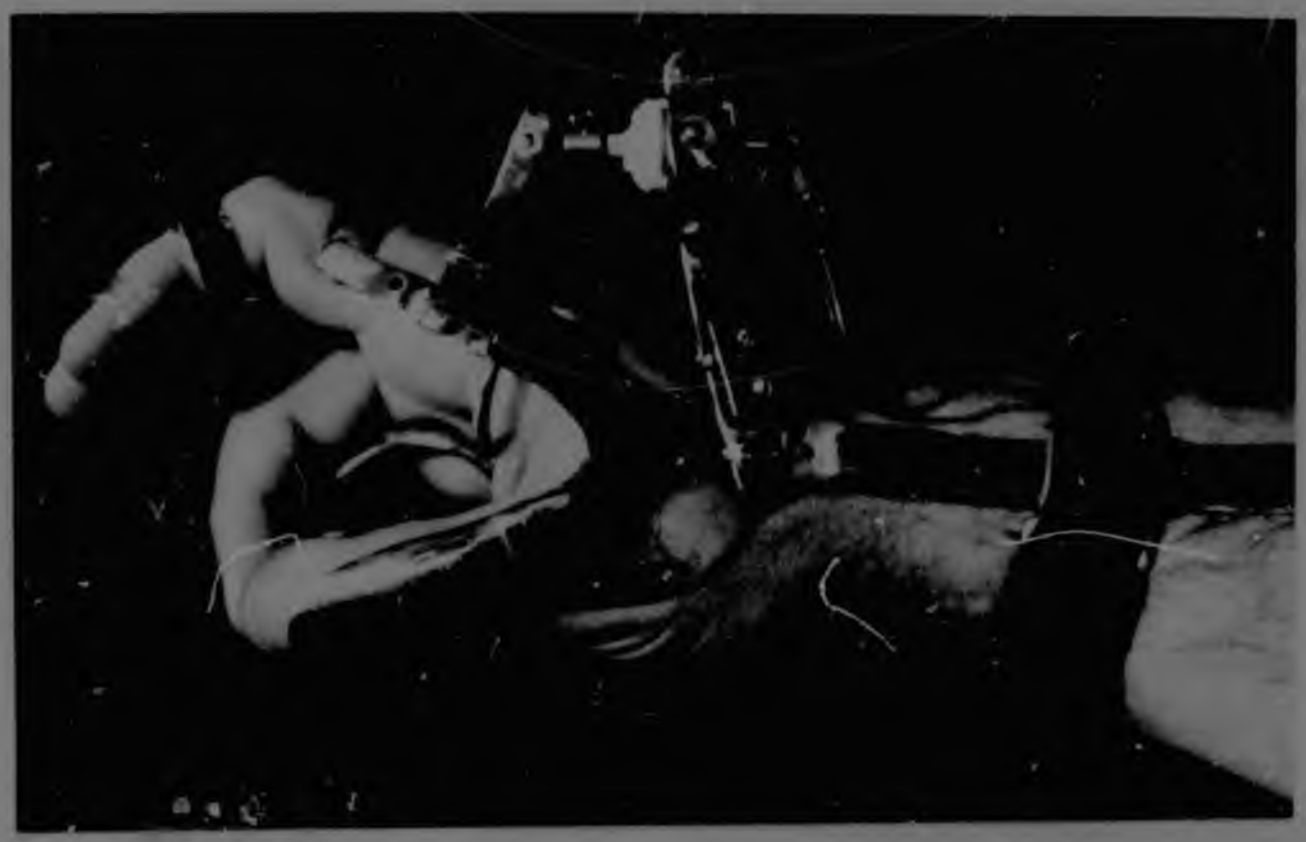


Fig.6.12. Mechanical Situation of Micro-switch on Orthesis (System 2).

6.4.7. Difficulties Encountered with System 2.

The system has been designed for patients who generally have reduced activity in their control muscles and have a fatigue thresh-hold substantially lower than normal. The control muscle must be able to maintain a constant level of activity when the hand is being kept in a fixed position, and must provide an output signal greater than this level when closing the hand. Too much is really required from the muscle. It does not take more than a few minutes of splint use for the control muscle to become fatigued and therefore for the system to become unreliable and frustrating for the patient to use. In practice, at best a jerky, unsure control results. (For the foregoing reasons a position-servo control system cannot be considered as an optimum control system for hand splints for quadriplegics.)

6.4.8. Operation of System 3.

The "3 state control" system first described by the Canadian workers, Dorcas and Scott[7] for hand prosthesis control, is the basis for the operation of this system. (c.f. Chapter 4.5.). They used the biceps for control of a prosthesis in the following way. When the control muscle was relaxed the prosthesis remained inactive. At a certain predetermined level of myo-electric activity, the prosthesis opened. At a second predetermined level of myo-electric activity, the prosthesis closed. Their system has been modified to suit the requirements of a myo-electric hand splint control. Such splints have been used with success by a number of our patients.

The muscles available, i.e. extensor carpi radialis and/or brachii radialis provided the controlling signal. The electrodes are situated over these muscles and the signals monitored are fed to the differential-input first stage of the myo-electric amplifier. Smoothing and rectifying then follows and a D.C. signal proportional to the muscle activity is obtained.

The output signal is graded into three ranges which may or may not be equal depending on the quality of the patients control signal. (See fig.6.13.) If the myo-electric activity is such that the output signal falls into the range C-B, the orthesis motor remains stationary. When the controlling muscle signal is such that the D.C. signal output falls in the range B-A, the orthesis motor rotates in such a manner as to close the hand. When the output signal falls in the range B-C, the orthesis motor rotates in the opposite direction opening the hand.

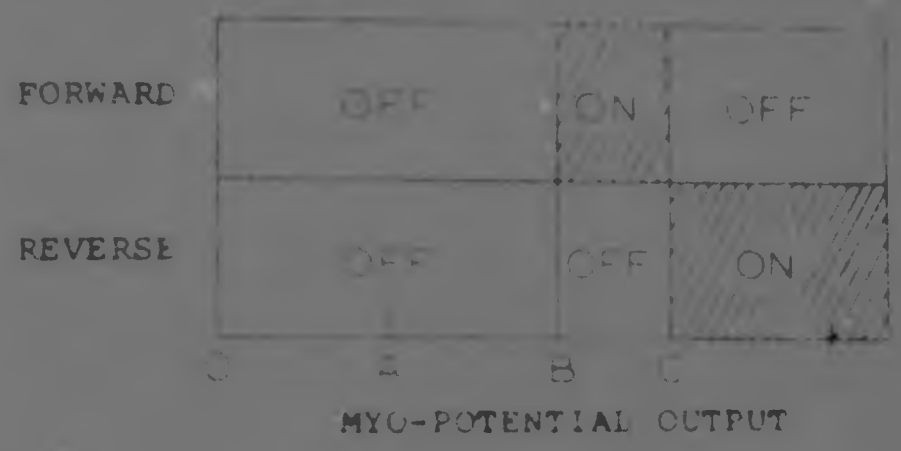


Fig.6.13. Table representing operation of 3 State System (after Dorcas and Scott).

6.4.9. Details of Electronic Circuitry.

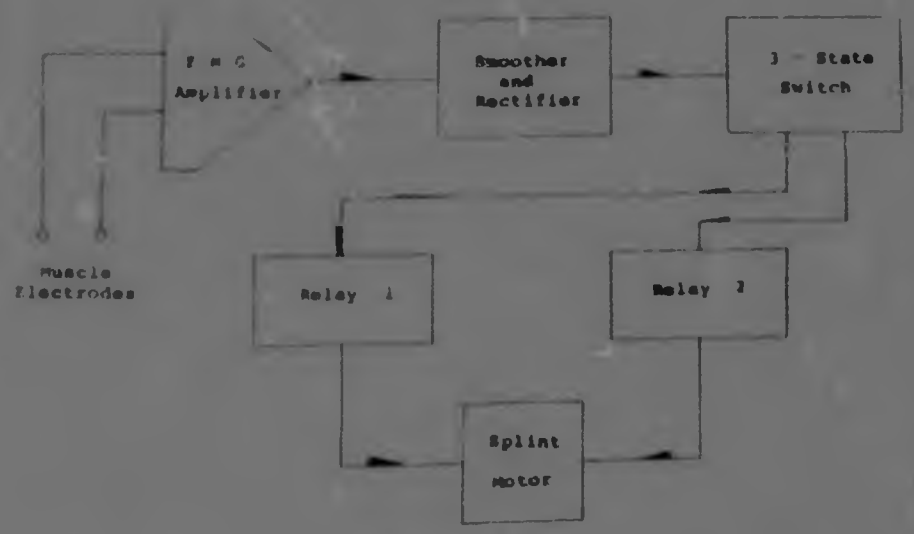


Fig.6.14. Block Diagram of System 3.

The above figure illustrates in block diagram form the system operation. The myo-electric amplifier is the same as used in systems 1 and 2 (Chapter 5.3.1.) as is the smoothing and rectifying circuit. The three level discriminator consists of a modified Schmitt Trigger (Chapter 4.5.) (Fig.6.15.).

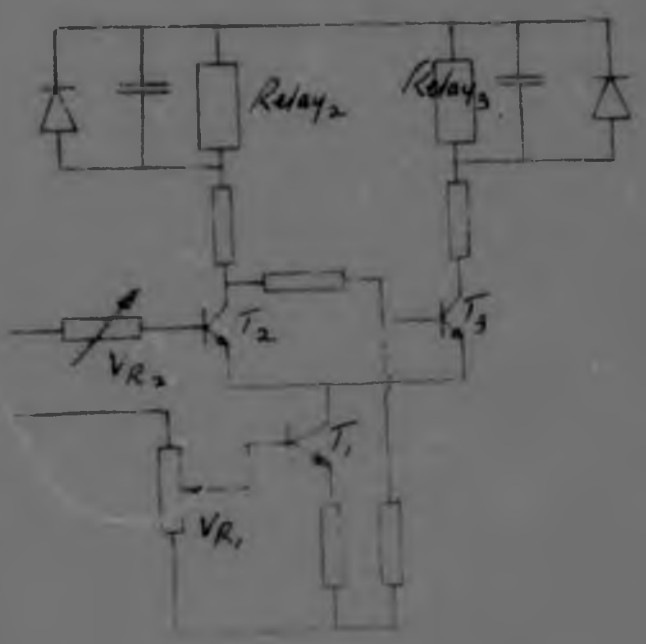


Fig.6.15. Modified Schmitt Trigger.

When the output of the rectifier and smoother reaches a level that permits enough base current to flow into T_1, such that the current flowing through the collector of T_1, and through the collector-emitter of T_3, is sufficient to close Relay 3, the motor is connected in such a way as to close the splinted fingers. When the input level exceeds a certain preset level (set via VR_1), T_2 is switched on and T_3 is switched off, (standard Schmitt Trigger action), and the motor is now connected to open the splinted fingers. If the input level is insufficient to switch T_1 on, the motor will be static and the fingers will remain stationary. The two different switching levels are set up to suit each particular patient; this is effected by adjusting the two variable resistors V_{R1} and V_{R2}.

6.4.10. Practical Assessment of System.

System 3 has been found to work very well and has consequently been used extensively in the control systems of the quadruplegic patients treated. The success of the systems may be attributed to the fact that the patient is required to provide a signal only at the time at which he wishes to change the position of his fingers. He does not have to maintain a constant level of muscle activity (as required in system 2), to maintain a desired finger position. Also, the system is relatively insensitive to transient signals caused by mains interference (resulting from switching on or switching off of electrical appliances in the immediate proximity of the pickup electrodes). The transient signal might cause a slight momentary rotation of the orthesis motor but the control is not affected as system 1 would be.

6.4.11. Other Control Systems.

In cases where there is more than one control muscle available as in quadruplegics who have lesions at levels of C_6/C_7, it is possible to simplify the control system quite considerably. The combined extensor carpi radialis and brachii radialis muscle group are no longer required to provide all the orthesis control information, and consequently, it is not necessary to resort to sequential movement selection or 3-state myo-electrical switch systems as described above. The muscles from which the second control signal is derived, must be anatomically remote from the first muscle group, that is, from the combined extensor carpi radialis and brachi radialis. If the two muscle groups are too near to each other, 'cross-talk' or 'overlapping' between the two control signals may occur through the skin. This results in unreliable and unpredictable orthesis behaviour.

The first of the patients fitted with this dual muscle control system[8] had good wrist extensors and flexors.

The pick-up electrodes were placed over these two muscles, and when the patient required to open the hand splint, he merely extended his wrist or produced an isometric contraction of his wrist extensors. When wishing to close the hand splint, a corresponding set of conditions were volitionally generated in the wrist flexors. Because the wrist extensors and flexors are psychologically associated with normal hand opening and closing, the patient experienced little difficulty in learning how to operate his orthesis. As wrist extensors and flexors were both present, external stabilisation of the wrist as previously described was not at all necessary. To accommodate this extra degree of freedom, that of wrist extension and flexion, the splint was made to extend only over the M.P. and I.P. joints, thus leaving the wrist joint free. A 'dead zone' is inherent in the electronic motor control circuitry. This 'dead zone' permits myo-potential activity of intensities below a certain level to be ignored by the control circuitry. The patient may therefore carry out slight wrist movements without triggering the control circuitry, but has to limit his wrist activity in order to maintain complete control over the orthesis. The electronic system is illustrated in fig.6.16. and is similar in all respects to the system described in Chapter 5.

The second patient fitted with a dual myo-control system used pronator teres as the muscle controlling orthesis closing. Extensor carpi radialis and brachi radialis are used to control orthesis opening. In this case it was necessary to re-educate the pronator teres so that the patient associated this muscle with orthesis closing. This the patient was able to accomplish quite easily but could not operate his orthesis in all positions of elbow extension.

Since the speed of the orthesis opening and closing is slow, and because the patients control muscles are weak, it was not considered necessary to make the operation of the orthesis motor proportional to the e.m.g. control signal. Adequate splint operation is available using an 'on-off' relay operated system as shown in fig.6.17. which replaces the circuitry in fig.6.16.

6.5. General Difficulties Experienced with All Systems.

1. The signals used for orthesis control are extremely weak and consequently an extremely high gain amplifier is required to provide a level of signal suitable for processing by the motor control circuitry. Amplification factors of up to 10^5 need to be employed. Because of the high output impedance of the signal source, and its consequent susceptibility to mains interference, the first stage differential amplifier must possess good common mode rejection. This can best be achieved by carefully selecting and matching the transistors used in the input stage (fig.5.2.). Even then, one cannot be sure that the common mode rejection will be adequate to balance out interference. Stray capacitances, unshielded leads, or leads of different lengths can also contribute to the deterioration of common mode rejection. The problem can best be overcome by including a filter with maximum rejection at 50cps. The filter employed is a 'parallel-T' type similar to that employed in the Myo-electric Prosthesis circuitry. (In the case of the hand prosthesis the filter was not absolutely necessary but was added as a sophistication so that very intense electrical interference did not impede the reliability of the prosthesis action. In the orthotic control systems the filter becomes quite necessary because the myo-control signal is so weak, and the control reliability would be severely impeded if it was not included.)

2. The orthesis motor generates a large amount of high frequency noise when operating due to the commutation of the brushes on the rotor. At high amplification sensitivity, the motor noise interference becomes so great that it can no longer be tolerated by the system and is interpreted as part of the control signal. As the motor ages the noise increases progressively to a point where at even low amplification sensitivity the control system becomes completely unreliable and the patient cannot use his apparatus with confidence.

Fortunately, the noise generated by the motor consists of high frequency spikes - the motor rotates at ±18,000rpm. The noise frequency is therefore much higher than the maximum myo-electric signal information carrying frequencies, and it (the noise) can conveniently by excluded without any loss of myo-signal by desensitising the amplifiers at high frequencies. The feedback capacities in the common emitter amplifier stages (fig.5.4.) were adjusted in an attempt to reduce the high frequency gain. This was partially successful but the high frequencies were not attenuated at a fast enough rate. The problem of motor noise interference was finally overcome by employing another filter. The filter was required to cut off sharply at about 750-1,000cps., and thus eliminate any interference satisfactorily.

Low pass filters are easily constructed using inductors and capacitors of various values in a number of circuit configurations. The winding of inductors is rather tedious, and the calculations for their design can be time consuming; their bulk and weight are usually inhibitive for low frequency filter applications. For these reasons, it was decided to use an active filter. The type of active filter used is of the Rauch type (fig.6.18.) and the capacitors used are so selected to give a third order Chebishev filter with maximum high frequency cut off slope. (See Appendix F). Although the requirements of the capacitor selections are rather stringent as far as tolerances are concerned, the filter was constructed using standard mylar capacitors. The operational amplifier A, (fig.6.19) consists of an integrated circuit, Texas INstrument Type SN724. The filter has been found to operate satisfactorily and has been included where necessary in the orthesis control systems.

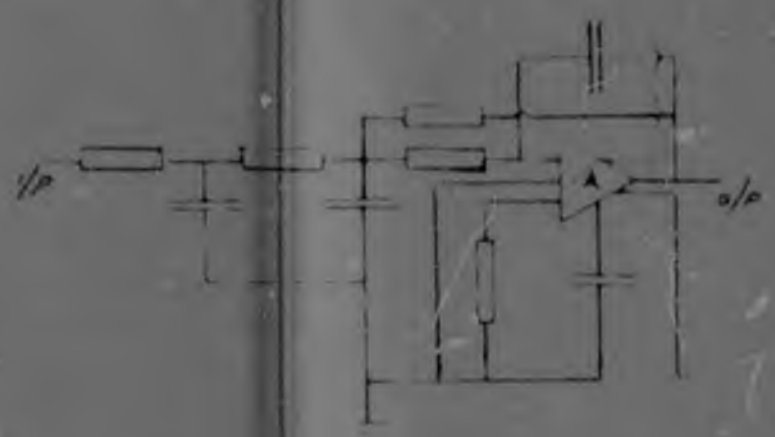


Fig.6.18. Rauch Low-Pass Filter.

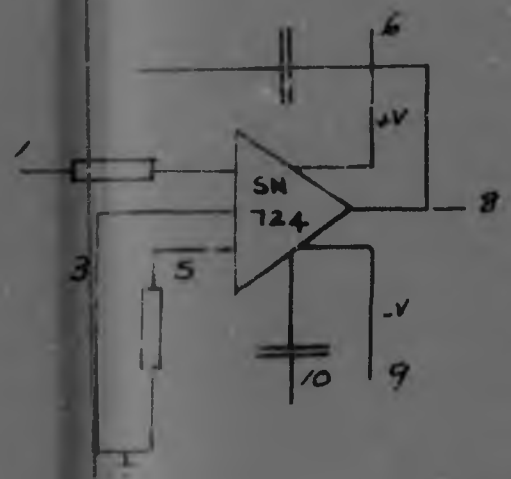


Fig.6.19. Connections of Operational Amplifier (Type SN724), as employed in Rauch Filter.

3. The problem of providing power for the control electronics and motor was overcome by using three 6 volt motor cycle batteries, housed in a container mounted in the rear of the patient's wheelchair. (See photograph 6.10.) The same container provides a housing for the electronic unit, and the on/off switches and sensitivity control are mounted on the front panel of the container. Recharging of the batteries is effected by using a commercially available battery charger, and charging is required normally only once every month.

4. The electrodes that are used to monitor the myo-potentials are made out of stainless steel rod approximately $^1/_2$" in diameter. The electrodes are then mounted on a stiff plastic separator which is strapped about the muscle being monitored. Unfortunately, it is found that the electrodes tend to slip off the muscle, resulting in a diminished myo-potential and consequent unreliable orthesis control. Commercially, a number of electrodes are available. These might help to overcome the problem of electrode movement, but they would also result in increased operating cost. In order to overcome the electrode problem it might become necessary eventually to use implanted percutaneous electrodes, or, to minimise the risk of infection, to use endoradiosonde type electrodes transmitting myo-potentials to an external radio receiver. Also, it has been suggested that 'skin pockets' be made for the patient, using suitable plastic surgery techniques, in order to keep the electrodes in position.

5. The motor and gearbox used to power the splints have been found to be underpowered at their rated operating points. Even when the motor ratings were substantially exceeded to an extent that reduced the useful motor life drastically, the splint speed and grip were not adequate. It is this problem that must be overcome before success can really be claimed in orthesis operation. A larger motor must be used to provide the necessary requirements

of speed and torque. This means that the motor will no longer be situated on the splint itself but will perhaps be mounted on the patient's wheelchair, Power to the splint will be conducted via a Bowden cable system.

6.6 Patient Training.

After the patients have been fitted with their splints and are able to control them satisfactorily, the patients are referred to the Occupational Therapy Department for further training. Here, a training programme has been especially devised for quadruplegic patients and they are encouraged to take part in activities designed to improve dexterity and proficiency in splint control. The patients are taught the basic writing patterns and practice results in a neat and legible script. Writing takes much of the patients' time because any realistic rehabilitation programme must take into account the fact that the patient is confined to a wheelchair, and can only undertake light sedentary work. Most clerical jobs could be attempted by the quadruplegic and a minimum requirement of such work would be the ability to write. Figure 6.20 and figure 6.22 show patients engaged in writing and dexterity activities respectively. A sample of a patient's writing is illustrated in figure 6.21.



Fig.6.20. Patient Writing with Myo-Electric Hand Splint.

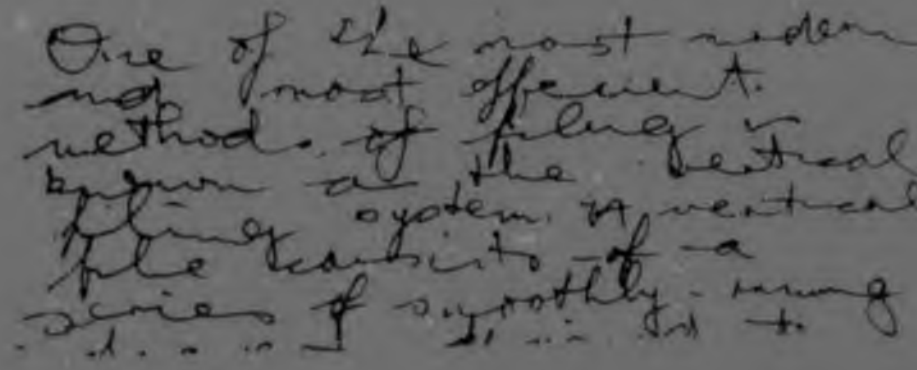


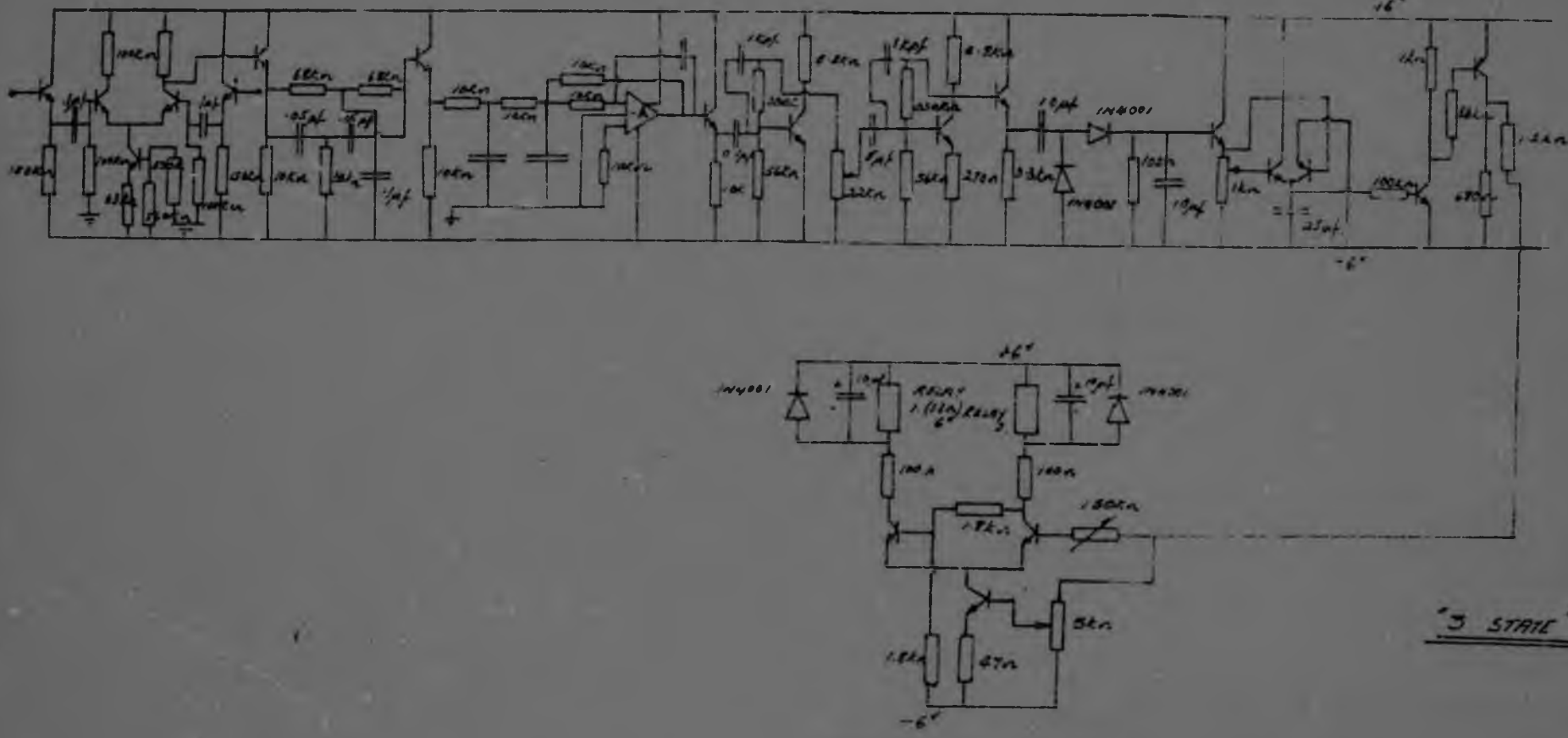
Fig.6.21. Example of Patient's Writing, using Powered Orthesis.

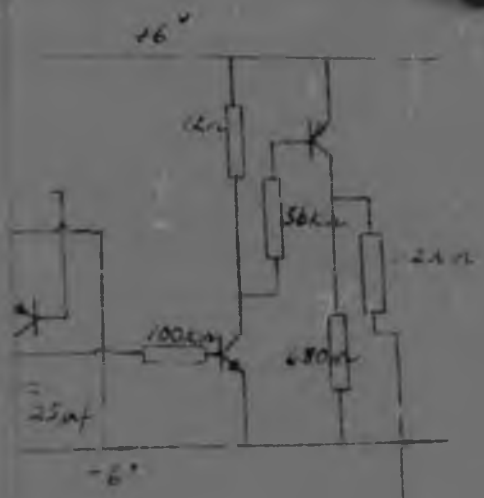


Fig. 6.22. Patient practising dexterity exercises.

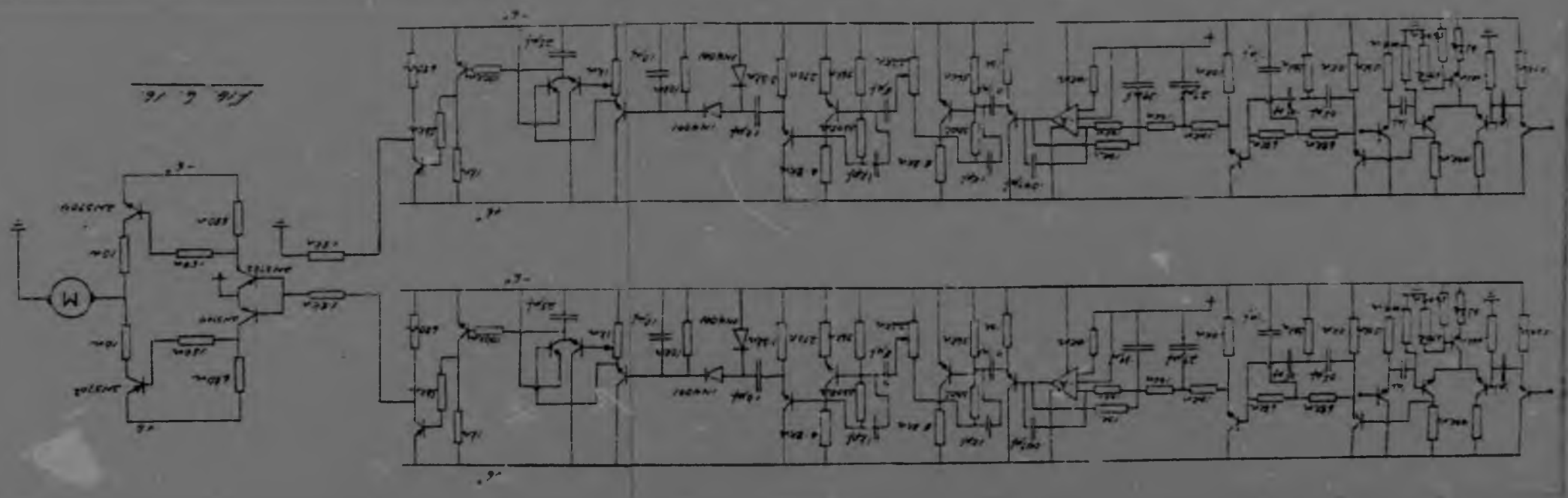


Fig.6.23. A toast to a brighter future.


+6"
-6"
1N4001
1N4001
1N4001



"3 STATE" ORTHESIS CONTROL SYSTEM



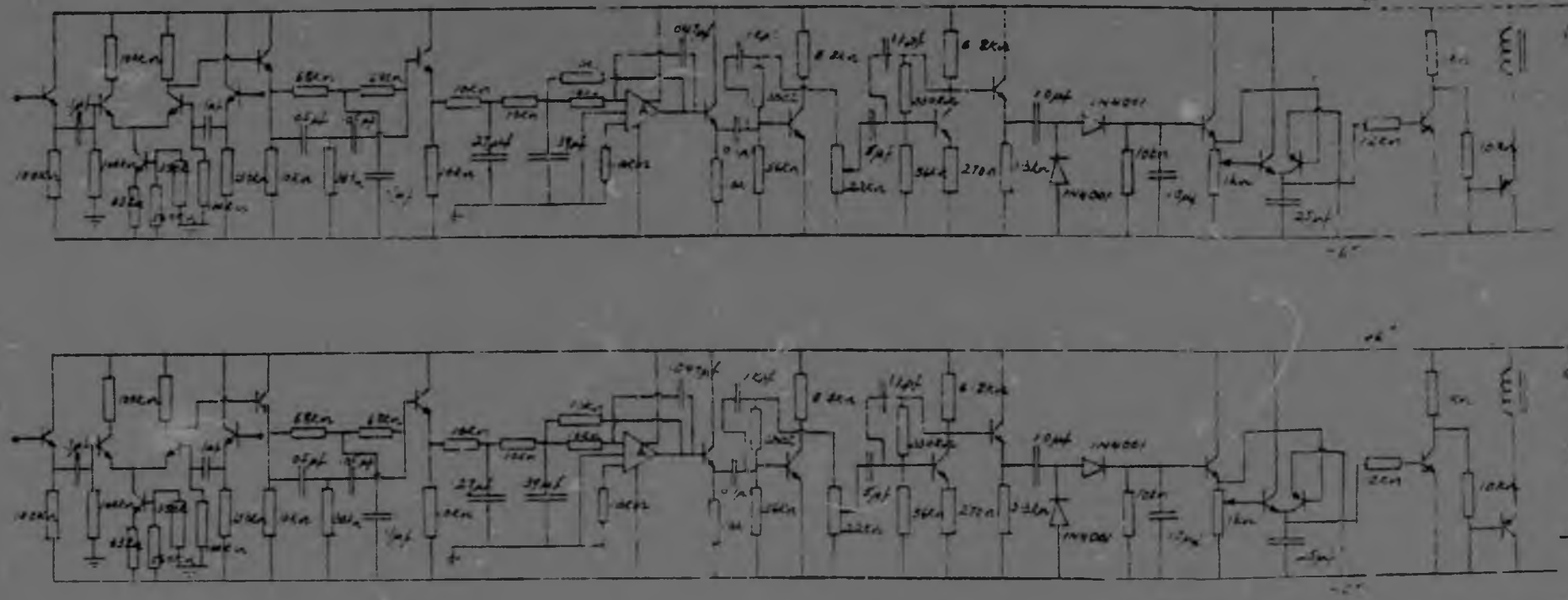


FIG. 6.17.

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CHAPTER 7.

The appliances developed and presented in Chapters 5 and 6, have by no means reached the final states of development. There are a number of improvements which can be made to the prostheses and orthoses.

7.1. Possible Improvements to the Myo-electric Hand Prostheses as Presented in Chapter 5.

7.1.1. Prosthesis to Patient Feedback.

As has been stressed in section 2.3., the concept of prosthesis to patient feedback is a most important one. Some means of knowing what the prosthesis output is, must be available to the patient. The Bottomley Hand (Chapter 4.3.) does attempt to accomplish this by relating the velocity and force output of the prosthesis to the volitional e.m.g. output from the controlling muscles.

In the system described in Chapter 5, although speed and force output are related to the myo-electric signal output, unlike the Bottomley Hand, there is no feedback to the patient. After a time however, the patient does learn what the effect of the various levels of volitional myo-potential that he can elicit will be. He will know, for instance, what gripping force corresponds to that particular musc. tension he is exerting. In the myo-electric prostheses, as is the case with other kinds of prostheses, visual feedback is virtually a necessity. Sight must play a most important part in learning how to use the prosthesis effectively.

It is suggested that some kind of proprioceptive feedback (section 2.3.3.) should be designed for the myo-electric system. Pressure sensitive pads, mounted in the finger tips of the prosthesis could relay information of the pressure being exerted to the patient, and similar pressure sensitive areas on the palm could serve the same purpose. The author has conducted some experiments

to see if this idea could be put into practice.

A strain-gauge was mounted on a piece of copper sheeting moulded around the finger of an experimental prosthesis. The strain-gauge output was influenced by the deformation of the sheet caused by pressure on the finger tip. The pressure signal so obtained, was used to modulate a simple square wave generator. The generator output pulses were fed to two electrodes attached to the arm, thereby providing a method of feedback by causing a mild stimulus at the electrode sites. As the pressure on the finger increased, the amplitude of the pulses at the two electrodes increased proportionately, causing the intensity of the electrical stimulus to increase. The electronic circuitry required to achieve this direct type of feedback is rather bulky and therefore cannot be included in a practical prosthesis system at present. The results were promising and it is felt that this is one method of direct prosthesis patient feedback that should be further investigated. An immediate application is for sufferers of leprosy who have lost all sensation in the fingers.

7.1.2. Speed of Operation of Prosthesis.

The speed of operation of the prosthesis is not yet fast enough (approximately $1^{1}/_{2}$ radians/second maximum). This speed should be at least doubled to result in a more functional prosthesis. The motor powering the prosthesis terminal device (type A.E.C. 0015) provides substantial force output ($\pm$ 30lbs.) with the present gear ratios, and it is felt that a sacrifice in force output could be made to obtain a faster maximum speed. It is hoped that a laboratory will soon be suitably equipped to enable the construction of the required gear-train mechanisms.

7.1.3. The Weight of the Batteries.

The weight of the battery pack normally employed in the myo-electric hand prosthesis is 642grms. This is not

too cumbersome for an adult, but it is felt that the pack is rather heavy for children to carry. As suggested in Chapter 5.3.8., a set of smaller batteries might be used, although this would entail having to change the batteries more than once daily. Until batteries with a better energy density have been developed, this would seem to be the only practical solution.

7.1.4. Prosthesis Fitting.

Not much experience has been gained in this country in using the Münster technique (Chapter 5.3.6.) Consequently, difficulty has been experienced in obtaining correct socket-stump fittings. However, as the myo-electric type of prosthesis becomes more popular, local limb-fitting will no doubt improve, and it is merely a matter of time before the correct techniques are learnt. Special machinery to facilitate plastic pouring for the fabrication of the sockets is to be installed in a proposed upper limb prosthetics unit in Pretoria.

7.2. A Glimpse into the Future.

7.2.1. Multi-channel Control.

The possibility of using implanted or percutaneous electrodes is a very real one, and the advent of suitable techniques for facilitating the use of these electrodes will certainly open new vistas in the field of prosthesis control. As many channels as required for individual finger movements and/or pronation and supination, and wrist extension and flexion, could be monitored and prosthesis control would be greatly improved. The mechanical problems of suitable finger linkages, etc., and motor location seem to be the major problems associated with these advanced prostheses.

7.2.2. Hybrid Systems.

In Chapter 4.5. the myo-electric control system of

Dorcas and Scott is discussed and attention is drawn to the fact that a hybrid system is employed, in the sense that both myo-electric control and shoulder harness control are used. Normally the patient would not have been able to operate both the elbow prosthesis and the hand prosthesis simultaneously; he is however, able to do this with the hybrid control system.

The above-elbow amputee is an ideal candidate for a hybrid type control. He normally has both biceps and triceps muscle groups available for hand prosthesis control and can inspan a shoulder harness for elbow control, or vice versa. This could form the basis for an interesting investigation, and it is planned to develop such hybrid systems when the opportunity arises.

A hybrid system could be employed to provide pronation and supination as well as finger extension and flexion in a myo-electric prosthesis for mid-forearm amputees.(1) Finger extension and flexion are effected in the usual way (myo-electrically) and the mechanical movement of the amputation stump may be used to trigger two micro-switches mounted in the socket. Usually the amputee does retain some measure of pronation and supination and this could be used to provide the necessary control signals.

7.3. Possible Improvements to the Orthotic Splint Systems as Presented in Chapter 6.

7.3.1. Speed of Operation.

The speed of operation of the splint is totally inadequate, ($\pm$ $^1/_2$ radian/sec. maximum), and to foster patient acceptance of powered orthotic hand splints it is necessary to increase the speed in some way. The motor employed at present cannot provide adequate power and it should be replaced by a larger motor. The larger motor however, would no longer be suitable for mounting on the splint itself, but would have to be mounted on some

convenient part of the wheelchair. Power should then be transmitted to the prosthesis via a Bowden cable system.

7.3.2. Difficulties of Fitting Hand-splints.

Each patient is individually fitted by the orthotist to ensure a good fit. It would be extremely convenient if a splint that had standard component parts in various sizes made to fit different size hands, could be developed. The orthotist could then assemble the hand splint from standard components available 'off the shelf'. Also, the orthesis is made of stainless steel and patients feel the metal parts on their hands draw added attention to themselves, and thus are a source of embarassment. This could be overcome by making the component splint parts out of a transparent plastic. The use of plastics too, might facilitate fabrication of the splint if it could be moulded directly on the patients' hands.

7.3.3. Electrodes.

As discussed in Chapter 6.5.4., the electrodes used for monitoring the control signals are inadequate and a method of signal monitoring must be developed to increase the reliability of the system. The possibilities suggested in Chapter 6.5.4. should be investigated. The possibility of using percutaneous or implanted electrodes should also be considered, and it is perhaps in this area that the best solution lies.

7.4. General Remarks on Teamwork in a Unit for Quadruplegics.

The engineer cannot attempt the rehabilitation of a quadruplegic patient by himself. His work is only part of a long process of rehabilitation and he must consider himself as a member of a team striving for a common goal - to return to the patient as much independence as is possible under the circumstances.

Without good physiotherapy, the patient's muscles can never be of a quality to control an orthesis. Without occupational therapy, all the engineer's work will be of no avail - the patient must be taught how to use his splint effectively. He must be made to realise that the splint is a part of himself, to be used all the time if maximum benefit from the appliance is to be gained.

Good nursing and medical attention is necessary to ensure that the patient's health is maintained in spite of his general weak condition. The patient's mental health is another problem of concern. The frustration and depression that accompanies quadruplegia cannot be imagined, and it is the psychologist who has the most difficult task of all - that of persuading the patient that he can still be a productive member of society in spite of his condition. The patient's parents and family too have to be educated as far as the needs, both mental and physical, of the patient are concerned.

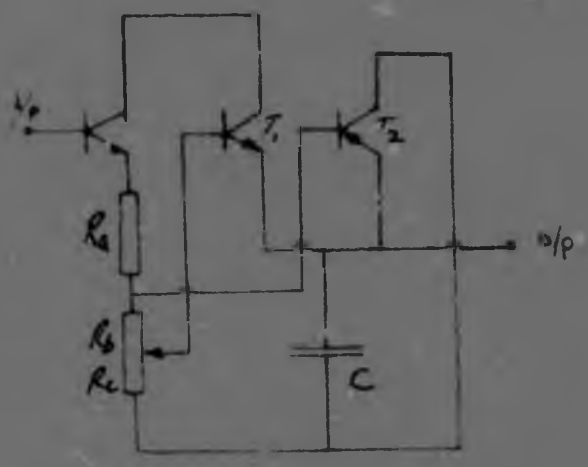
From the foregoing, it is seen that teamwork is extremely necessary and important in any rehabilitation project for quadruplegics. (These remarks apply to amputees and in fact all disabled people to a certain degree.) Doctor, psychologist, nurse, physiotherapist, orthotist, systems engineer, and occupational therapist all have vital parts to play in this team. It is only through this concerted interdisciplinary co-operation that worthwhile results can be expected.

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APPENDIX A.

AUTOGENIC BACKLASH GENERATOR OPERATION.



BACKLASH GENERATOR.

Assume $R_a = R_b = R_c$ then if the input voltage to the circuit is -9v then $V_{R_c} = -3$ Volts, $V_{R_b} = -6$ Volts and $V_{R_a} = -9$ Volts.

Due to emitter follower action T_{1e} is also at -3 volts. T_2 remains off as V_{T2b} is less than V_{R2e}. Current flows into capacitor C via [illegible] and it charges up to -3 volts.

Now assuming that the input voltage goes to -6 volts, the voltage at Rc will then be -2 volts. This is more positive than the voltage at B and consequently T_1 will turn off. T_2 still remains in the off state and the output is still at -3 volts.

Finally let us assume that the input voltage goes to -3 volts. $V_{Rc} = -1$ volt, $V_{Rb} = -2$ volts and $V_{Rc} = -3$ volts. By the same reasoning as in previously T_1 remains off. However, since $V_{Rb} = -2$ volts and is greater than the voltage at B, T_2 will switch on and the capacitor will discharge to V_{Rb} through T_2.

It is clearly seen that the degree of backlash is determined by the value of resistor R_b. For a more accurate analysis V_{eb} drops of the transistors must be considered.

APPENDIX B.

MYO-ELECTRIC SIGNAL AMPLIFIER FREQUENCY RESPONSE.

Graph 1. Myo-electric signal amplifier frequency response of the amplifier used in the hand prosthesis.

Graph 2. Myo-electric signal amplifier frequency response of the amplifier used in the hand orthesis.

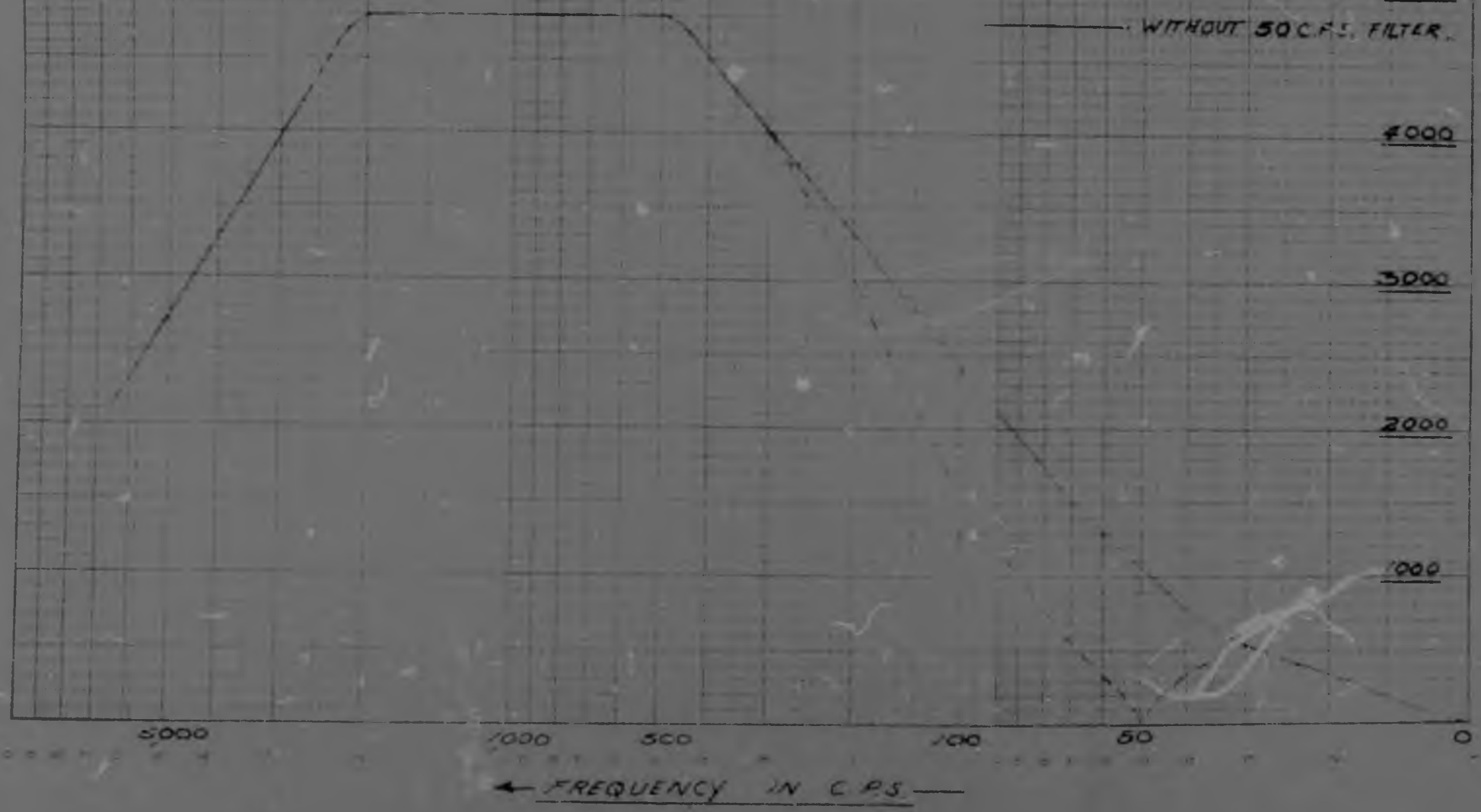
AMPLIFIER OF HAND PROSTHESIS

1.

MYO-ELECTRIC SIGNAL AMPLIFIER FREQUENCY RESPONSE

– – – – – WITH 60 C.P.S. FILTER

——— WITHOUT 50 C.P.S. FILTER.



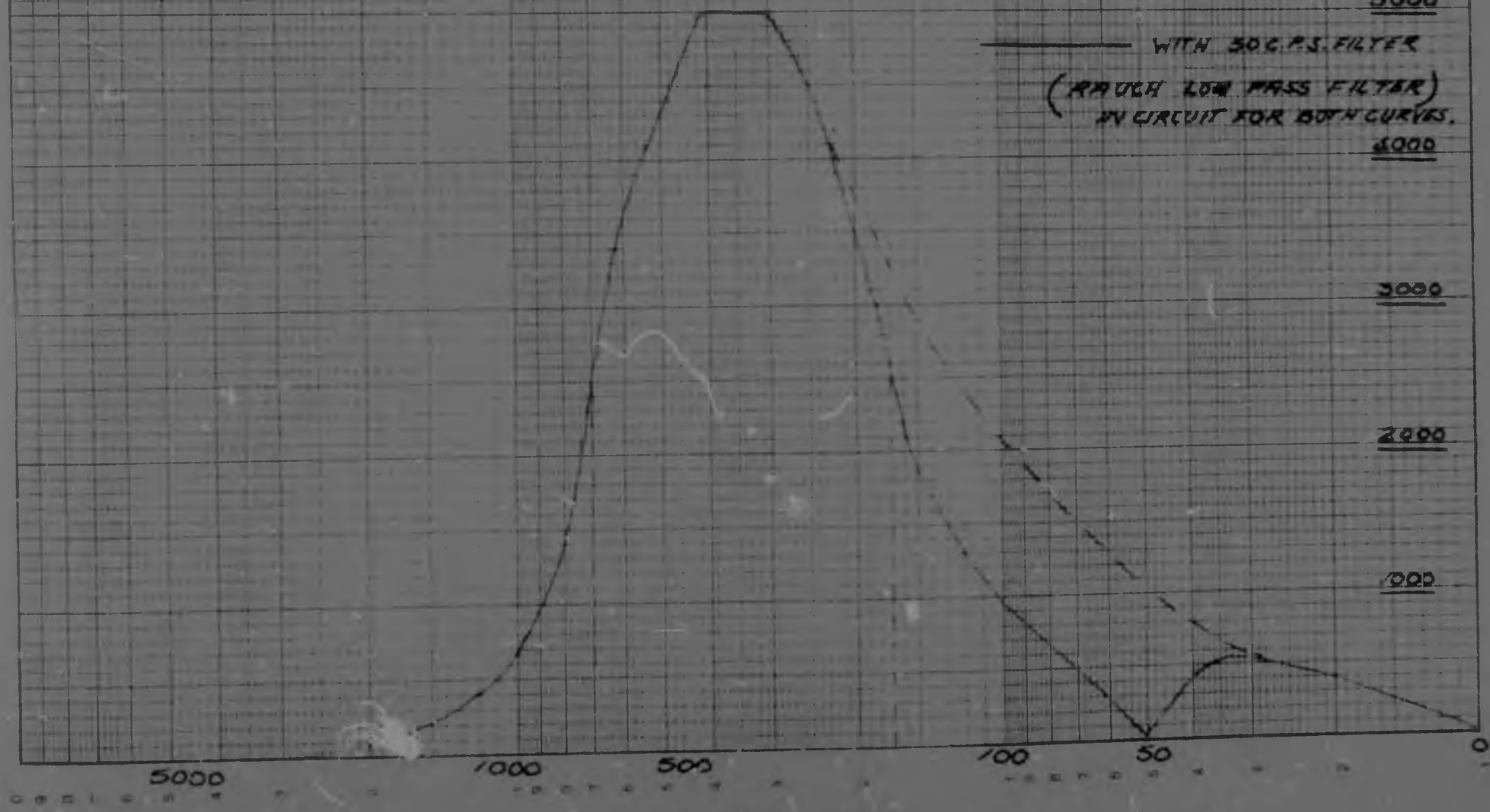
AMPLIFIER OF HAND ORTHESES.

MYO-ELECTRIC SIGNAL AMPLIFIER FREQUENCY RESPONSE

– – – – – WITHOUT 50 C.P.S. FILTER

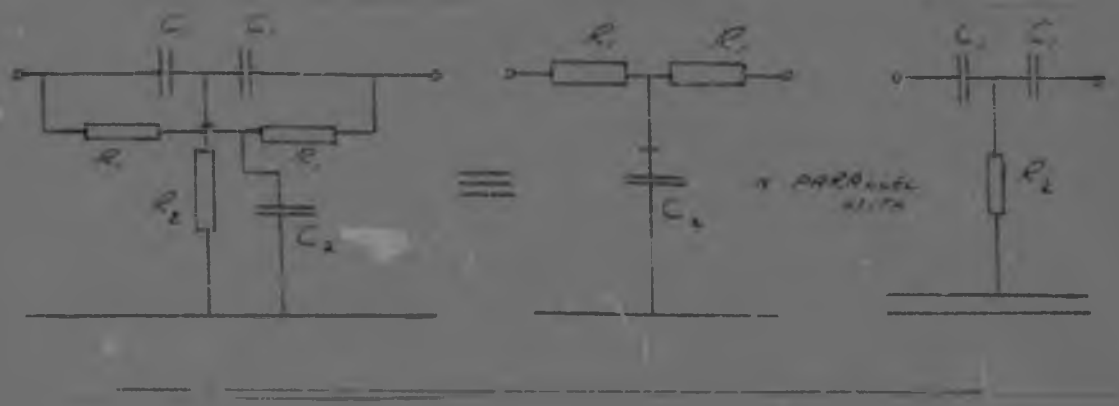
——— WITH 50 C.P.S. FILTER

(ROUGH LOW PASS FILTER) IN CIRCUIT FOR BOTH CURVES.



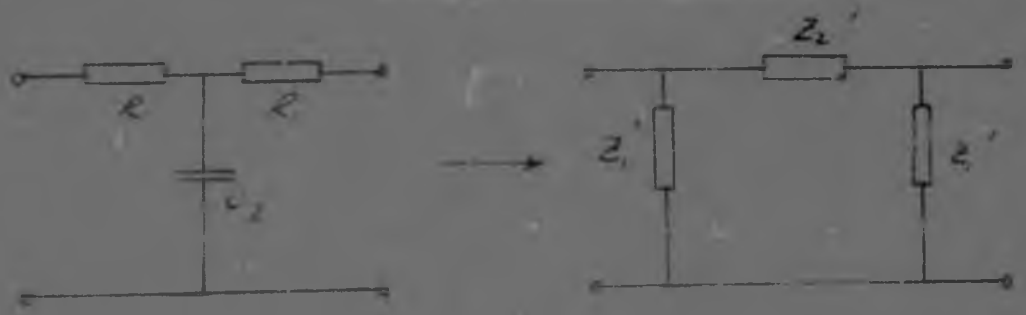
APPENDIX C.

TWIN T FILTER WITH MAXIMUM REJECTION AT 50 C.P.S.



To analyse the twin T circuit above the equivalent 'star' circuits above are transformed to their corresponding 'delta' circuits as follows.

(a)



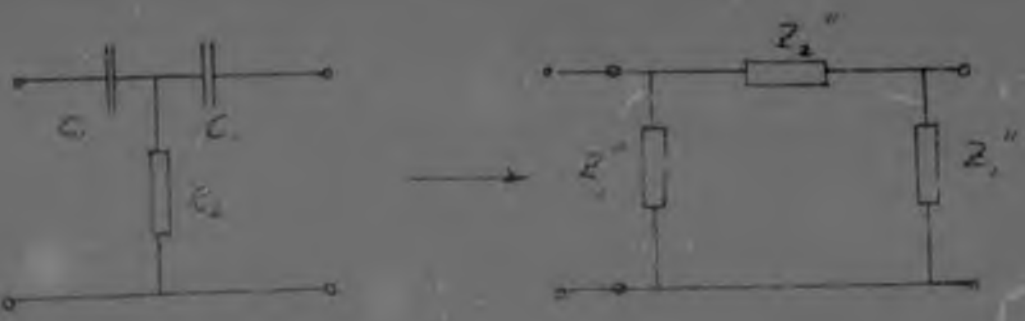
$$z_1^1 = R_1^2 \left(\frac{1}{R_1} + \frac{1}{R_2} + \frac{1}{-jX_2}\right)$$

$$= 2R_1 - \frac{R_1^2}{jX_2}$$

$$z_2^1 = -R_1 jX_2 \left(\frac{2}{R_1} + \frac{1}{-jX_2}\right)$$

$$= R_1 - 2jX_2$$

and (b)



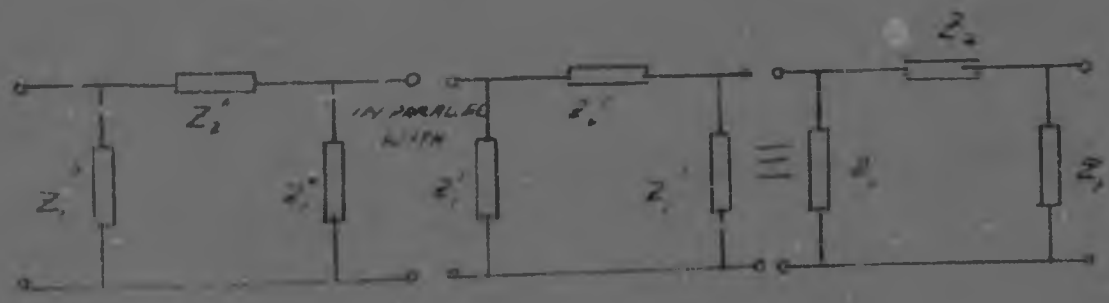
$$z_1^{11} = (-jX_1)^2\left(\frac{1}{-jX_1} + \frac{1}{-jX_1} + \frac{1}{R_2}\right)$$

$$= -2jX_1 - \frac{X_1^2}{R_2}$$

$$\text{and } z_2^{11} = (-jX_1)R_2 \cdot \left(-\frac{2}{jX_1} + \frac{1}{R_2}\right)$$

$$= 2R_2 - jX_1$$

Considering the two delta circuits in parallel:-



$$z_1 = \frac{z_1^1 z_1^{11}}{z_1^1 + z_1^{11}}$$

$$\text{and } z_2 = \frac{z_2^1 z_2^{11}}{z_2^1 + z_2^{11}}$$

The rejection of the signal becomes infinate when z_1 becomes infinite, i.e. when $z_1^1 + z_1^{11} = 0$

The transmission factor is defined as

$$= \frac{Z_2}{Z_1 \; Z_2}$$

at max. rejection Z_1

therefore Z_2 can be neglected with respect to Z_1

$$\therefore \quad = \frac{Z_2}{Z_1}$$

$$= \frac{Z_2^1 \cdot Z_2^{11}}{Z_2^1 + Z_2^{11}} \cdot \frac{Z_1^1 + Z_1^{11}}{Z_1^1 \; Z_1^{11}}$$

near balance i.e. near max. rejection $Z_1^1 + Z_1^{11} \longrightarrow 0$ and it is expected that this term will influence most at this point.

Let $Z_1^1 + Z_1^{11} = \beta$

therefore $\beta = 2R_1 - \frac{R_1^2}{R_2} + (-2jX_1 - \frac{X_1^2}{R_2})$

$$= 2R_1 - \frac{X_1^2}{R_2} + j(\frac{R_1^2}{R_2} - 2X_1)$$

at balance $\beta = 0$

$$\therefore \; 2R_1 = \frac{X_1^2}{R_2} \text{ and } \frac{R_1^2}{X_2} = 2X_1$$

$$\text{or } R_2 = \frac{X_1^2}{2R_1} \text{ and } X_2 = \frac{R_1^2}{2X_1}$$

For a maximum slope twin T circuit

$$R_1 = X_1$$

$$\therefore \quad R_2 = \frac{-X_1}{2} = \frac{R_1}{2_1}$$

$$\text{i.e. } R_1 = 2R_2 = X_{1_1}$$

$$\text{and } X_2 = \frac{R_1}{2} = \frac{X_1}{2}$$

$$\text{i.e. } X_1 = 2X_2 = R_1$$

therefore $R_1 = 2R_2$ and $C_2 = 2C_1$

Considering f_c = 50 cps.

choosing $C_1 = .05\,\mu f$

$\therefore \quad C_2 = 0.1\,\mu f$

$\therefore \quad X_1 = \frac{1}{\omega c.}$

$$= \frac{1}{2 \quad 50 \times 0.05 \times 10^{-6}} \text{ ohms.}$$

$$= \frac{10^6}{5 \times} \text{ ohms.}$$

$$= \frac{10^6}{100} = 64 \text{ K-ohms.}$$

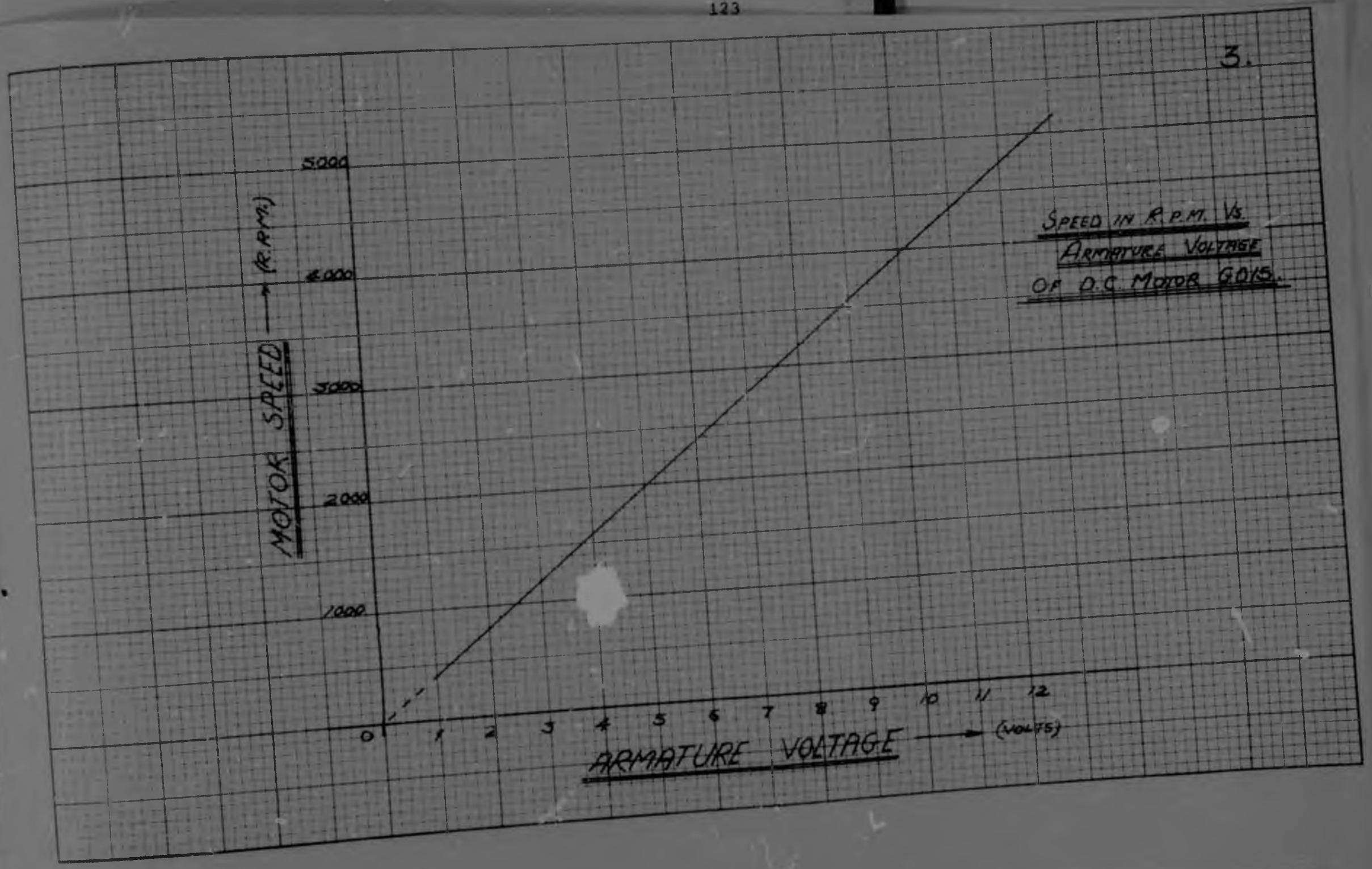
$\therefore \quad R_1 = 64$ K-ohms.

and $R_2 = 32$ K-ohms.

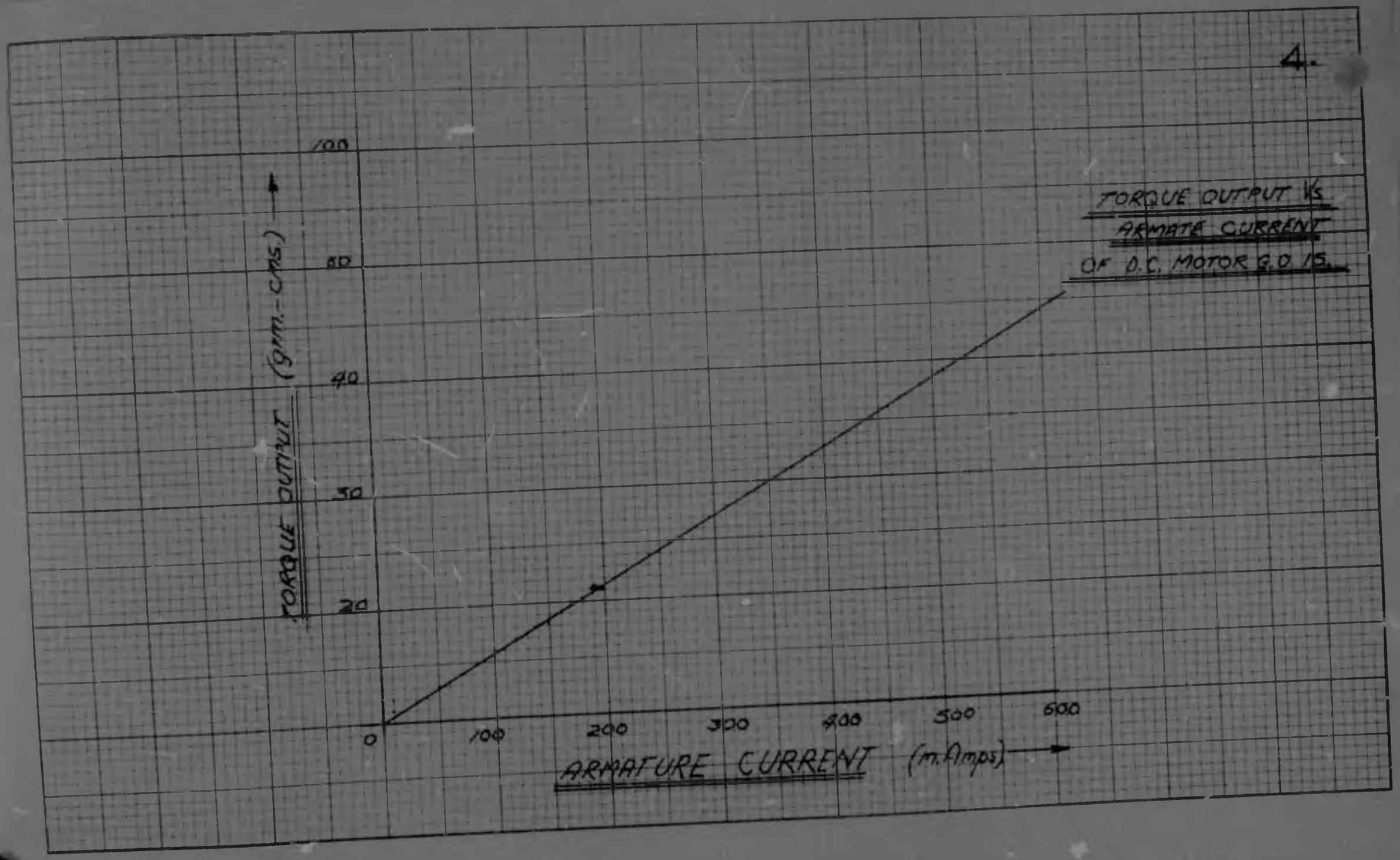
The nearest standard values to R_1 and R_2 are 68 K-ohm. and 33 K-ohm. respectively. As shown in the graphs of the Myo-electric signal amplifier response (Appendix C), the choice of 68 K-ohm. and 33 K-ohm. is adequate.

3.

SPEED IN R.P.M. VS.
ARMATURE VOLTAGE
OF D.C MOTOR GO15.

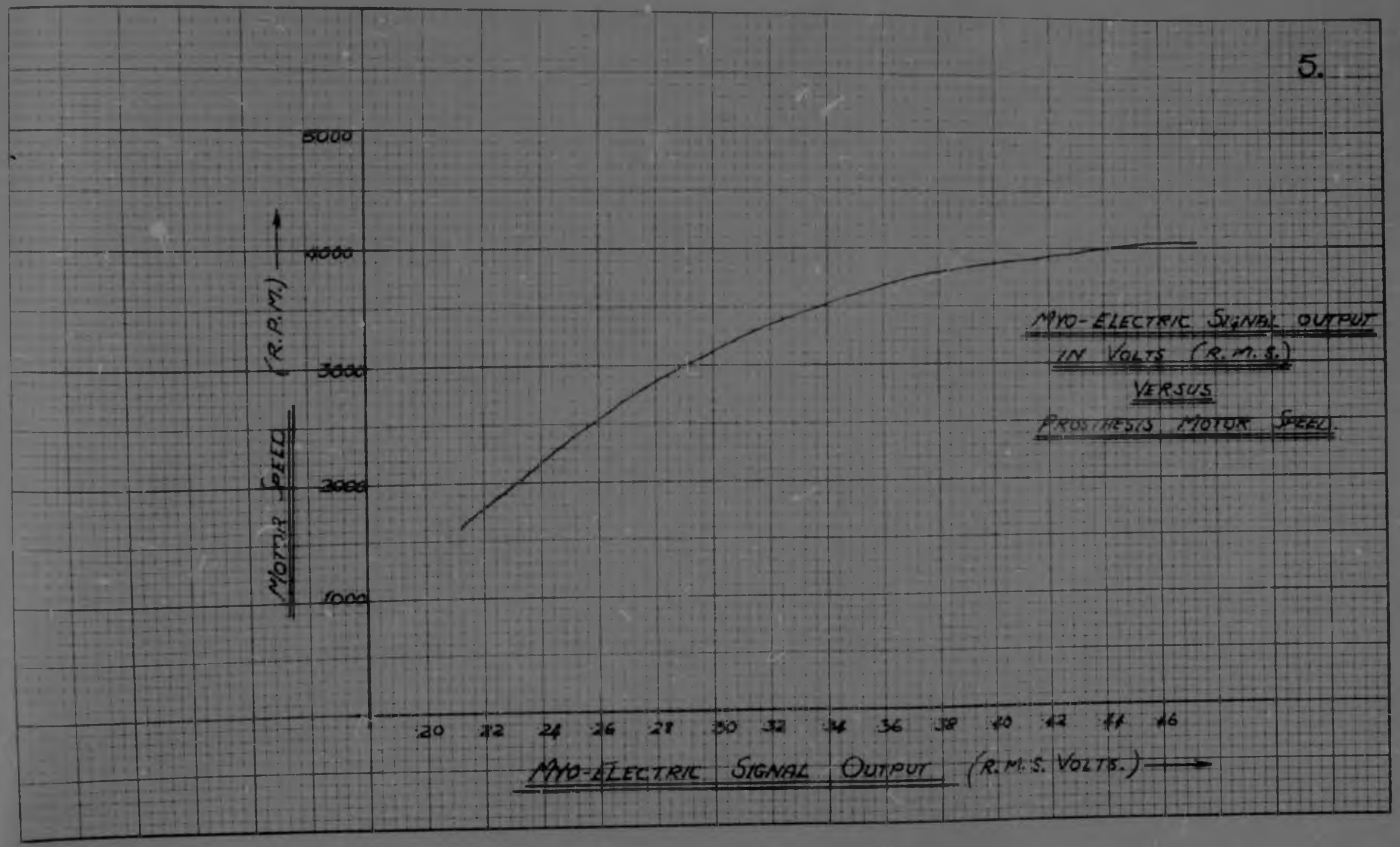


TORQUE OUTPUT Vs
ARMATE CURRENT
OF D.C. MOTOR G.O 15.

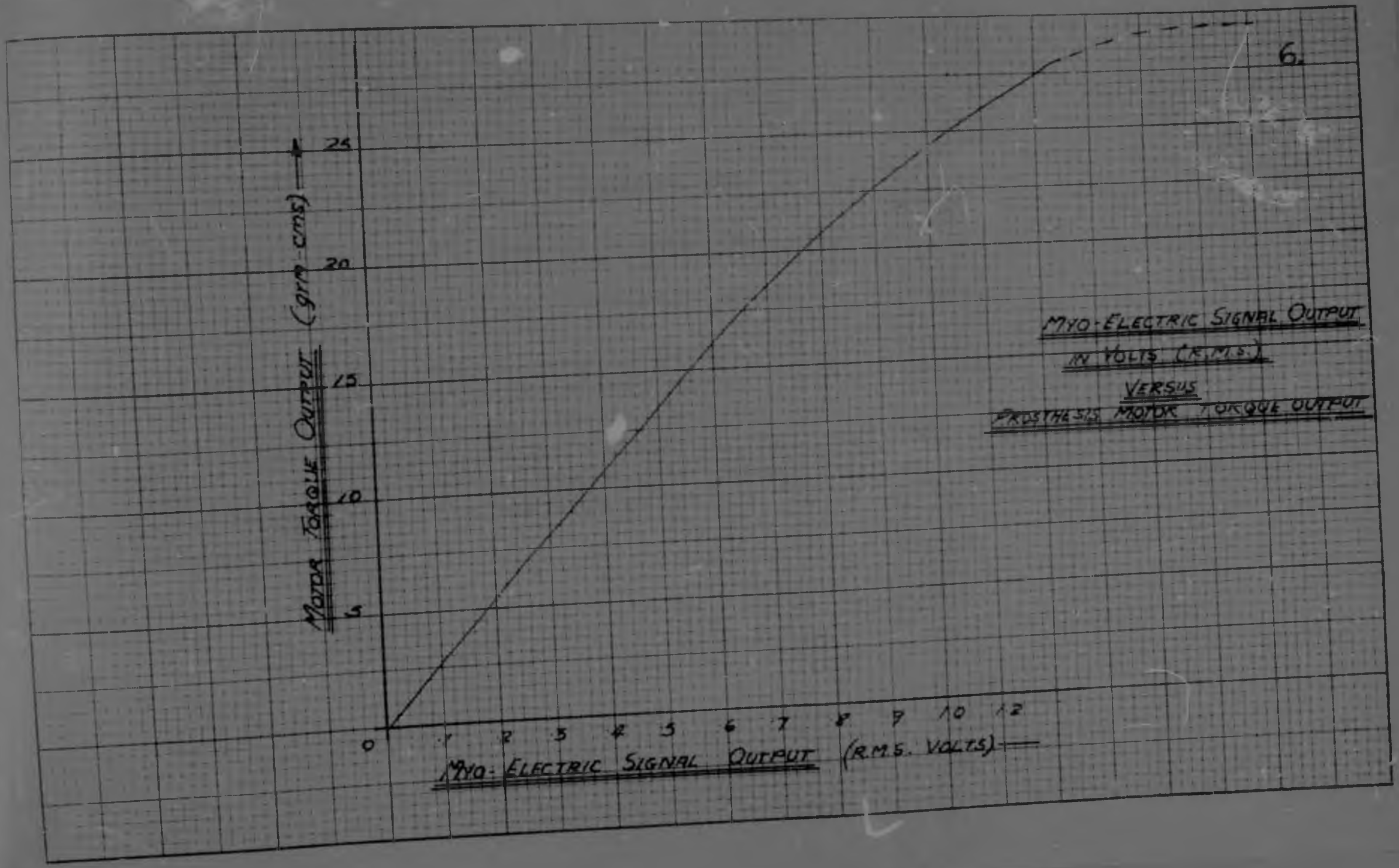


5.

MYO-ELECTRIC SIGNAL OUTPUT IN VOLTS (R.M.S.) VERSUS PROSTHESIS MOTOR SPEED.



6.


MYO-ELECTRIC SIGNAL OUTPUT
IN VOLTS (R.M.S.)
VERSUS
PROSTHESIS MOTOR TORQUE OUTPUT
MOTOR TORQUE OUTPUT (gram-cms)
25
20
15
10
5
0
1
2
3
4
5
6
7
8
9
10
12
MYO-ELECTRIC SIGNAL OUTPUT (R.M.S. VOLTS)

APPENDIX D.

PREHENSION MOTOR CHARACTERISTICS.

Graph 3. Motor speed versus armature voltage.

Graph 4. Motor torque output versus armature current.

Graph 5. Myo-electric signal output versus motor speed.

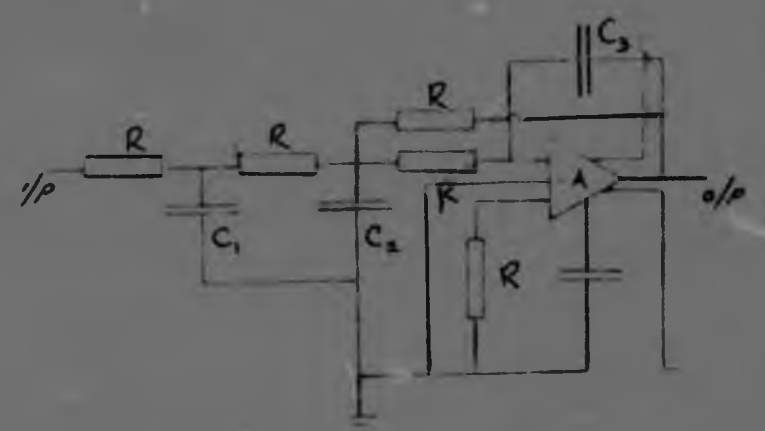
Graph 6. Myo-electric signal output versus motor torque.

From graph 5 it is seen that maximum speed is obtainable at a myo-electric voltage output of .46 volts R.M.S. Maximum torque is available at approximately three times the myo-electric voltage output as that necessary for maximum speed (graph 6). The myo-electric voltage output was measured at the base of the transistor of the final voltage amplifier stage.

APPENDIX F.

LOW PASS ACTIVE FILTER DESIGN AS USED IN THE ORTHESIS AMPLIFIERS.

A third order Chebishev filter with a $\pm$ 3 dB ripple is used in the amplifier to exclude the effect of high frequency noise generated by the orthesis motor. (c.f. Chapter 6). A table of capacitor values normalised with respect to resistor choice has been published by Foster (1965) for use in a Rauch filter configuration.



RAUCH TYPE FILTER.

The relationship between P and C_n (where n = 1,2 --- n, depending on the order of the filter) is

$$C = \frac{C_n}{Rw_o}$$ where w_o = Cut off frequency.

Choosing R = 10kΩ and choosing the cuttoff frequency to be at 400 c.p.s. then

$$w_o = 2\pi f = 2\pi.400$$
$$= 2500 \text{ radians/sec.}$$

TABLE VI

NORMALIZED CAPACITOR VALUES FOR ±3 dB RIPPLE CHEBYSHEV FILTER

N	C_1	C_2	C_3	C_4	C_5	C_6	C_7	C_8	C_9
1	1.00238								
2	4.65188	0.30365							
3	6.81594	9.87459	0.11858						
4	17.61184	0.06287	7.29501	0.69946					
5	11.29701	27.22941	0.03913	10.44517	0.25395				
6	39.23620	0.02669	14.36163	0.13344	10.51340	1.07107			
7	15.83205	53.14343	0.01944	19.02057	0.08382	13.16263	0.37195		
8	69.51525	0.01477	24.41049	0.05823	16.31038	0.19106	13.82731	1.43794	
9	20.37988	87.55034	0.01165	30.52752	0.04312	19.92561	0.11869	16.24343	0.48615

NORMALISED CAPACITOR VALUES FOR ± 3dB RIPPLE CHEBYSHEV FILTER.

For a third order filter (n=3) the values of C_1, C_2 and C_3 from the above table are

$$C_1 = \frac{6.81594}{F_c} = \frac{6.81594}{2500 \times 10^4}$$

$$= \frac{6.81594}{2.5 \times 10^7} = .273\mu f.$$

$$C_2 = \frac{9.875}{2.5 \times 10^7} = .394\mu f.$$

and

$$C_3 = \frac{0.1186}{2.5 \times 10^7} = .0474\mu f.$$

The capacitors C_1, C_2 and C_3 were made up from standard mylar capacitors and it was found that even though the values of capacitance were not exact the low pass filter operated satisfactorily (Graph 2, Appendix 'C').

REFERENCE: Foster Edward J. (1965)
"Active Low Pass Filter Design"
I.E.E.E. Transactions on Audio, Vol. AU.-13,5.

APPENDIX F.

RELATIVE RUNNING COSTS OF PNEUMATICALLY AND ELECTRICALLY POWERED PROSTHESES.

Assuming that one 11b. cylinder of CO_2 gas is used daily for powering the pneumatic prosthesis, the daily cost of filling the cylinder is 50 cents. The average yearly cost is then at least R180. The yearly cost of battery recharging is minimal and is hardly noticeable on a normal monthly electricity bill.

It is necessary to replace the storage batteries once yearly at a cost of approximately R30. This is still much less than the cost associated with the pneumatic system. The initial cost of the pneumatic cylinder is high, each cylinder costing about R9.00 excluding the regulator required to reduce the stored gas pressure from ± 1000lbs/sq. inch to a working pressure of 80lbs/sq. inch. The total cost of one cylinder plus regulator is R22.50.

Author Ruch C
Name of thesis Myo-electric control of prostheses for mid-forearm amputees and for orthotic hand splints for quadruplegics 1968

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