PROGRES WITH THE BRITISH MYOELECTRIC HAND

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Introduction

The use of EMG signals for on/off control of a prosthesis was first described by Battye[1] et al., 1955 and has since been successfully applied clinically by Kobrinski[2] et al.

The EMG signal does on the other hand allow the possibility of proportional control, and it was to investigate the difficulties of proportional control in use on patients that the British hand was designed. The nature of the surface EMG signal gives rise to several problems when attempts are made to use it for proportional control. These problems were discussed at the previous ETAN conference in Opatija[3,4] and several principles were suggested as to how to overcome them. The design of the present hand closely follows these principles so that they will only be briefly discussed below.

Problems in Using EMG for Corntrol and Some Methods of Overcoming Them

These may be divided into two. The first is the presence of crosstalk—apparent activity in one muscle when the patient attempts to perform an action involving a distant or even opposing musle. Crosstalk may be due to real but inadvertent activity in the adjacent muscle, sometimes referred to as physiological crosstalk. This may of course be eliminated by careful training, but even if this is successful, some real or electrical crosstalk always remains which is due to conduction of signals through the tissues. Electrical crosstalk from opposing muscles in the forearm may be as low as 5 — 10%.

The method used to eliminate crosstalk in the current British hand is to take signals from two opposing groups. After rectification and smoothing to obtain a D.C. level representing the intensity of the A.C. signal, the two signals are subtracted from one another in a differential amplifier. If electrodes are well placed, a muscle will always produce a stronger signal through electrodes on its surface than through the electrodes placed over its antagonist. In this way therefore the relative magnitudes of the two signals determine whether the output of the amplifier is positive or negative, and this in its turn the direction in which action is produced in the prosthesis.

The second main problem is that of smoothing the rectified signal. A simple C. R. network is used for this and a time constant must be chosen which gives a reasonably brisk response to changes in signal level; 100 m.sec. is used in this systems. Ufortunately random variations in the smoothed signal occur even when steady effort is maintained, and this is true even if considerably longer time constants are used. Other averaging methods have been suggested[5,6], but none produce a solution to this problem.

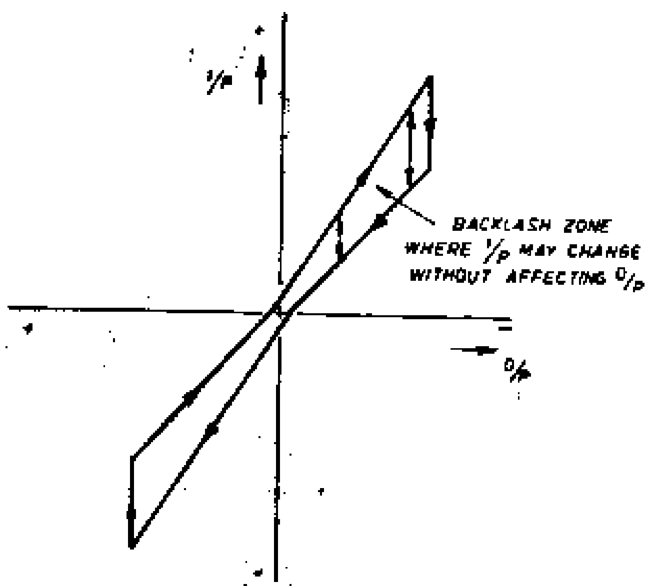

Figure 1. Transfer function of prototype hand. I/P is a positive or negative signal depending on which muscle group is producing more activity. O/P is force either opening of closing the hand

In the prototype hand described by Bottomley and Cowell[7] a backlash technique was used to eliminate these random variations. Their amplitude seems to vary with the intensity of the signal (i.e., with any steady effort there is a variation of about 20% in the averaged EMG level). The prototype system, therefore, was arranged as a closed-loop force servo in which the error signal was ignored unless above a certain minimum. Since EMG intensity is proportional to force, this minimum was arranged to be increased as O/P force increased. Figure

1 shows the transfer function of the system working in an isometric situation (i.e., exerting force with no movement).

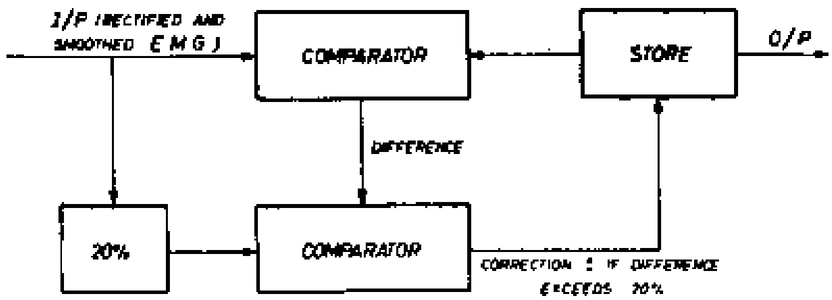

Figure 2. Block diagram of the function of the autogenic backlesh unit

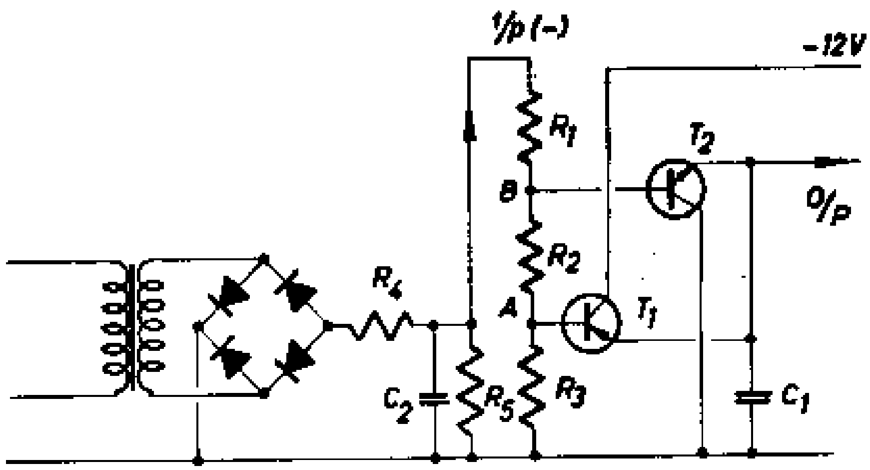

Figure 3. Circuit of the autogenic backlash unit. If R_1, R_2 and R_3 are regarded as a potential divider the capocitor C_1 will charge up to the voltage at point A, through transistor T_1 but will not discharge until point B has dropped to this level when transistor T_2 will turn on. Percentage of backlash may be changed by adjusting the value of R_2 since this affects the relation between voltages at A and B

In the production system a slightly different technique is used as described by Bottomley[8]. It has been called an »autogenic backlash generator« and consists in effect of a nonlinear filter in which O/P signal does not change unless I/P signal — O/P signal is greater than a fixed percentage of I/P signal. Figure 2 shows a block diagram of its operation, Figure 3 the circuit used, Figure 4 the transfer function of such a circuit, and Figure 5 the effect on a sample of rectified and smoothed EMG.

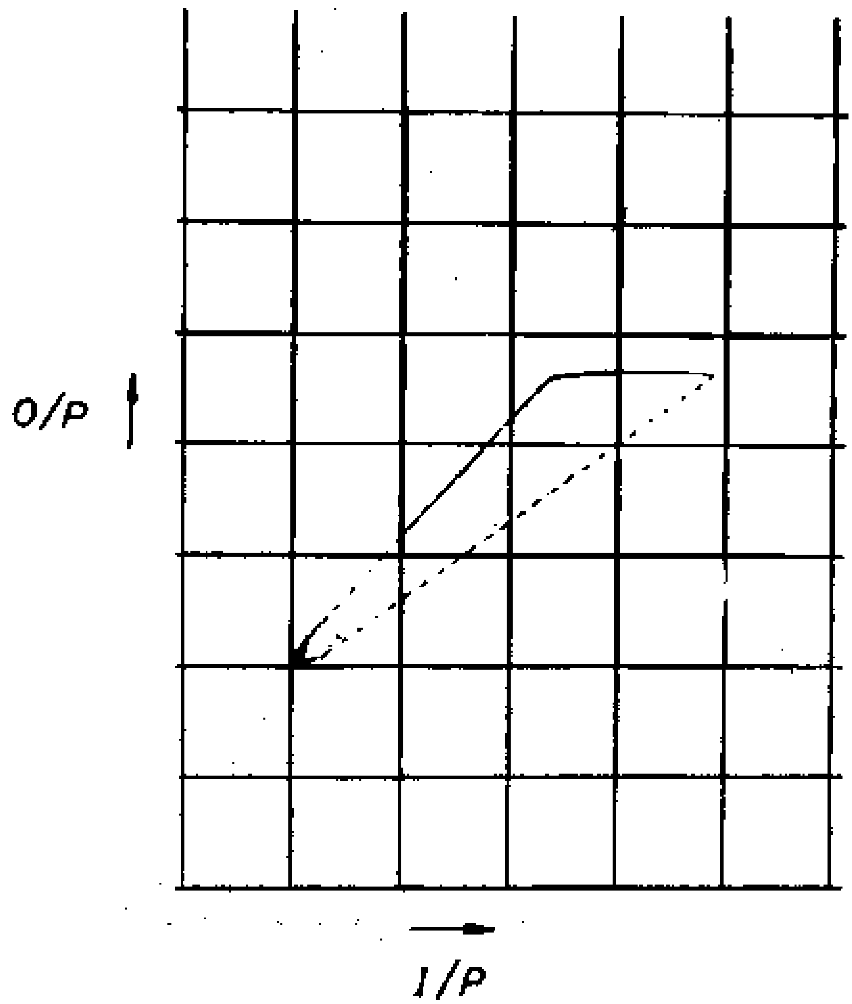

Figure 4. Transfer function of the circuit shown on Figure 3

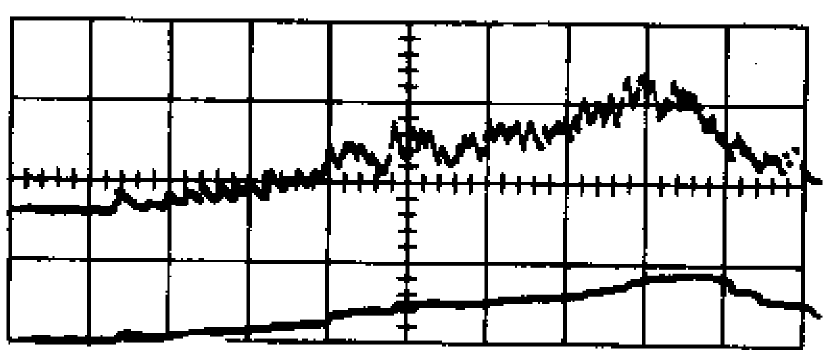

Figure 5. A sample of rectified and smoothed EMG before (upper trace) and after (lower trace) backlash is applied. Scale: 500 ms/div

Design of the Hand

Although many powered prosthetic devices use pneumatic motors and cylinders of liquid CO_2, it was decided to use an electric motor, since EMG control necessitates the use of transistors and hence the presence of a battery. It would obviously be an inconvenience to the patient to have to carry two power sources. The choice of prehension for our initial trials is an almost inevitable consequence of this decision. Hands are very seldom required to do real work. Either they are moved towards the object exerting no force, or once it has been grasped they exert force with no movement. In both cases the product (force times distance = work) is zero. Provided an electric motor is followed by a nonreversible gear and arrangements are made for it to be turned off when the required force is attained it is economical in both these situations. As stated above, muscle produce tension in proportion to the signal reaching them. It was decided therefore to make a hand which would exert force in proportion to the EMG signal. This may be achieved by inserting a force transducer in the drive train beyond the nonreversible gear and using its output as the feedback signal in a closed loop servo (see block diagram, Fig. 7). This will also turn off the motor when the required force is attained, since in this situation the error signal will be zero. To provide control when the hand moves freely and before an object is grasped, a second feedback loop is arranged using velocity feedback. It was arbitrarily decided to arrange that full velocity should be produced by an EMG intensity that will give 1/3 maximum force. This seems to work well in practice.

The Production Version

The prototype hand was designed by the author in co-operation with the Medical Electronics Department of St. Thomas' Hospital. We had, however, no facilities available for miniaturization and production so that it was decided to get the help of a larger organization: the United Kingdom Atomic Energy Authority who are responsible for the production version. The Army Prosthetics Research Laboratory hand was used and fitted with a size eight permanent magnet D.C. servo motor (Rank-Pullin Controls Ltd.) followed by a spur gear train and a critically nonreversible lead screw and nut. A beam attached to the nut is arranged to pull the two moving fingers of the hand against the thumb. Semiconductor strain gauges are cemented to the beam and give the force signal required for control (Fig. 6). The velocity signal is derived from the back E.M.F. of the motor. Thirteen D.E.A.C., Type 900 D (900 m.A.H.) cylindrical nickel cadmium cells are used and a full charge seems to last about 2½ days in continuous use. A maximum force of 9 lbs. is available at the finger tips and the fingers close in .8 seconds, at full speed. Position-operated limit switches restrict the fingers and prevent jamming. Fig. 7 shows a block diagram of the system.

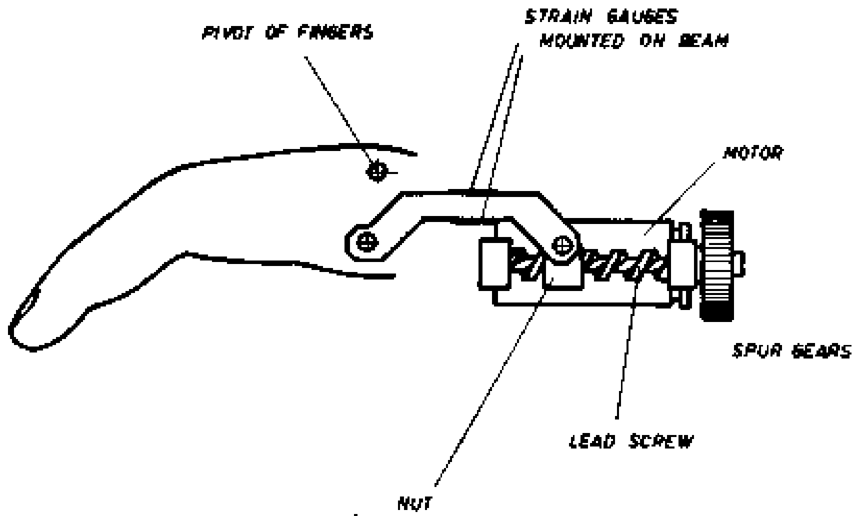

Figure 6. Arrangement of strain gauges in the production hand to give a signal proportional to output force

Three provisions have been introduced which have not so far been described and are to ensure that battery current is not wasted.

1. Clearly while the patient produces a signal and the motor is capable of producing a force to match it, the error signal can be restored to zero and the motor current turned off. If, however, he exceeds this level the motor will then receive stall current continuously and to no useful effect. To prevent this the »EMG clamp« (8 in Fig. 7) is introduced which limits the input to a level just below the maximum the motor can achieve. Having a nonreversible gear, the hand is potentially capable of producing passive forces greater than the maximum of the motor—as in lifting heavy objects. In order to arrange that the motor does not then open the hand it is necessary to introduce the »Force clamp« (13 in Fig. 7) at the same level. In practice this means that within the dynamic range of the system output force is proportional to effort, but that outside it the hand remains locked.
2. If the nonlinear filter succeeds in getting rid of wobble on the input signal, it could happen that the system would see an error signal sufficiently large to consume current but insufficient to cause the motor to produce the necessary correction. The function of the second summing junction (17 in Fig. 7) is to detect the level at which successful correction will be produced by the motor. Below this level the drive amplifier is disconnected by the Schmidt trigger (19 in Fig. 7). In practice this means choosing a small error signal which will give full stall torque from the motor. The velocity feedback is arranged so that this occurs at very low speed.

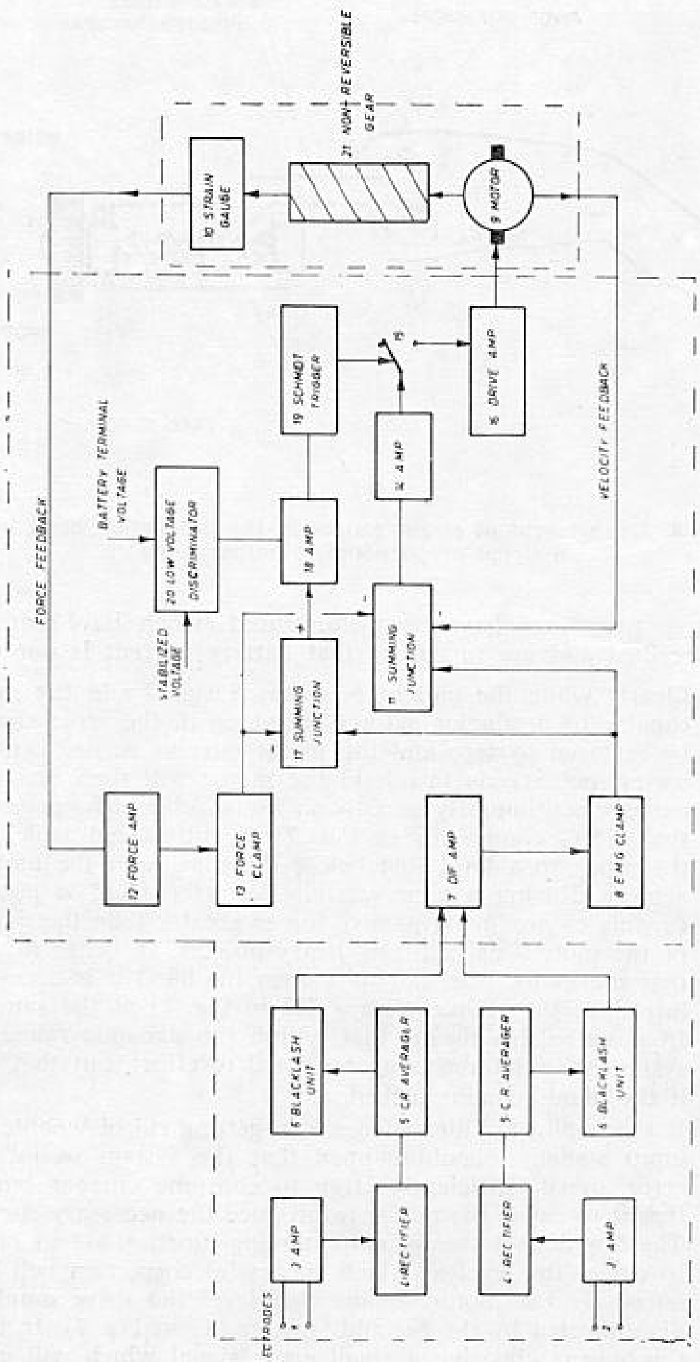

Figure 7. Block diagram of the production hand

3. Even if nickel cadmium batteries are used, there comes a time when a battery is nearing the end of its discharge cycle when voltage begins to drop. Under these conditions the »EMG clamp« will begin to fail in its job of protecting the motor from receiving stall current when it fails to achieve the appropriate force. To prevent this happening the low voltage discriminator (20 in Fig. 7) is introduced, which quenches the drive circuit at this point.

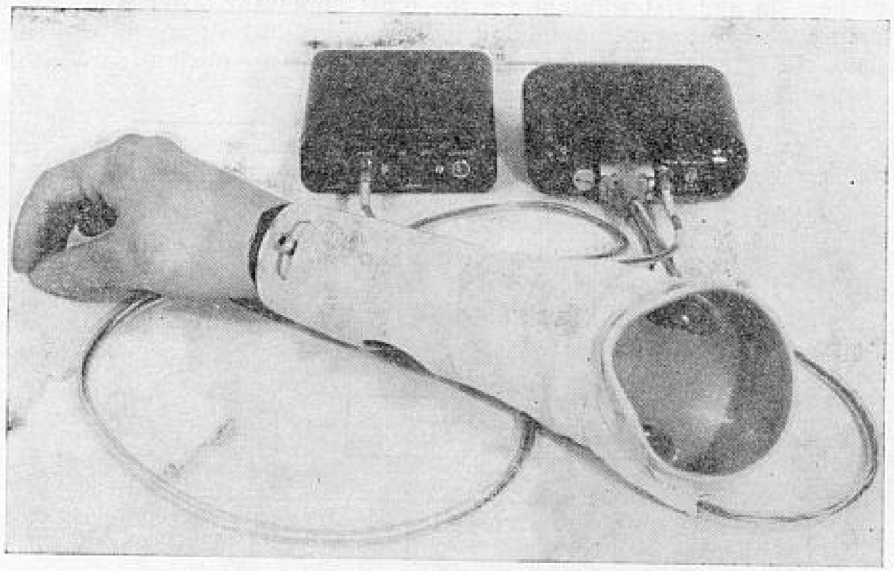

Figure 8. The production hand with a Münster socket fitted with electrodes

Figure 8 shows a photograph of the production hand, and the electrodes fitted into a Münster type total-contact socket. Electrodes are of pure silver and are lightly sprung onto the skin surface using a silicone rubber grommet pictured in Figure 9. Only six patients have so far been fitted, but their reactions to the prosthesis have been favourable.

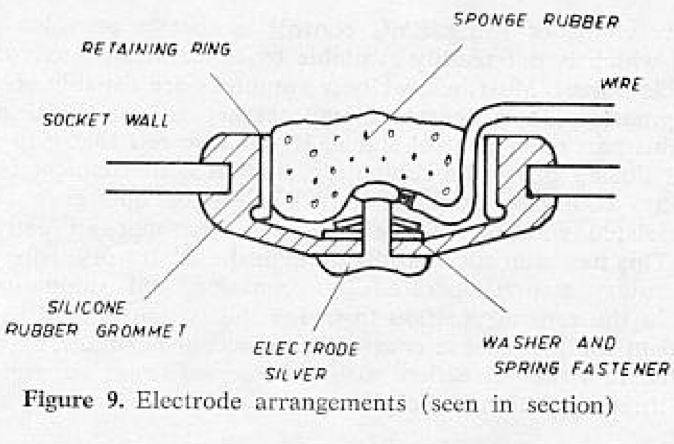

Figure 9. Electrode arrangements (seen in section)

Additional Movements

With careful training crosstalk may be reduced to a sufficiently low level to enable on/off control of 2 or even 3 low independent movements in below-elbow amputees. It is also possible by special techniques such as described by Dr. Beeker of Delft (q. v.) to achieve proportional control of at least two movements with EMG control in the below-elbow amputee. If two movement are to be provided probably prehension and rist rotation are the most important. After some preliminary work towards obtaining independent EMG signals to control these two movements it was decided to adopt a more obvious approach.

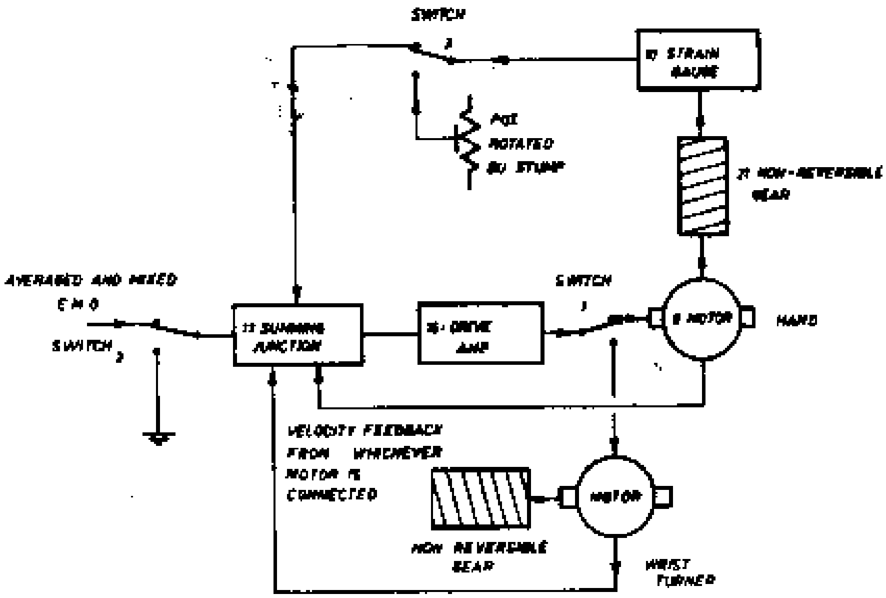

Figure 10. Simplified version of Figure 7 showing how the same control circuit may be used to operate prehension from EMG signals and wrist rotation from a mechanical input

The only virtue of using EMG control is that it provides a control channel which is not readily available by other means (except perhaps by muscle bulge). Most below-elbow amputees are capable of pronation and supination. It is therefore unnecessary to use EMG signals to derive this particular control signal. It was also felt that grip is seldom changed during pronation/supination so that a mechanical rotation of the stump could be used to control a motor and gear box giving power-assisted rotation of the prosthesis, without an extra control circuit. This has been successfully done in the laboratory using a printed circuit rotary switch operated by pronation and supination of the stump. In the central position in which the switch is held by a detent mechanism the prehension control is connected normally. If the stump is rotated in either direction with a force sufficient to overcome the detent, three operations take place serially.

1. The motor used for prehension is disconnected (switch 1, Fig. 10).
2. The EMG signal is switched off (switch 2, Fig. 10) and a potentiometer is substituted for the strain gauge bridge normally used for force feedback (switch 3, Fig. 10).
3. The motor drive circuit is connected to a motor and gear box for wrist rotation by switch 1.

Increased effort to rotate the stump then progressively gives increased velocity of rotation of the prosthesis in the appropriate direction, by moving the potentiometer referred to above.

The extremes of pronation and supination some interference is produced in the EMG channel but this is unimportant since the prehension drive circuit is then switched off and the hand locked. Using clossed-loop velocity control rather than a position servo, continuous rotation of the wrist is possible, though opinions differ as to whether this is desirable. Since the same drive circuit is used, it was necessary to use an identical motor for this movement and to provide force-operated limit switches. Unexpectedly the motor gives very geneous output for wrist rotation.

Higher Level Amputees

For the above-elbow amputee it would be possible to use biceps and triceps for operating a hand. However, it is unlikely that more than one channel will be available from muscles in the upper arm, and probably the key to the use of EMG in the future is its skillful combination with mechanical control of external power and with orthodox cable control. If this is so, it might be preferable to use EMG from biceps and triceps to control elbow flexion or wrist rotation. To achieve control of many movements in the shoulder disarticulation amputee probably an implanted electrode would be necessary, in which case signals may be obtained from single motor units, even several independently from the same pair of electrodes as described by Basmajian[8].

References

1. Battye, C. K., Nightingale, A., and Whillis, J.: The Use of Myoelectric Currents in the Operation of Prostheses. J. Bone & Jt. Surg., 37B, pp 506—510, 1955.

2. Kobrinski, A. E. et al: Problems of Bioelectrical Control. First IFAC Conference, Moscow, pp 1119—1122, 1960.

3. Nightingale, A. and Bottomley, A. H.: Myoelectric Control of Muscle Substitutes. Proceedings of the Symposium on Automatic Control in Prostetics, Belgrade, Yugoslav Committee for Electronics and Automation, 1962.

4. Bottomley, A. H.: Working Model of a Myoelectric Control System; ibid, 1962.

5. Weltman, A. and Lyman J.: The Effects of Electronic Transformations on the Pattern of Myoelectric Activity During Arm Movements. Biotechnology Laboratory Report No. 6, University of California, Los Angeles, 1960.

6. Scott, R. N. and Thompson, G. B.: Orthotic Systems Research. Progress Report No. 3, University of New Brunswick, Department of Electrical Engineering Research, Report 64.2, June 1964.

7. Bottomley, A. H. and Cowell, T. K.: An Artificial Hand Controlled by the Nerves. New Scientist, 21, pp 668—671, 1963.

8. Bottomley, A. H.: Amplifier Design and Signal Processing for Myoelectric Control of Powered Prostheses. Sixth International Conference on Medical Electronics and Biological Engineering, Tokyo, 1965.

9. Basmajian, J. V.: Conscious Control of Single Nerve Cells. New Scientist, 20, pp 62, 1963.