A NEW VERSATILE MYOELECTRIC CONTROL UNIT FOR CLINICAL APPLICATION

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Introduction

The use of myoelectrically controlled prostheses has been spreading throughout the world. Although the first myoelectric prosthesis was produced by Reiter /1/, as early as 1948, clinical application of myoelectric control only became possible with the availability of semiconductor technology. Present-day refinements of this technology in the form of micropower integrated circuits make possible an even more sophisticated and versatile artificial limb control /6/, and experience with presently exsiting systems has provided information which points the way to further developments.

Past Experiences

Following clinical experimentation with the Soviet myoelectric hand, it was decided to produce a control system with emphasis on improved performance and miniaturization. An on-off type unit was completed in 1966 for which size and weight reduction was achieved by using integrated circuit technology /2/. This on-off control device has since been applied routinely at this Institute.

Analog Control

In 1968, continued design efforts led to the development of our first open loop proportional control unit. For this unit any increase of the rectified myoelectric signal produces a proportional increase in the DC power available to the motor. Clinical results obtained with this control unit show that the time delay between the command input signal and the actuator movement has been reduced considerably when compared to the on-off system. This reduction has permitted a more dynamic response of the controlled device, effecting a closer physiological link between the user and his prosthesis. Furthermore, it is now possible for the user to manipulate fragile objects and to perform fine work with pliers and tweezers.

Analog Control Drawbacks

One of the drawbacks is the high energy dissipation in the output transistor. A second drawback relates to the following: the electrical energy applied to a motor is transformed into mechanical energy only when it reaches a certain percentage of the nominal energy. This percentage is a function of the load and finally, in an open loop system these latter two aspects when combined with a conscious or subconscious residual myoelectric activity could be the cause of a current drain through the motor although no displacement of the driven device is observed. Consequently, unnecessary power consumption may exist when the system operates within the neutral resting zone and the amputee would be oblivious of this situation. These inconveniences may be overcome by taking advantage of a closed loop system where the input signal would be compared continuously to a feedback signal of velocity, force, or both. Random variation in the EMG input signal and the required complexity of a closed loop system lead us to retain an open loop system, but using pulse frequency modulation instead of an analog system.

Control Through Pulse Modulation

Recently, Ulen and Wager /4/ reported that motor control through pulse modulation is not more efficient than through analog control. Nevertheless, this technique was thought to offer some advantages when applied to prostheses or orthoses controlled by an open loop system.

The advantages are:

(1) Possibility of adjusting the minimal pulse width applied to the motor so that to each pulse corresponds to a certain motor rotation. This eliminates the second drawback of the analog system. Moreover, the discrete tremor generated by the pulses informs the patient that his prosthesis is operating. However, the acoustic level of the tremor is minimal and not noticeable to the ear.

(2) Output transistors operate only at saturation level which corresponds to a minimum energy dissipation in these transistors.

(3) Possibility of higher power output without sacrificing the miniaturisation.

Design criteria

Before listing the main characteristics of this new control unit, it should be emphasized that all current prosthetic and orthotic devices require the patient to accept some inconveniences and discomfort. To offset these considerations, the device must significantly improve his function or appearance, or both. Some factors which generally contribute to the rejection of a myoelectric prosthesis by the patient are: poor prosthetic performance, unreliability, difficulties in obtaining service, battery maintenance, excessive weight or dimensions, and high cost. The prosthesis may also reject a device on account of its complexity and difficulties with fitting or repairs.

Considering the above and trawing upon past expirience we have included the following characteristics in our most recent control system (Fig. 1.):

(1) Plug in type modular construction of disposable units which could easily be assembled or repaired by any qualified prosthetist.

(2) Electrodes and myoelectric amplifier encapsulated in a single block; this unit is also disposable, and, in event of failure, may be replaced easily by a new one at reasonable cost.

(3) Reduced overall dimension of the assembly so that it can be incorporated easily into the laminated socket of any type of prosthesis (Fig.2).

(4) Power supplied by a single battery.

(5) Negligible standby power requirements. This factor is very important since it is in effect at rest as well as during activity.

(6) Versatility and flexibility of utilization. The unit can control proportionally an electric prosthesis, and with the aid of pulsed electromagnetic valves, a hydraulic or pneumatic prosthesis /5/. Moreover, it accomodates either myoelectric control or a displacement transducer which could be adapted to some anatomical movement. Using a transducer, the input signal is directly fed

into the pulse generator at TP,1 (Fig.3).

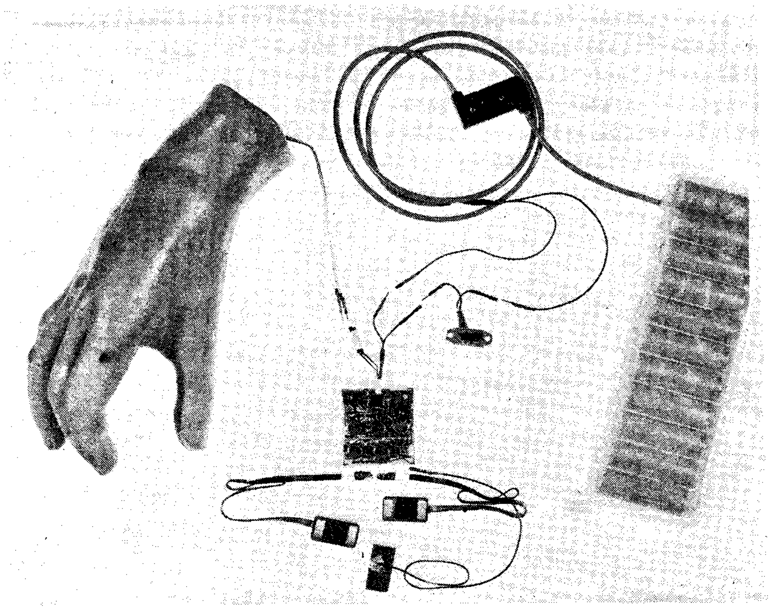

Fig. 1. Myoelectric proportional control system kit assembly.

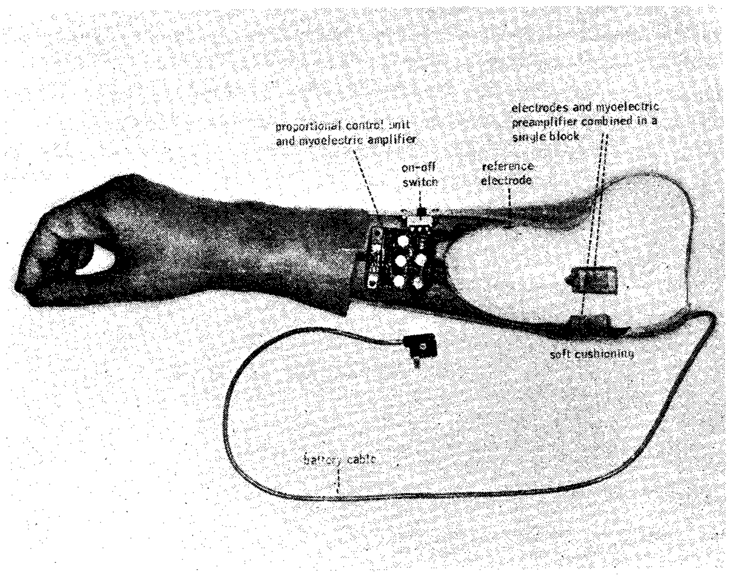

Fig. 2. Sectional view of a myoelectrically controlled prosthesis with a Munster-type socket for a below elbow amputation.

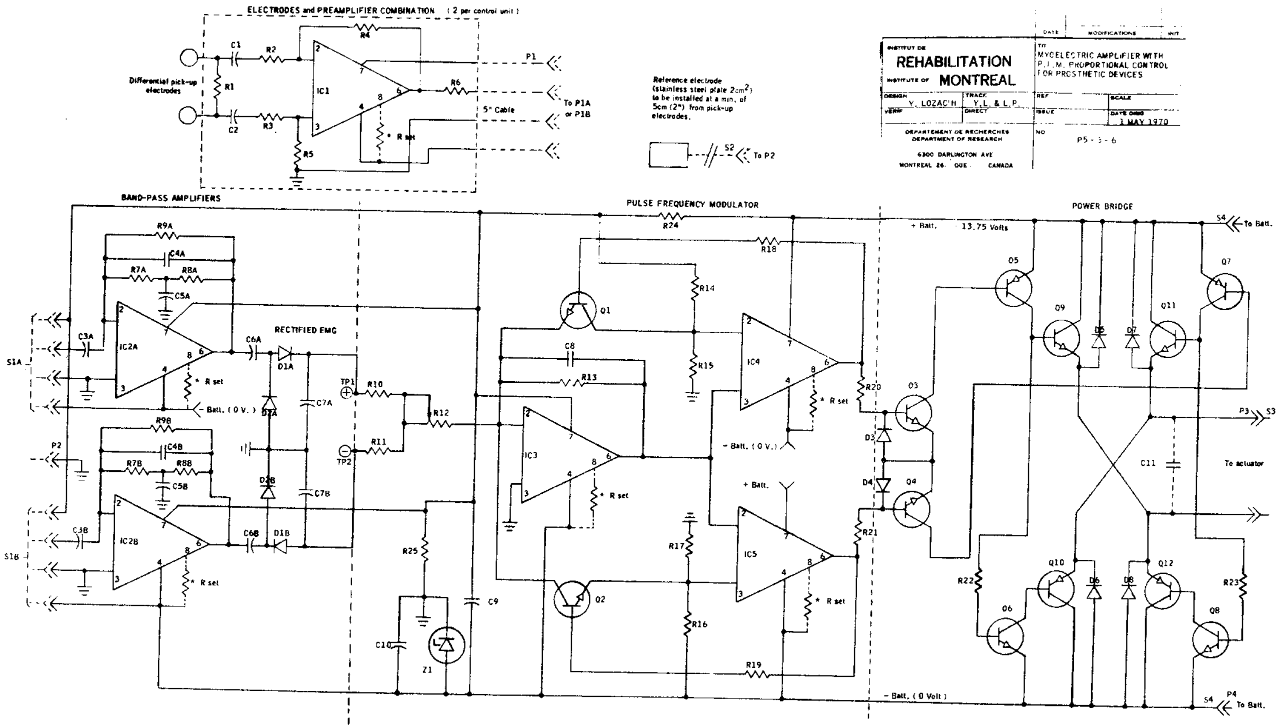

Fig. 3. Complete diagram of the myoelectric proportional control system. The total drain is 0.9 mA. All IC's are micro-power operational amplifiers (UC4250).

Control unit description

Two types of proportional control using pulse technique have been designed and tested in our laboratory. The first operates at fixed frequency while the pulse width is varied proportionally to the electromyographic signal. The second is a pulse frequency modulation type /3/. Both types gave equivalent results; however, the latter was selected for clinical application since it was designed around a single type integrated circuit allowing for substantial saving in cost with small-number production planning. In this type of control, the velocity and the force are functions of the width and frequency of the electrical impulses applied to the motor. The pulse frequency is itself proportional to the average rectified electromyographic signal which implies that the number of applied pulses per second is directly proportional to the voluntary muscle contraction.

Electrodes

Stainless-steel surface electrodes and the myoelectric preamplifier have been integrated into a single unit (Fig. 1). Using this arrangement, a better separation of the signal from the background interferences and indirectly a more dynamic response of the controlled device have been obtained.

The preamplifier is an input differential stage type with a voltage gain of one hundred. Its frequency response is limited by the input capacitors and the amplifier´s own characteristics (UC 4250).

Band-Pass Amplifier

The second myoelectric amplification stage is capacitively coupled to the first stage. It is a narrow band-pass amplifier ranging from 100 to 1000 Hz with a maximum voltage gain for frequencies between 400 and 500 Hz. The output of this stage is connected to a rectification system, positive for one channel and negative for the other. After filtering and smoothing, the sum of these positive and negative values is then fed to the input of a variable pulse frequency generator (at point TP,1). Filter-time constant has been minimized in order to obtain a faster response of the controlled device.

Detailed Operation of the Pulse Generator

When a signal E_i is fed to the integrator's input (Fig. 2), the resultant at the output is a variation proportional to the time integral of the input signal. The ouput signal E_o is then compared to the reference voltage V_r of the same polarity. The output of IC 4 or IC 5 will become strongly positive as soon as the absolute value of E_o is equal to or greater than the absolute value of V_r. This will switch transistor Q1 or Q2 and start the discharge of the integration condenser C8 through R14 or R16. Thus the absolute value of input and output signals of the integrator will decrease until E_o is again less than V_r. At this point, the output of the comparator will regain its initial negative value and the discharge of C8 will be stopped. The pulse width (T) is approximately calculated according to the following formula: $T = CRx$ where $Rx = \frac{R14 . R15}{R14+R15} = \frac{R16 . R17}{R16+R17}$

The frequency (F) is determined by the following formula:

$$F \cong \frac{1}{C8Ry} \cdot \frac{Rz}{Vr} \cdot E_i$$

where $Ry = R10+R12 = R11+R12$

and $Rz = \frac{R15}{R14} = \frac{R16}{R17}$

E_i= input voltage of the integrator.

Power Stage

The power stage is bridge made up of transistors Q9, Q10, Q11, and Q12. These transistors are controlled respectively by Q5, Q6, Q7, and Q8 which themselves are controlled by Q3 and Q4. Its operation is such that for any positive pulse from the comparator's output, Q3 or Q4 becomes a conductor. If Q3 becomes a conductor, Q5, Q6, Q9 and Q10 conduct at saturation level thus directing the current in the load (motor); if the polarity of the comparator's input signal is reversed, Q3 is blocked and Q4 becomes a conductor bringing Q7, Q8, Q11 and Q12 to saturation level and, consequently, reversing the current in the load.

The diodes D5, D6, D7 and D8 are included to clamp the current spikes obtained with high inductive loads.

The power stage can deliver a current as high as 3 amperes without requiring any heat sink.

General Specifications

Input impedence of the differential preamplifier (400 Hz)	150 K ohm
Band-pass amplifier frequency range	100 - 1000 Hz
Band-pass amplifier rejection factor	-40 dB at 60 Hz
Pulse frequency rate (to be adjusted according to the motor characteristics)	0 to 100 Hz max.
Pulse width (to be adjusted according to the motor characteristics)	10 milliseconds
Battery (nickel-cadmium)	13.7 volts,500mAH
Total current drain	0.9 mA
Control unit dimensions	4.3cm X 3.9cm X 0.7
Control unit weight	20 gr
Electrodes and preamplifier combination block dimensions	2.4cm X 1.4cm X 0.9
Electrodes and preamplifier combination block weight	7 gr

References

/1/ Reiter, R.,; "Eine neue Electrokunsthand", *Grenzgebiete der Medizin*, 4 : 133, 1948.

/2/ Lozac'h, Y., Lippay, L.A., Sherman, D.E., and Gingras, G.; "Helping Hands", *Electronics*, pp. 124-131, Aug. 7, 1967.

/3/ Pavlides, T., and July, E.; "Analysis of a New Class of Pulse-frequency Modulated Feedback System", *IEEE Trans. Automatic Control*, AC-10, pp. 35-43, 1969.

/4/ Ulén, E., and Wager, M.: "Driving Systems for Orthosis-Prosthesis Application", FOA 2 rapport 1C 2477-54, Forsvarets Forskningsanstalt, Avdelning 2, 104 50 Stockholm 80, Sweden,1971.

/5/ Kwee, H.H. and Bache, H.W.; "Myoelectric Control of Pneumatic Prostheses", Annual Report, Department of Research, Rehabilitation Institute of Montreal, Vol. V, October 1969.

/6/ Lozac'h, Y., Sherman, D.E., and Gingras, G.; "A New Versatile Myoelectric Proportional Control Unit", *Bulletin of Prosthetics Research*, Fall 1971.

Bibliography

1. Bigland, B., and Lippold, O.C.J.;"The Relation Between Force, Velocity and Integrated Electrical Activity in Human Muscles", *J. Physiol.*, 123 : 214-224, 1951.

2. Nightingale, A.;""Background Noise" in Electromyography", *Phys. Med. Biol.*, 3:325-338, 1959.

3. Bottomley, A.H., Kinnier-Wilson, A.B., and Nightingale, A.; "Muscle Substitute and Myoelectric Control", *J. Brit. I.R.E.*, 26 : 439-48, 1963.

4. Sherman, D.E., Lippay, A.L., and Gingras, G.; "Prosthesis Given New Perspectives with External Power", *Hospital Management*, 100 : 44-49, Nov. 1965.

5. Godden,A.K.;"The Techniques of Myoelectric Control of Prostheses and the Prospects of this Type of Control for Thalidomide Casualties", Engineering Laboratory Report No.1, 048, 68, University of Oxford, England, Mar. 1968.

6. Gavrilović, M.M., and Marić, M.R.; "Positional Servo-mechanism Activated by Artificial Muscles", *Med. and Biol. Engng.*, 7(1): 77-82, Jan. 1969.

7. Parker, P.A., and Scott, R.N.; "Electrode Studies", University of New Brunswick, BioEngineering Institute, Progress Report no.8, pp. 23-38, March 1969.

8. Kadefors, R., Monster, A.W., and Petersen, I.; "A New Aspect of Electrode Design in myoelectric Control Systems", *8th ICMBE*, Chicago, Illinois, Session 5-5, July 20, 1969.