A NEW VERSATILE MYOELECTRIC PROPORTIONAL CONTROL UNIT

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INTRODUCTION

The realization of the first myoelectric prosthesis in 1948 by Reiter (1) led the way for a new type of control. The development of the transistor has since then considerably increased the potential application of this technique. Micropower integrated circuits should be another important step toward a more sophisticated and versatile artificial limb.

Several laboratory prototypes and commercial units exist today; however, a group of Soviet researchers produced the first commercially available bioelectic prosthesis.

Our laboratory, after having experimented clinically with the Soviet system, decided to produce its own control unit placing the emphasis on improved performance and on miniaturization. This miniaturization is essential when a complete integration of the control unit with the prosthesis is envisaged; the use of the integrated circuit made this possible. The unit was completed in 1966 and described in the article "Helping Hands" (2). An on-off mode of control was used. In this type of control the maximum power is always applied to the motor. Clinical experience obtained with the on-off mode of control has shown that it would be more desirable for the patient to be able to proportionally control the velocity and the force output of his terminal device as a function of myoelectric activity. A new control system has been designed named Analog Proportional Control, and its working principle is such that to any increase of the rectified myoelectric signal there is a corresponding proportional increase of the d.c. power available to the motor.

Clinical results obtained with this control unit have shown that the following objectives have been attained: The time delay between the command input signal and the actuator movement has been considerably reduced. This reduction has permitted a more dynamic response of the controlled devices, 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 the help of pliers or tweezers (Fig. 1).

FIGURE 1.—Manipulation of delicate objects is now possible.

One of the drawbacks of the analog system is the high energy dissipation in the output transistors. A second drawback is the fact that the electrical energy applied to a motor is transformed into mechanical energy only when this electrical energy represents a certain percentage of the nominal energy. This percentage is also a function of the load. These drawbacks when combined with a conscious or subconscious residual myoelectric activity could be the cause of a current drain through the motor although no output displacement of the controlled device is observed. Consequently, high power consumption may exist when the system operates within the neutral resting zone and the amputee would be oblivious to this situation. To overcome these inconveniences a system based on pulse frequency modulation was chosen.

Before describing this unit, it may be in order to indicate the reasons which dictated the choice of the major criteria used in this design. All prosthetic and orthotic devices require the patient to accept some inconveniences and discomfort. In return, the device must significantly improve his functional freedom. Some factors which generally contribute to the rejection of a myoelectric prosthesis by the patient are: poor performance of the prosthesis, unreliability, unavailable servicing, battery maintenance, excessive weight or dimensions, and high cost. The prosthetist may also reject a device for various reasons such as complexity and difficulties in fitting or repairs. In view of these possibilities, the following characteristics were included in our last unit:

1. Modular construction which could easily be assembled by any qualified prosthetist.
2. Electrodes and myoelectric preamplifier are encapsulated in one block. This unit is now disposable and can be replaced by a new one at a reasonable cost. This will permit the prosthetist to facilitate repairs in a much shorter time and at a predetermined cost.

3. The overall dimensions are such that the assembly can be easily incorporated in a laminated socket of any type of prosthesis (Table 1).

TABLE 1.—*General Specifications*

Input impedance 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 250 or 500 mAH
Total current drain	0.9 mA
Control unit dimensions	4.3 cm. X 3.9 cm. X 0.75 cm.
Control unit weight	20 gr.
Electrodes and preamplifier combination block dimensions	2.4 cm. X 1.4 cm. X 0.9 cm.
Electrodes and preamplifier combination block weight	7 gr.

4. Power is supplied from a single battery; at present this appears to be the most convenient solution for reasons of simplicity, cost, and maintenance.
5. Negligible standby power of the control unit. This factor is important since consumption of energy should be calculated for an entire day.
6. Versatility of utilization: this unit can proportionally control an electrical prosthesis and also a hydraulic or pneumatic prosthesis with the aid of pulsed electromagnetic valves. Furthermore, this unit accommodates both a myoelectric control or a displacement transducer adapted to some anatomical movement. Using a transducer, the input signal is directly connected to the pulse generator at TP1 (see Fig. 2).
7. Proportional control by pulse modulation: the first characteristic of this type of control is the possibility of adjusting the minimal pulse width applied to the motor, so that each of the pulses corresponds to a certain motor rotation thereby eliminating the second inherent problem of the analog system. Moreover, the discrete tremor generated by the pulses informs the patient that his prosthesis is working. However, the acoustic level of the

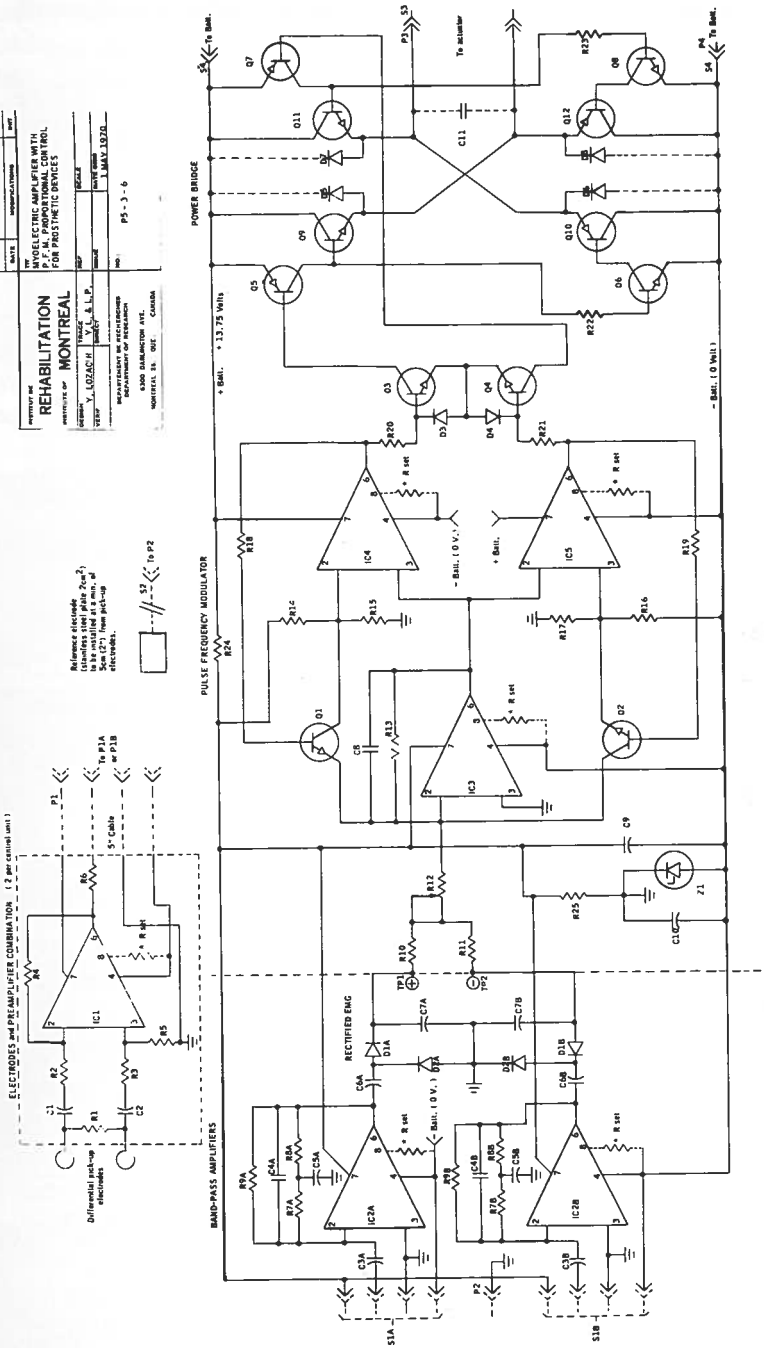

FIGURE 2.—Complete diagram of the myoelectric proportional control system. The total current drain is 0.9 mA. All IC's are micropower operational amplifiers (UC 4250).

tremor remains weak and is not noticeable to the ear. Secondly, when control is obtained through pulse modulation, the output transistors operate only at saturation level, the level at which the energy (I^2R) dissipated in the transistors is minimal. This allows economy of battery power together with higher efficiency and the possibility of controlling higher power output without sacrificing the miniaturization.

By comparison, using the same terminal device, the prehension force obtained is approximately 40 percent higher than the prehension obtained with the analog system.

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.

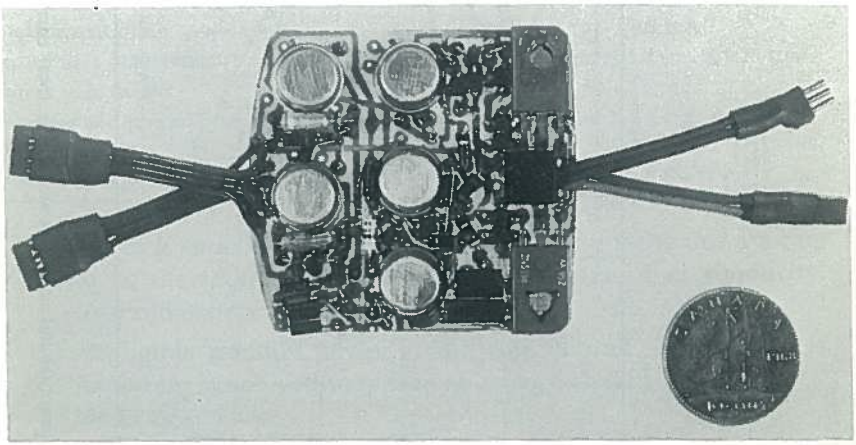

FIGURE 3.—Proportional control unit by pulse frequency modulation. This unit includes two band-pass amplifiers (channels A and B), the pulse generator, and the output stage.

Both types gave equivalent results; however, the latter was selected for clinical applications due to the fact that it was designed around a single type of integrated circuit (Fig. 3). This is quite important since it allows a substantial cost decrease when the production planning is based on a small series.

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.

GENERAL DESCRIPTION

Electrodes

The stainless-steel surface electrodes and the myoelectric preamplifier have been integrated into a single unit (see Fig. 4). 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.

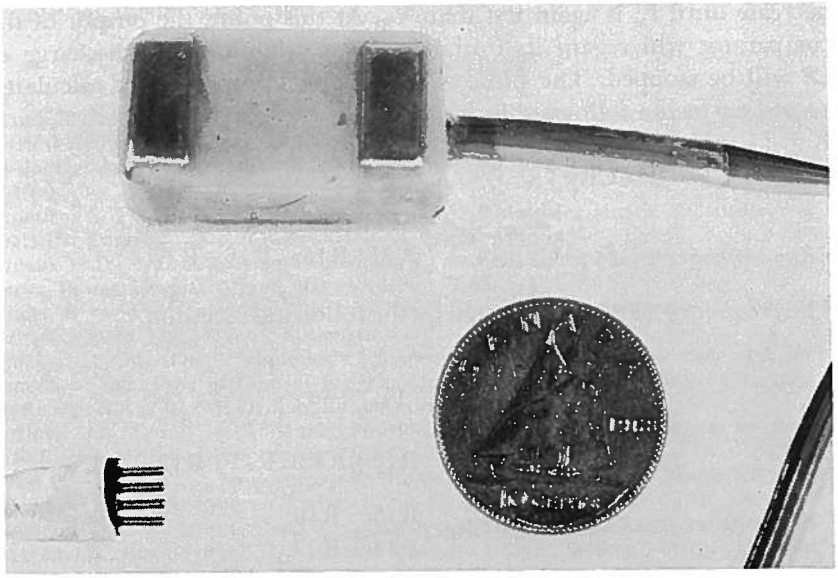

FIGURE 4.—Stainless-steel electrodes and myoelectric preamplifier integrated in one unit.

The preamplifier is an input differential stage 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 stage of the myoelectric amplification is capacitively coupled to the first stage. It is a narrow band-pass amplifier ranging from 100 to 1000 cycles with a maximum voltage gain for frequencies between 400 and 500 cycles. 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 TP1). Filter-time constant has been minimized in order to obtain a faster response of the controlled device.

Detailed Operation of the Pulse Generator (Fig. 2)

When a signal E_i is fed to the integrator's input, the resultant at the output is a variation proportional to the time integral of the input signal. The output signal E_o is then compared to the reference voltage V_R of the same polarity. The output of IC4 or IC5 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:

$$T \cong CRx$$

$$\text{where } Rx = \frac{R14.R15}{R14+R15} = \frac{R16.R17}{R16+R17}$$

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

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

$$\text{where } Ry = R10+R12 = R11+R12$$

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

E_i=input voltage of the integrator.

Power Stage (Fig. 2)

The power stage is a 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.

CONCLUSION

Following 1 year of clinical evaluation, the results obtained are very encouraging. We believe that this present system can control an artificial hand which at its maximum velocity approaches the velocity of a normal human hand.

REFERENCES

1. Reiter, R.: Eine neue Electrokunsthand. Grenzgebiete der Medizin, 4:133, 1948.
2. Lozac'h, Y., A. L. Lippay, E. D. Sherman, and G. Gingras: Helping Hands. Electronics, pp. 124–131, Aug. 7, 1967.

GENERAL REFERENCES

Bigland, B. and O. C. J. Lippold: The Relation Between Force, Velocity and Integrated Electrical Activity in Human Muscles. J. Physiol., 123:214–224, 1951.
Nightingale, A.: "Background Noise" in Electromyography. Phys. Med. Biol., 3:325–338, 1959.
Bottomley, A. H., A. B. Kinnier-Wilson, and A. Nightingale: Muscle Substitute and Myoelectric Control. J. Brit. I.R.E., 26:439–48, 1963.
Sherman, E. D., A. L. Lippay, and G. Gingras: Prosthesis Given New Perspectives with External Power. Hospital Mgmt., 100:44–49, Nov. 1965.
Pavlides, T. and E. July: Analysis of a New Class of Pulse-frequency Modulated Feedback System. IEEE Trans. Automatic Control. AC–10, pp. 35–43, 1965.
Godden, Alex 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.
Gavrilovic, M. M. and M. R. Maric: Positional Servo-mechanism Activated by Artificial Muscles. Med. and Biol. Engng., 7(1):77–82, Jan. 1969.
Parker, P. A. and R. N. Scott: Electrode Studies. University of New Brunswick, Bio-Engineering Institute, Progress Report No. 8, pp. 23–38, Mar. 1969.
Kadefors, R., A. W. Monster, and I. Petersen: A New Aspect of Electrode Design in Myo-electric Control Systems, 8th ICMBE, Chicago, Illinois, Session 5–5, July 20, 1969.