

- [54] **MYOELECTRICALLY CONTROLLED PROTHESIS**
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- [22] **Filed:** Feb. 10, 1971
- [21] **Appl. No.:** 114,262

- [52] U.S. Cl.....3/1.1, 3/12.3
[51] Int. Cl.....A61f 1/00, A61f 1/06
[58] Field of Search.....3/1-1.2,
3/12-12.7; 128/418, 2.06 E

- ## [56] References Cited

UNITED STATES PATENTS

3,631,542	1/1972	Potter.....	3/1.1
3,418,661	12/1968	Allison et al.	3/1.1
3,620,208	11/1971	Higley et al.	128/2.06 E
3,628,527	12/1971	West.....	128/2.06 E
2,561,383	7/1951	Larkins et al.....	3/12.7

OTHER PUBLICATIONS

"A Three-State Myo-Electric Control" by D. S. Dorcas et al.,

"Medical & Biological Engineering," Vol. 4, July 1966, pages 367-370 and page facing page 370.

"Myo-Electric Control of Powered Prostheses" by A. H. Bottomley, "The Journal of Bone & Joint Surgery," Vol. 47B, No. 3, 1965, pp. 411-415.

"Muscle Voltage Moves Artificial Hand" by G. W. Horn, "Electronics," Vol. 36, October 11, 1963, pages 34-36.

"The Control of External Power in Upper-Extremity Rehabilitation," National Academy of Sciences - National Research Council, "Signal Processing in a Practical Electro-Myographically Controlled Prosthesis" by A. H. Bottomley, pp. 271-273.

Primary Examiner—Richard A. Gaudet

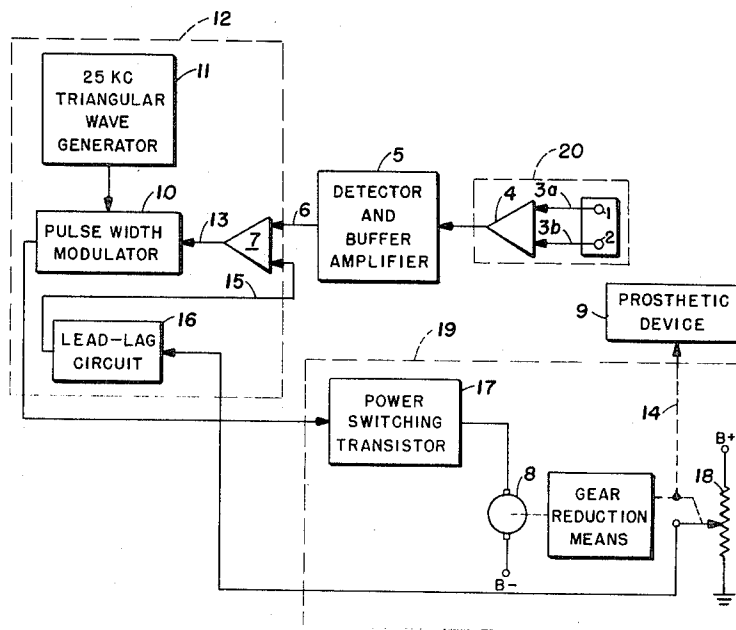
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[57] ABSTRACT

A control system that is suitable for use with prosthetic and orthotic systems utilizing a single site, closed-loop servo system. The closed-loop, position servo system comprises a sensor unit, signal amplifier, control unit, and a power pack unit. The terminal device opens in direct proportion to control signal amplitude as the muscles contract. The control unit conserves battery power by minimizing quiescent current when it is not necessary that the drive motor be powered.

22 Claims, 21 Drawing Figures



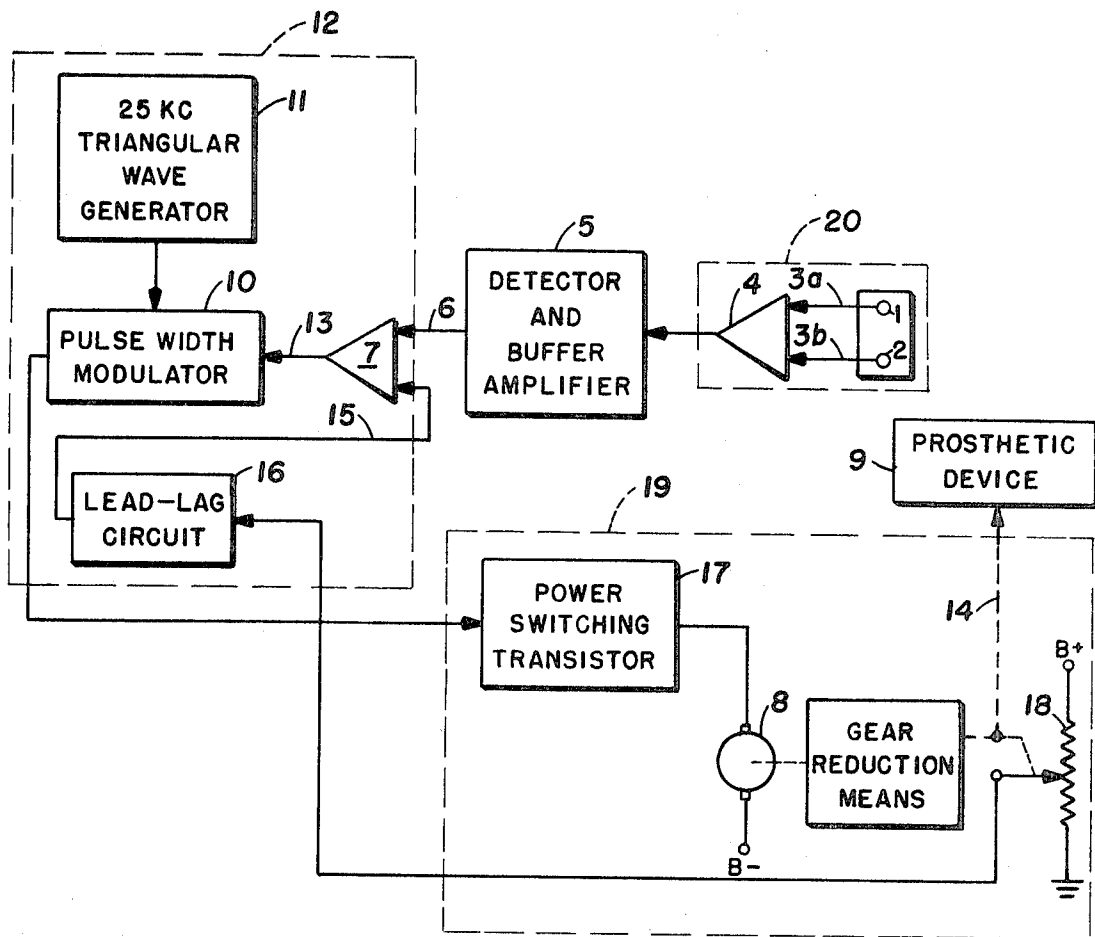


FIG. 1

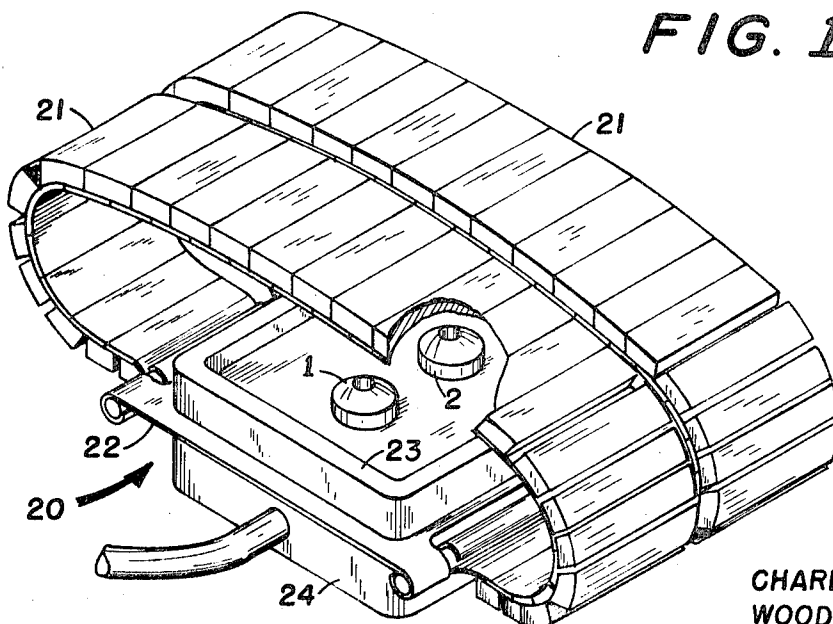


FIG. 2

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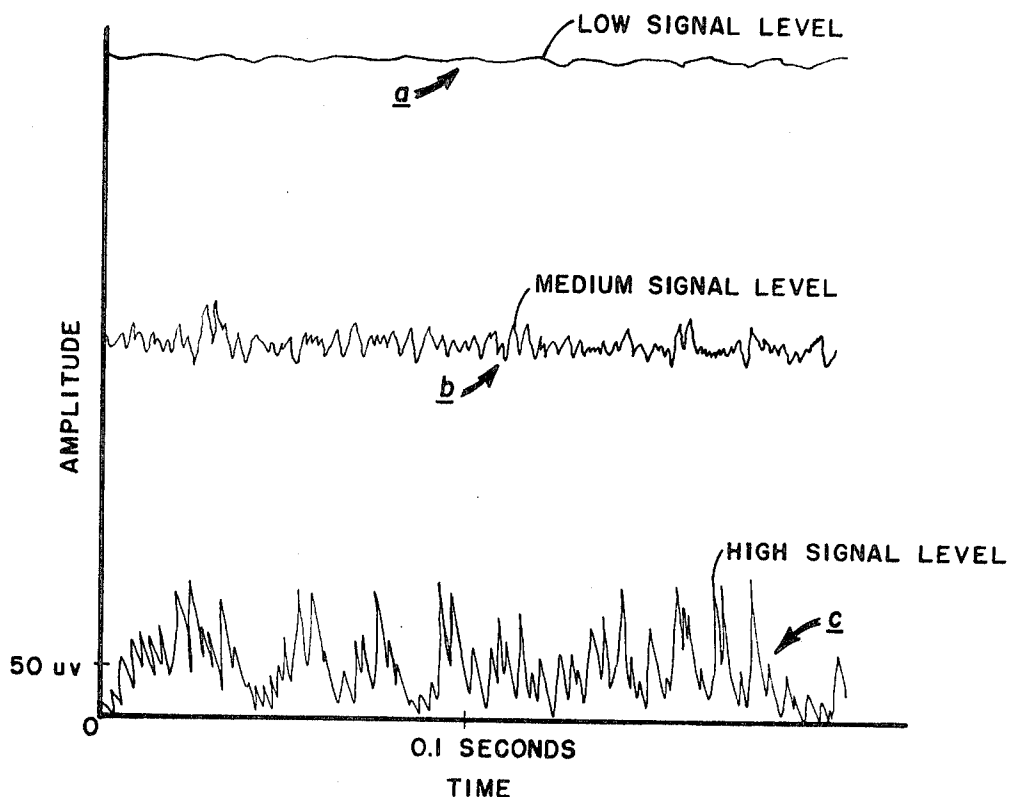


FIG. 3

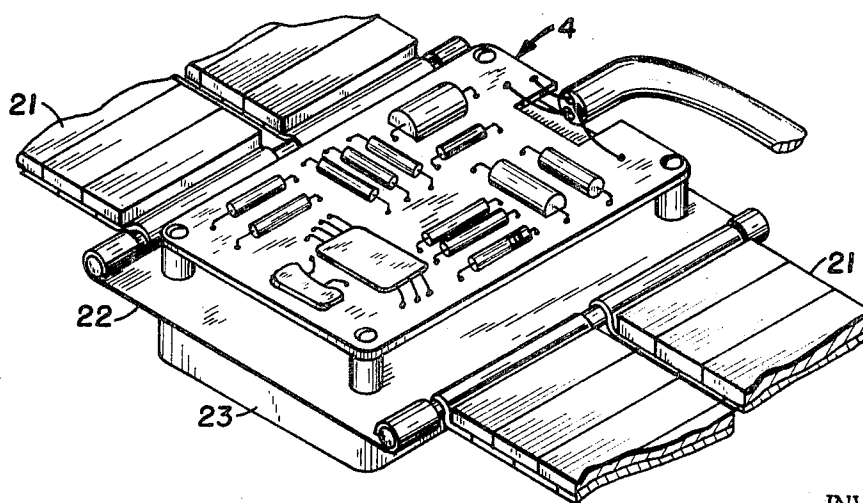


FIG. 4

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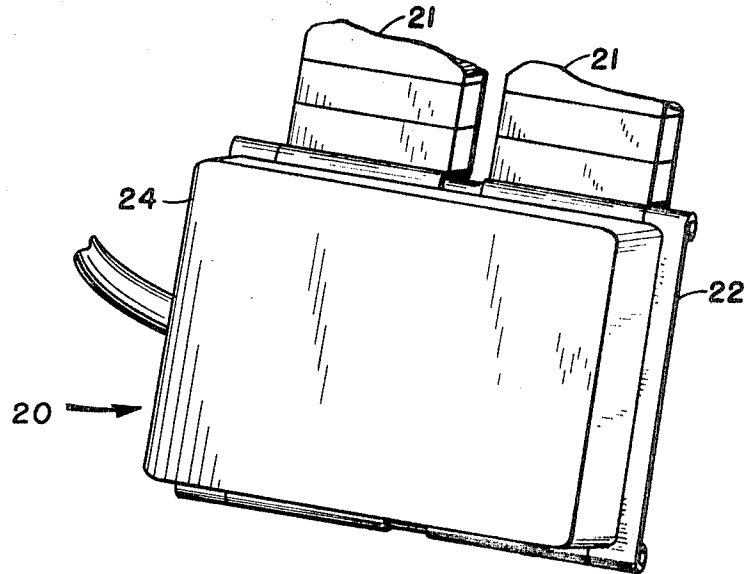


FIG. 5

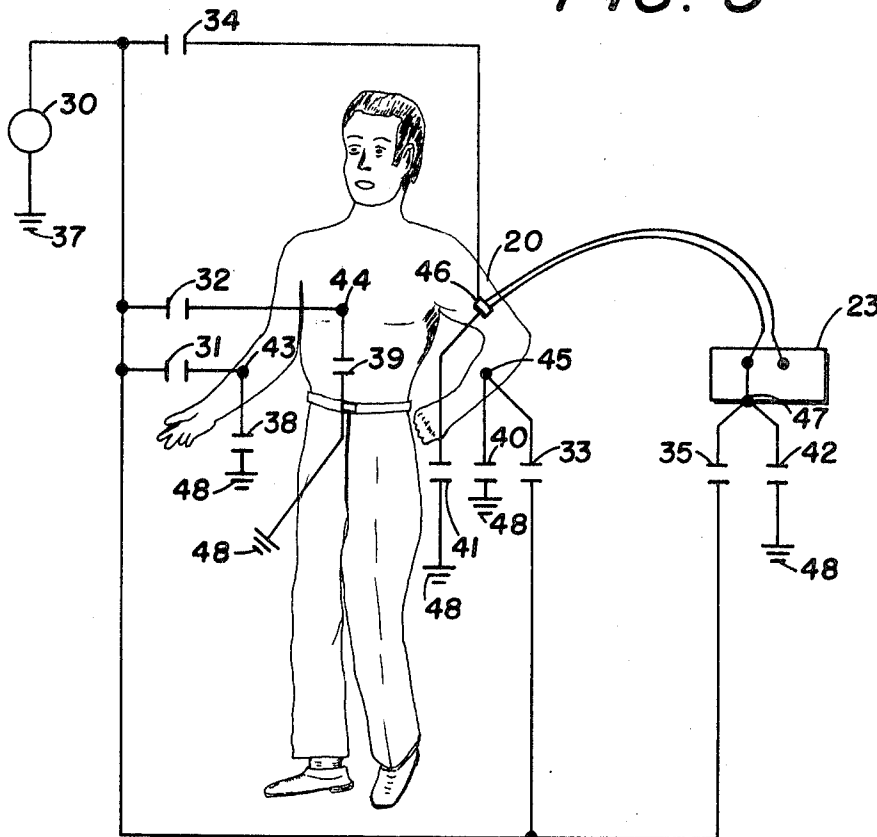
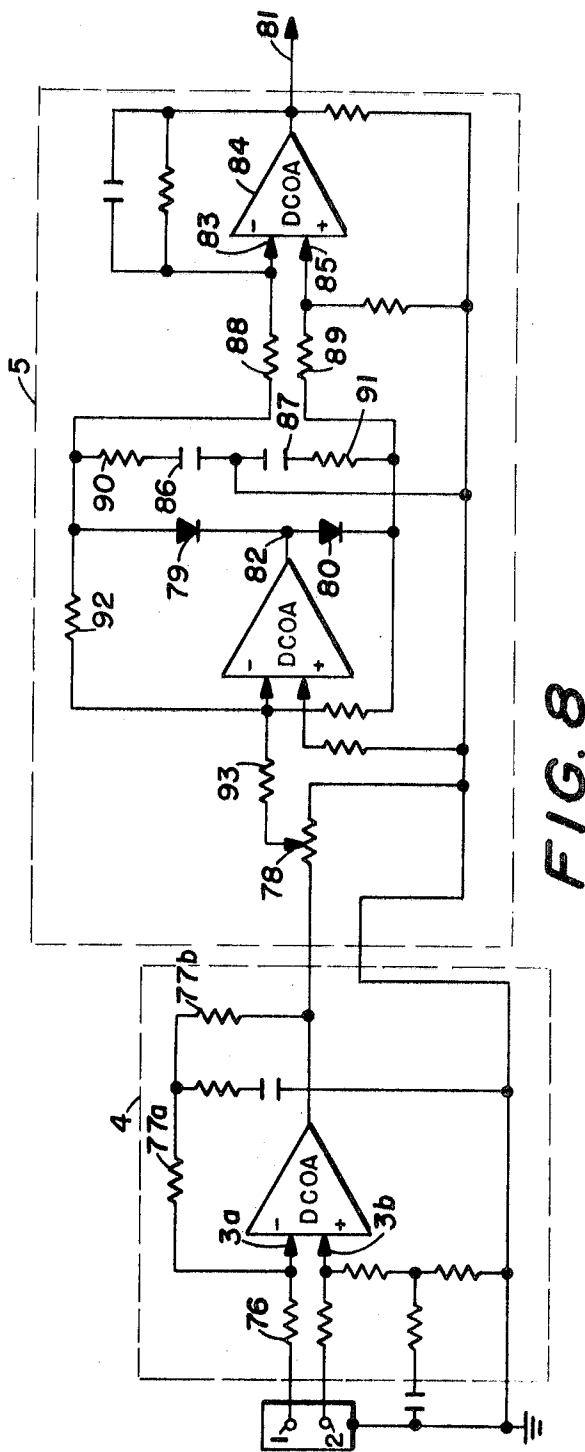
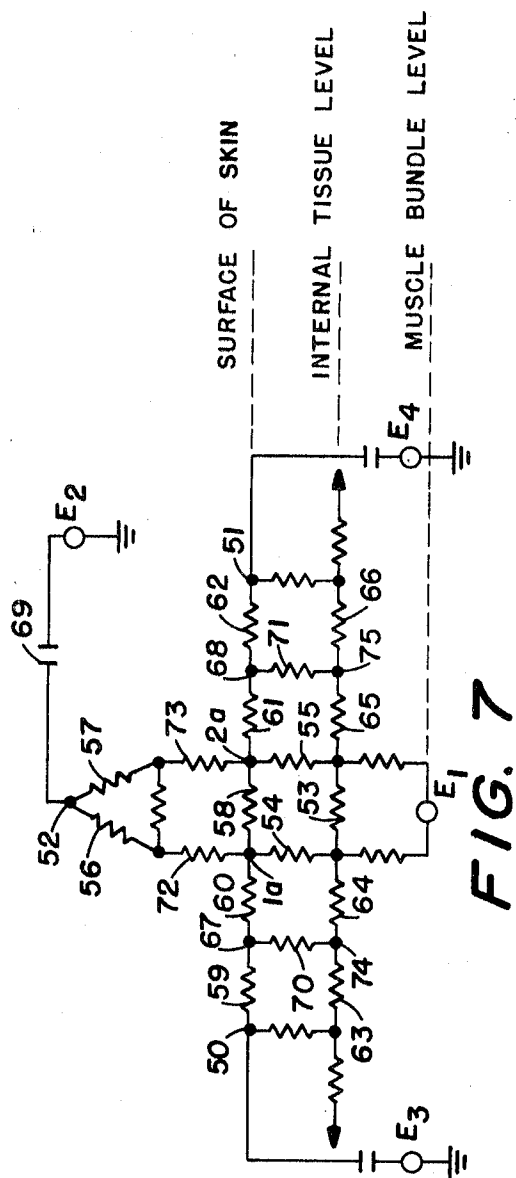


FIG. 6

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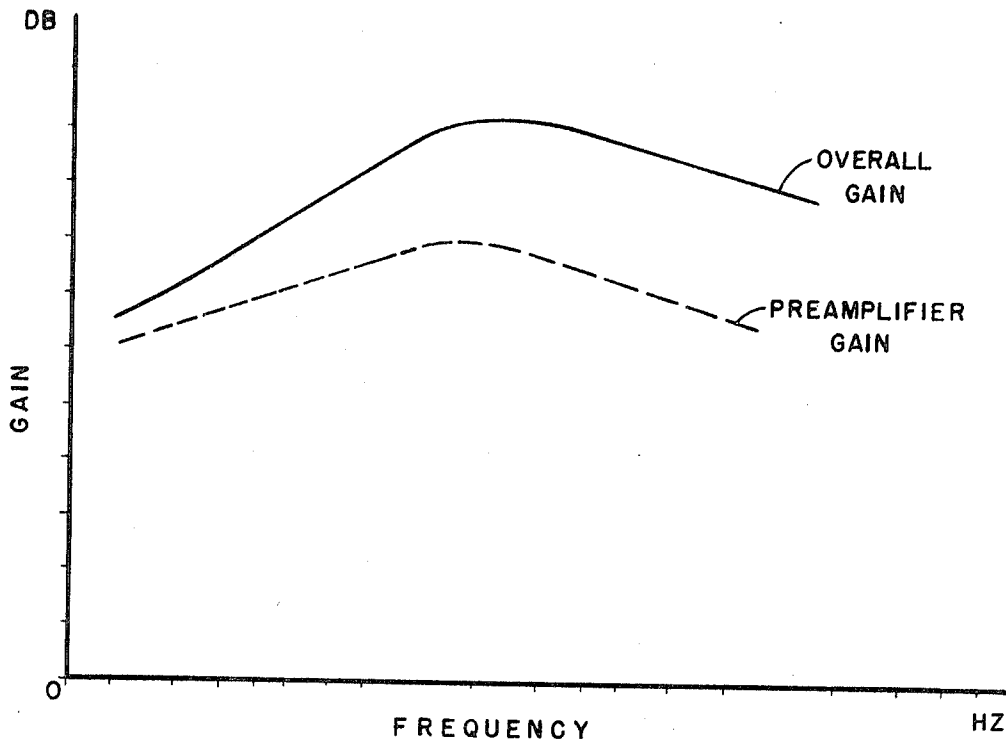


FIG. 9

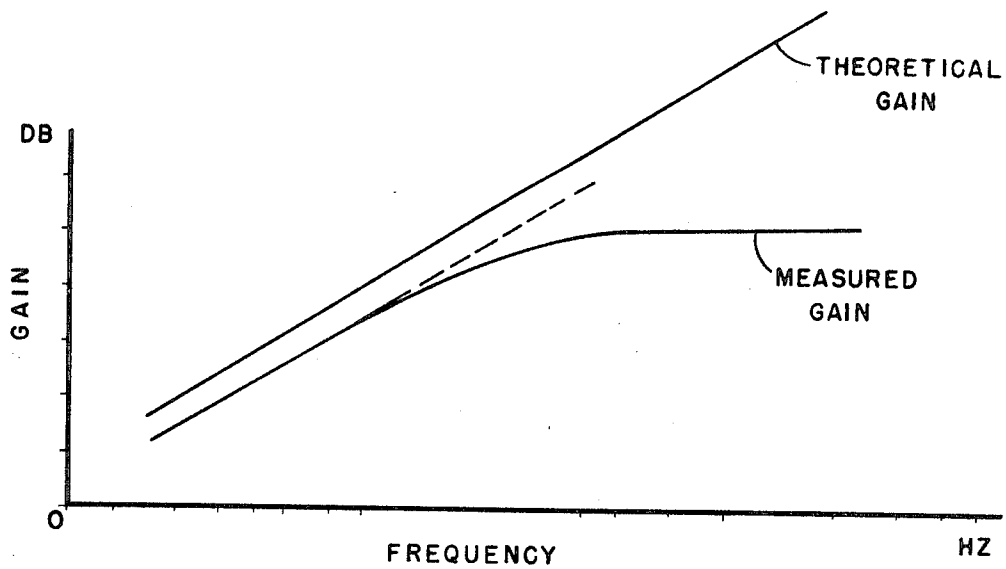


FIG. 10

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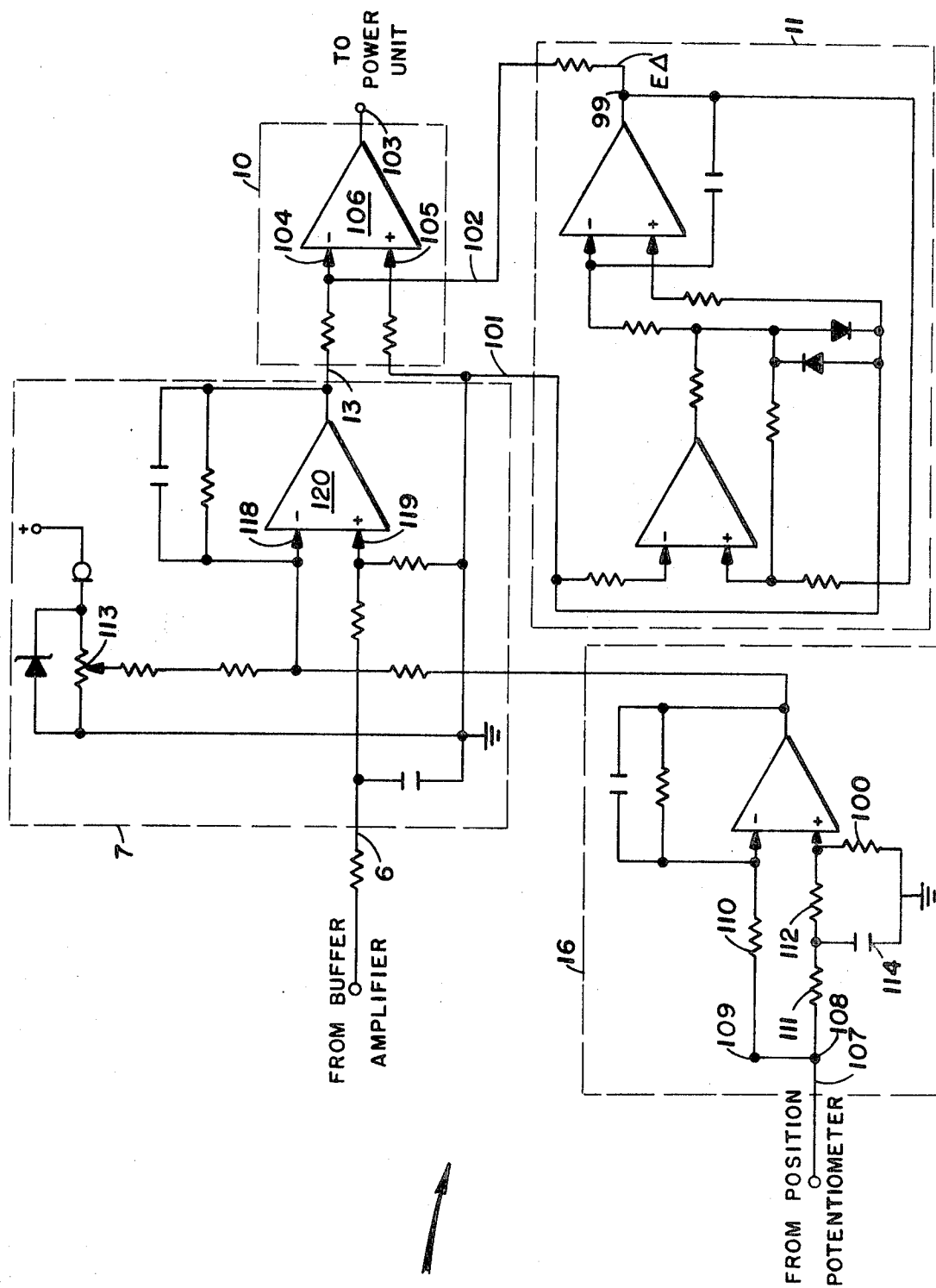
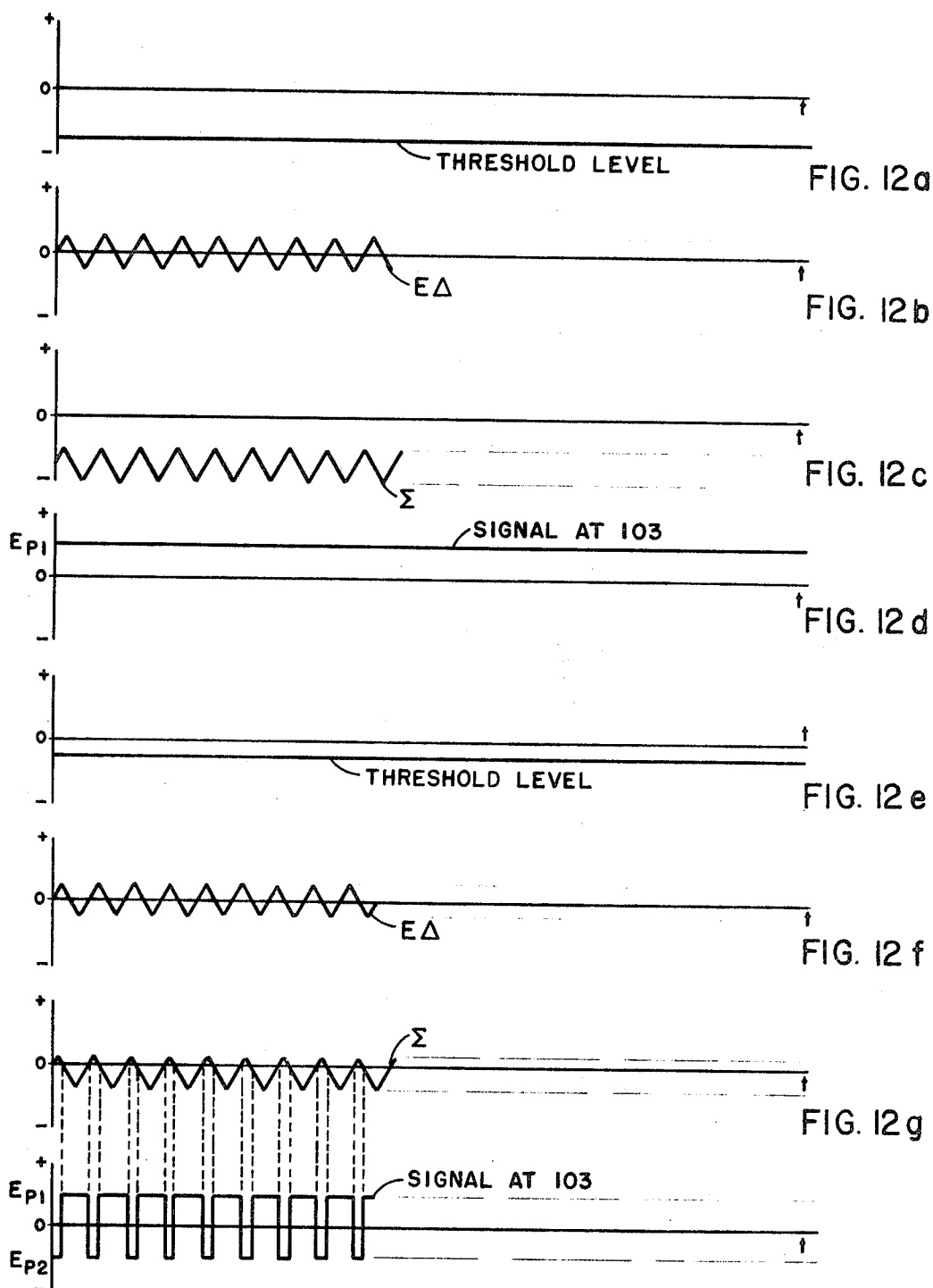


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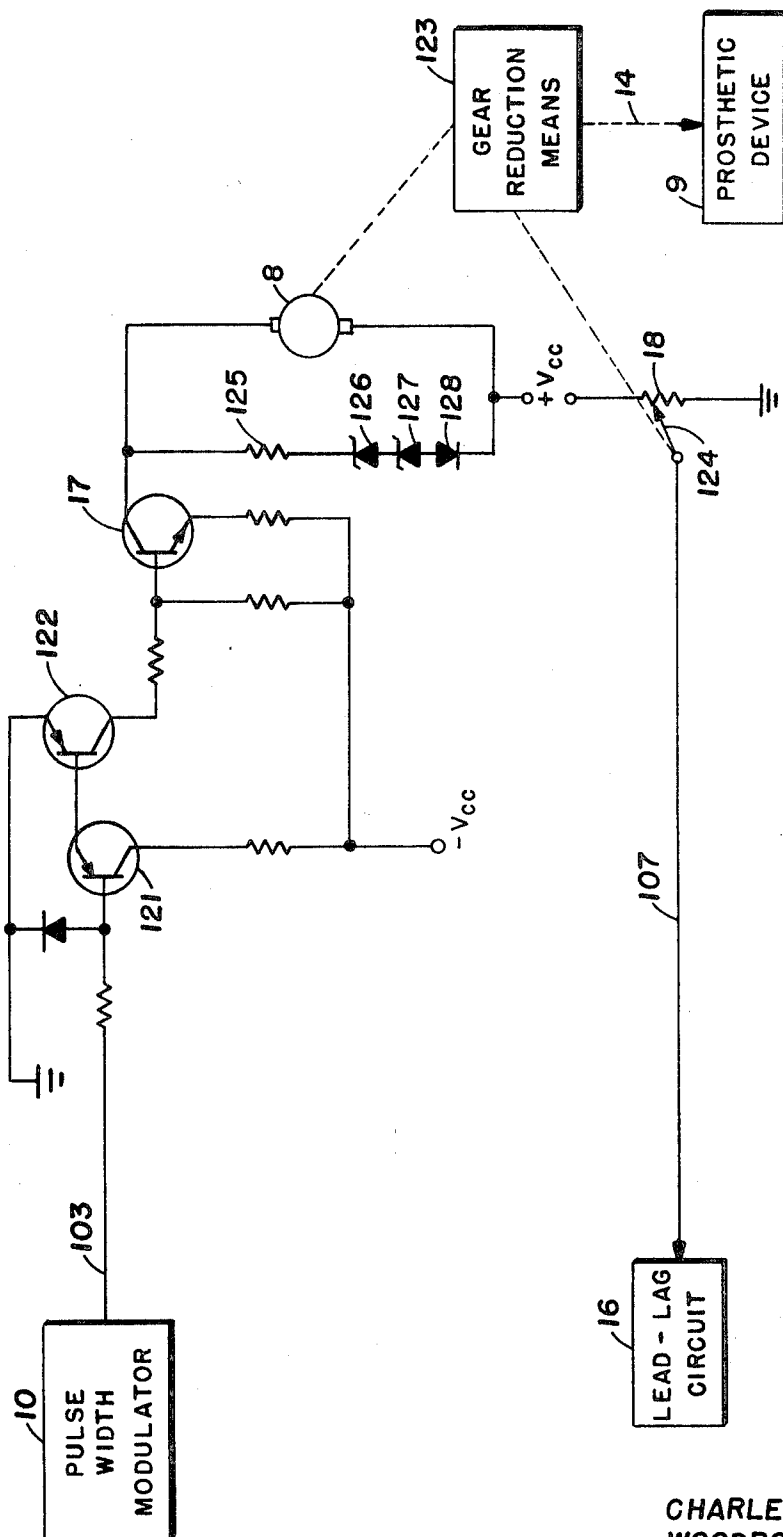


FIG. 13

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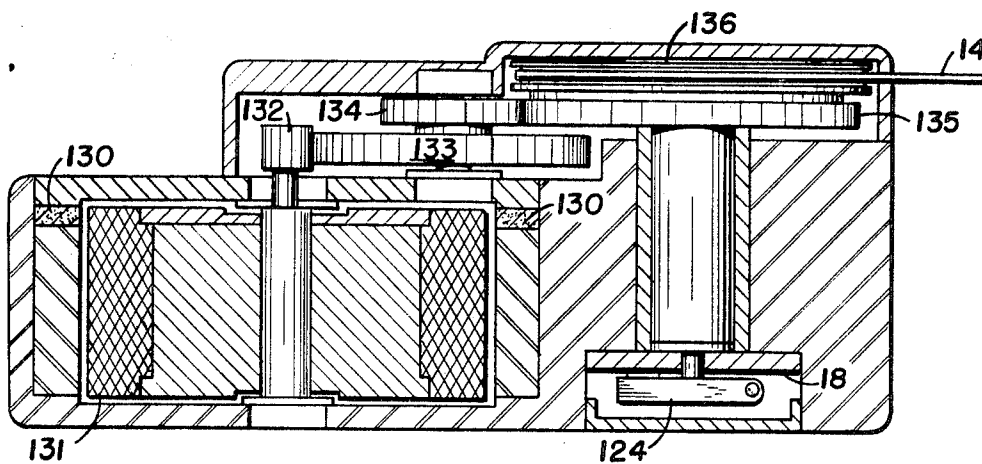


FIG. 14

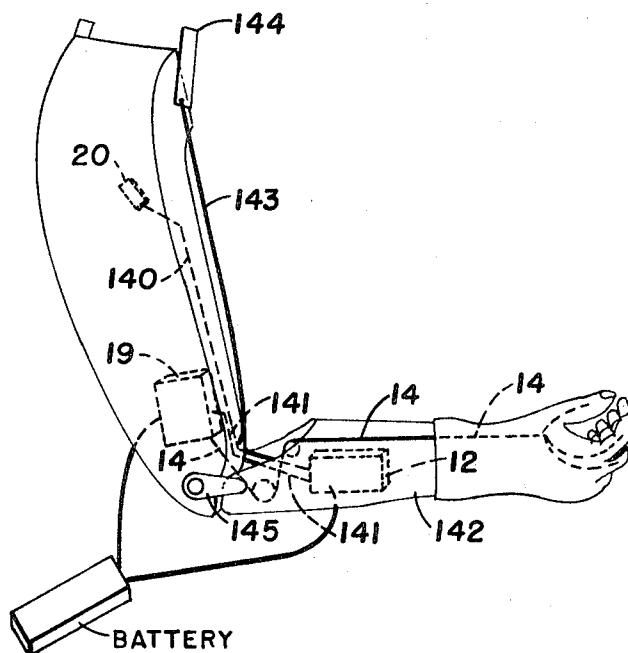


FIG. 15

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MYOELECTRICALLY CONTROLLED PROTHESIS

BACKGROUND OF THE INVENTION

This invention relates to a myoelectric control system for prosthetic or terminal devices. More particularly, the subject invention relates to a closed-loop myoelectric control system that utilizes only one control site for proportionate mode control of an externally powered prosthetic device.

When muscles contract they produce minute electrical potentials. The potentials produced by the voluntarily controlled muscles of an amputee are ideal control signal for prostheses. A prosthesis that responds to muscle contraction in the same way that the replaced part of the body responded could be used "naturally" with a minimum of retraining.

Muscle emf's are known as myoelectric potentials. Electrical conduction of these potentials through body tissue and fluids results in potential differences that can be sensed on the skin. Myoelectric potentials measured on the skin are much attenuated relative to the amplitude of the signals at their point of origin in the muscle. They are composite signals from many muscle fibers. Surface electrodes in contact with (but not penetrating) the skin are used for prosthesis control, despite the weak intensity of the signal, because of the formidable problems encountered in the use of percutaneous electrodes for any length of time.

The myoelectric signals acquired on the skin cannot be precisely described because they are affected by many factors. Among these are: (1) muscle type, function, and condition (including fatigue); (2) characteristics of the tissue, bone and skin that lie between the muscle and the electrodes; (3) electrode material, surface texture, geometry, and spacing; and (4) the location of the electrode relative to the muscle. However, some characteristics of the myoelectric signal acquired on the skin are typical. These are: (1) the signal is an AC voltage that is roughly proportional (in amplitude) to the force developed by the muscles that generate it; and (2) the power spectrum is such that the major portion of the power lies between 30 and 500 Hz. Signal amplitudes on the order of 100 microvolts rms are typical for healthy muscles developing modest tension. Paralyzed muscles often produce myoelectric voltages, but their amplitude is usually much lower than for healthy muscles. Some prostheses that are controlled by myoelectric potentials are unsatisfactory because of difficulties encountered in obtaining signals that are both sufficiently large in amplitude and relatively free of noise and "crosstalk." Crosstalk results when unwanted signals produce by antagonist (and other) muscles are sensed along with the desired signals.

Since myoelectric potentials were first used to control prosthetic devices, many types of systems and control mechanisms have been developed. In most prior art systems the electrode structure typically contains two stainless steel electrodes which make contact with the surface of the skin over the muscle site or, more commonly, sites; tensing of the muscle causes the generation of an electrical signal — the emg (electromyographic) signal — which is AC in form with major frequency components in the 30 to 500 Hz spectrum. The electrical signal sensed by the electrodes is fairly low in level, generally in the 10 μ v rms to 1,000 μ v rms range, depending on the degree of tension in the muscle and other factors, and hence requires amplification

if it is to be useful in driving the synthetically powered device. However, such amplification is usually beset with noise problems because of the low level of the signal. The electric signal processing circuitry must be designed to minimize these problems and provide a useful DC signal, largely free of noise, for the externally powered prosthetic device. Any particular signal level may be generated by tensing the muscle to the desired degree. Typically, a small, high speed DC motor is geared down to drive a mechanical linkage that, in turn, operates the moving parts of the prosthesis. For a self-contained drive mechanism to be satisfactory for a wide range of applications it must fit within the space envelope of the prosthetic device and must satisfy such diverse criteria as moderately high opening and closing velocities and must be light, silent in operation, and low in power consumption.

SUMMARY OF THE INVENTION

The major advantages of the control system of the subject invention accrue from its utilization of only one myoelectric control site and the associated simplicity of control. In these applications the rest position, i.e., terminal device closed, corresponds to minimum control signal voltage, i.e., the muscle is relaxed. The terminal device opens in direct proportion to control signal amplitude as the muscles contract. The technique of utilizing a prosthetic control system in a proportional mode using only one control site effectively eliminates difficulties with electrical crosstalk. Also involved with the subject invention is a remote power pack concept which obviates the requirement for packaging the motor and drive system within the prosthesis. This concept also allows additional flexibility in the selection of components and in the mechanical design of the drive and control mechanisms and can also be used for elbow and wrist rotation.

The myoelectric signal is acquired by surface electrodes that are held in intimate contact with any muscle capable of producing suitable myoelectric signals. The signals sensed by the electrodes are amplified and detected. The output of the detector is applied to a control unit that controls an electric drive motor which operates the prosthetic device. With the muscles relaxed and minimum myoelectric signal generated, the prosthesis is in its closed position. However, when the muscles begin to contract, the electrodes pick up an increasing signal intensity which causes the control unit to drive the motor, consequently opening the prosthesis. The prosthesis, in this case a hand, begins to open until a feedback voltage proportionate to the hand opening position is equal to the control signal. In this manner, the hand is servo controlled for all positions between fully closed and fully open.

An object of the present invention is to provide a myoelectrically controlled prosthesis that utilizes a servo-mechanism control system.

Another object is to provide a prosthesis that is controlled by a closed loop servo system.

Still another object is to provide a prosthetic servo control system which requires only a single site sensor assembly.

A further object of the invention is the provision of a servo control system that can be utilized for both terminal devices and above-elbow prosthetic devices.

Still yet another object is to provide a prosthetic control system that is lightweight, sensitive, and reliable.

Yet another object of the present invention resides in the provision of a prosthetic control system that operates in a proportional mode.

A still further object is to provide a control system for use with a prosthetic device that is operated by a small, self-contained electric motor.

Another object of the present invention is to provide a prosthetic control system that has power control circuits which conserve battery power.

And a further object is to provide a unique prosthetic control system that can be utilized with conventional prosthetic devices.

BRIEF DESCRIPTION OF THE SEVERAL VIEWS OF THE DRAWING

FIG. 1 is a basic block diagram of the invention.

FIG. 2 is a perspective showing the electrode sensors and their associated support apparatus.

FIG. 3 is a chart showing myoelectric signal waveforms generated by neural stimulation of a muscle.

FIG. 4 is a detail perspective showing the physical arrangement of the preamplifier.

FIG. 5 is a perspective showing the preamplifier employed, with the shield cover in place.

FIG. 6 is a simplified diagram illustrating noise pickup resulting from body capacitances.

FIG. 7 is a simplified diagram illustrating the effects of skin impedances in the sensor region.

FIG. 8 is a circuit schematic of the preamplifier, detector and buffer amplifier circuits.

FIG. 9 is a chart showing preamplifier and overall gain characteristics.

FIG. 10 is a chart showing detector gain characteristics.

FIG. 11 is a schematic of the control unit.

FIG. 12a through 12g are charts showing the input and output waveforms of the pulse width modulator.

FIG. 13 is a schematic of the power unit.

FIG. 14 is a cross-section of the power unit.

FIG. 15 shows an arrangement of the subject invention physically located within an above-the-elbow prosthetic device.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENTS

A block diagram of the control system is shown in FIG. 1. The system is a closed-loop position servo with position feedback that follows the myoelectric signal generated by the associated muscles. The myoelectric signal is acquired by a pair of electrodes 1 and 2 that are held in intimate contact with the arm over the associated muscle. The electrodes are spaced approximately 1 inch apart and are placed close to the flexor or extensor muscles (not shown) that control the fingers or anywhere else on the body where suitable signals exist. The myoelectric signal sensed by electrodes 1 and 2 is applied at 3a and 3b to a preamplifier 4 wherein said myoelectric signal is amplified and applied to a detector and buffer amplifier unit 5. Output signal 6 of unit 5 is a DC control signal that is approximately proportional in amplitude to the amplitude of the sensed myoelectric signal. The output control signal 6 of detector 5 is applied to a control unit 12. In order to minimize electrical power consumption a pulse-width modulation system is used to control the output torque of a D.C. torque motor 8 which actuates a prosthetic device 9. More specifically, a pulse-width

modulator 10 utilizes the output of a triangle wave generator 11 and the output 13 of a summing amplifier 7 to provide a pulse-width modulated signal which controls motor current. The D.C. torque motor 8 is driven in one direction only. The output 13 of summing amplifier 7 corresponds to the difference between the amplitude of control signal 6 and the position of the control cable 14 as represented by shaped position signal 15 of lead-lag circuit 16. Since output 13 of summing amplifier 7 reflects the above mentioned difference it will hereinafter be referred to as error signal 13. If the amplitude of error signal 13 is small, current in motor 8 flows for a small part of the duration of the output of triangular wave generator 11. Accordingly, the "on" time of the motor is a function of the magnitude of the error signal 13. By operating a power switching transistor 17 in a power switching mode and causing motor 8 to actuate only when the output of lead-lag circuit 16 is less than the control signal 6, relatively little power is dissipated. When electrodes 1 and 2 are not sensing a myoelectric signal, standby power consumption in the electronic components is quite low, i.e., less than 300 milliwatts. No mechanical switches or special power cutoff relays or circuits are required to switch from "standby" to "operate" condition.

A potentiometer 18 provides a shaped position signal 15 via lead-lag circuit 16 to summing amplifier 7. This feedback arrangement provides high gain at low frequencies and less gain as frequency is increased. Such signal processing makes the opening of the terminal device very easy to control at all elbow flexion positions with or without an object in the terminal device, i.e., the hand. This control technique facilitates a simple interface between the amputee and his prosthesis. The amputee need generate only one signal when he desires to open the hand, and when the control muscle is relaxed the hand is automatically closed by rubber bands and maintains a grasp force without any further effort or attention on the part of the amputee. When the amputee wishes to disengage the hand, he contracts the control muscle. When the command signal provides the voltage needed for the hand opening, the fingers open and the object is freed.

Referring now to FIG. 2, there is shown a sensor assembly 20 in conjunction with an arm-band support apparatus 21. The sensor apparatus consists of an aluminum (or any suitable metal) mounting plate 22, a guard ring assembly 23, and two stainless steel electrodes 1 and 2. In this particular embodiment the electrodes are made of stainless steel but they can be made of some other material, e.g., silver, gold, platinum, or silver-silver chloride. (The sensor electrodes in FIG. 2 are shown positioned close together for purposes of illustration only. Actually they are spaced further apart.) As was previously mentioned, the sensor electrodes 1 and 2 make contact with the skin over the muscle at the control site. Neural stimulation of the associated muscle causes myoelectric signals to be generated. As was previously mentioned and is shown in FIG. 3, the myoelectric signals are AC signals in which major frequency components lie in the 30 to 500 Hz spectrum. Waveform a of FIG. 3 shows the myoelectric output when the muscle is relaxed. When the muscle is in low tension, a waveform such as that illustrated by waveform b of FIG. 3 is generated. Waveform c of FIG. 3 shows a typical myoelectric signal produced when the muscle is in high tension. The myoelectric signal to be sensed

is within the 10–1,000 μV rms range of amplitude values depending upon, among other factors, the tension of the associated muscle.

Amplification is required if the myoelectric signals are to be used to control an externally powered prosthesis. Amplifiers must be carefully designed to minimize noise relative to the low amplitude of the myoelectric signal. The subject invention was designed to minimize these problems and to provide a useful demodulated signal that is largely free of noise. The physical structure of preamplifier 4 of FIG. 1 is shown in FIG. 4. Preamplifier 4 is physically located on the reverse side of the sensor assembly 20 of FIG. 2. Referring back to FIG. 2, the preamplifier circuit 4 is mounted on the aluminum mounting plate 22 which forms the base for the guard ring 23 and the surface electrodes 1 and 2. The aluminum plate 22 serves as a shield between the preamplifier circuit 4 and said surface electrodes 1 and 2 and thus prevents unwanted electrical feedback. It also serves to minimize any direct pickup from stray 60 Hz electric fields. FIG. 5 shows the entire sensor assembly with a shield 24 covering the preamplifier circuit.

The analysis of the effects of 60 Hz noise pickup is a complex 3-dimensional problem. It involves distributed capacitances between the noise source and the body surface area, leakage along the skin (surface) and thru the internal tissues (sub-surface) of the body, and distributed capacitances between the body surface area and the external environment power ground to which the noise source is returned. To simplify the problem and attempt to indicate how noise pickup at the input of the preamplifier 4 can occur, a greatly simplified model describing the noise problem is depicted in FIG. 6. In this figure, the distributed capacitances have been replaced by lumped capacitances to major parts of the body. Noise currents generated by noise source 30 (usually a 60 Hz source) may enter the body parts by means of the stray lumped capacitances C_{N1} through C_{N5} shown at 31 through 35 respectively. There may even be more than one noise source, but for simplicity only one such source is shown at 30. Return of the surface and sub-surface currents to a power ground 37 is through the stray lumped capacitances C_{G1} through C_{G5} shown at 38 through 42 respectively. Because of the potential differences set up between various parts of the body—due to the capacitance voltage dividers being unlike—noise currents will flow on the skin surface and in the sub-surface (internal tissues) from one body part node or circuit node (i.e., from any of the nodes 43 through 47) to another.

The sensor assembly 20 of FIG. 2 was chosen because of the shielding of the two probes, against 60 Hz electric fields, provided by the guard ring 23 and the aluminum plate 22. The guard ring 23 when connected to signal ground 48 tends to establish an equipotential region (at signal ground potential) on the surface of the skin surrounding the two electrodes. As such, it tends to syphon off surface 60 Hz noise currents to signal ground without permitting them to develop a potential difference between the electrodes. If this equipotential region were to extend into the internal tissue region just below the skin surface, any internal tissue currents flowing as a result of 60 Hz noise fields would tend to be syphoned off to signal ground without permitting them to develop a potential difference (which would be sensed by electrodes) in the tissue underneath the elec-

trodes. Because the guard ring contacts only the skin surface, the degree to which the equipotential region can extend itself into the internal tissue region will depend on the skin surface to internal tissue conductivity under the guard ring. Initially this conductivity can be increased by moistening the skin in the contact area. Fortunately, this conductivity should improve and maintain itself with increased time of contact because of the likelihood of perspiration and insensible water penetrating the skin underneath the guard ring. Nevertheless, there will be some finite conductivity between the surface skin and internal tissues and, as a consequence, the internal tissue region underneath the guard ring and probes will not be an equipotential region in general. Hence some 60 Hz noise voltages will very likely develop in this region and be sensed by the probes.

Since the problem is 3-dimensional, internal tissue currents may flow from any direction underneath the probes. It is desirable, therefore, to minimize these currents by establishing an equipotential surface on the surface of the skin in all regions near the electrode structure. For this purpose, the arm band 21 which encircles the arm, supports the electrode structure and holds it in contact with the skin, should be metallic and tied to signal ground.

FIG. 7 depicts the noise situation in the region of the electrode structure. In FIG. 7, internal body capacitances between various skin and tissue layers have been omitted for simplicity. Three levels within and on the body are indicated: (1) skin surface; (2) internal tissue level; (3) muscle bundle level. The body resistances associated with each of these regions and between the regions are also shown. The potential differences between the noise sources developed at nodes 50, 51 and 52 cause currents to flow through the various body resistances (i.e., through skin surface, skin surface to internal tissue, and internal tissue resistances). If any current is permitted to flow through internal tissue resistance 53, an undesirable noise potential will be developed and sensed by the probes unless an equal and opposite potential is developed in the skin surface-to-internal tissue resistances 54 and 55—a very unlikely event in view of the highly variable nature of skin and tissue resistances. This noise voltage could develop even if the common node impedances 56 and 57 were infinite because current may flow through resistance 58 (and develop a noise voltage across the electrodes by virtue of current flow between nodes 50 and 51).

In the absence of a guard ring returned to signal ground, noise currents would flow through resistances 59, 60, 58, 61, 62, and 63, 64, 53, 65, and 66 in FIG. 7. Resistances 59 through 62 represent surface resistances along the skin between the respective nodes. Resistors 53 and 63 through 66 represent internal tissue resistances (i.e., subcutaneous tissues). Thus a noise potential difference could easily be set up between nodes 1a and 2a (the electrode contact nodes) and would be amplified by the preamplifier undesirably. (Those resistors not identified represent inter-region resistances).

If, however, a guard ring is installed (with connection to signal ground) around the probes contacting the skin surface at nodes 67 and 68, an equipotential surface will tend to be set up at least on the skin surface between nodes 67 and 68 between which are located electrode nodes 1a and 2a. Skin surface currents normally

flowing through resistors 59 and 62 to resistors 58, 60 and 61 would tend to be syphoned away from said resistors 58, 60 and 61 and would flow into the guard ring (which is equal to signal ground), to node 52 (which is also equal to signal ground) and back to power ground through stray lumped capacitance 69. These skin surface currents would therefore not be allowed to flow through resistor 58 and consequently no noise potential difference would be developed across the electrodes. If the skin surface to internal tissue resistances 70 and 71 were low, the equipotential surface would in effect extend into the internal tissue region, thereby syphoning currents that flow through the internal tissue resistances 63 and 66 to the guard ring and signal ground and away from internal tissue resistances 53, 64, and 65. The degree to which this syphoning action occurs will depend on how low resistors 70 and 71 can be maintained. With the guard ring and the "extended guard ring" (i.e., the metallic arm bands) installed, adequate noise rejection is possible even when no design precautions are taken to: (1) balance the impedance in both input arms (resistances 72 and 73) of the preamplifier; and (2) insure that common node impedances 56 and 57 are high. Note that use of the guard ring does not reduce the proportion of muscle bundle voltage (E_i) reaching the preamplifier inputs except for the increased loading caused by resistors 60, 61, 64 and 65 which are, in effect, returned to signal ground at nodes 67, 68, 74, and 75 (assuming resistances 70 and 71 are very small).

The effective source impedance of the muscle bundle voltage seen by the electrodes can be as high as 1 megohm with poor electrode contact and decreasing down to approximately 10K ohms with improved contact. To minimize the attenuation of the portion of muscle bundle voltage appearing at the electrodes, due to loading by the differential input impedance seen looking into the preamplifier at the electrodes nodes 1a and 2a, the latter should be made as high as possible, preferably exceeding 200K ohms.

Referring now to FIG. 8, there is shown a detailed schematic diagram of the preamplifier 4, and detector and buffer amplifier unit 5.

Based on the expected emg signal level in the 10 to 1,000 μ v rms range, the preamplifier differential gain should be in the 2,500 to 5,000 range so that its output may be detected at a useful level where additional noise pickup at the output will be of little consequence. The useful emg signal frequency spectrum has been stated to be in the 30 to 500 Hz range with much of the energy concentrated in the 50 to 150 Hz range. Some discrimination against 60 Hz noise can be built into the preamplifier by designing it to have a band-pass characteristic with its center frequency in the 150 to 250 Hz range.

The preamplifier 4 is a bandpass amplifier [which uses a National Semiconductor Type LM301A Monolithic DC Operational Amplifier (DCOA)] having a nominal gain of about 4,350 at a center frequency of about 150 Hz. The common mode impedance will be determined essentially by the impedances external to the monolithic amplifier which return to signal ground from each of the differential inputs; the internal amplifier impedances leading from the differential inputs to signal ground are usually much higher than these external impedances and may be neglected by comparison (since they parallel the external impedances).

With regard to equivalent input noise voltage of the preamplifier, when working from high source impedances (of the order of 500K ohms), this noise should preferably not exceed 5 μ v rms in the 30 to 500 Hz frequency spectrum. If held to this limit, the noise will be insignificant with respect to normal emg signals. DC currents will flow from the guard ring (which is at signal ground potential) through the skin and body to the amplifier input circuit by means of the probes. If the amplifier input DC bias currents are I_1 and I_2 and the input offset voltage is E_M , it can be shown that the skin and body currents flowing into the probes to the negative (inverted) and positive (non-inverted) preamplifier inputs shown at 3a and 3b respectively are I_a and I_b , where

$$I_a = [R_2/(R_1 + R_2)] I_2 + E_M/R_1 \quad (1)$$

$$I_b = [R_2/(R_1 + R_2)] I_2, \quad (2)$$

where I_2 is the amplifier non-inverted input DC bias current and R_1 is the value the DC input resistor 76, and R_2 is the combined value of resistors 77a and 77b. The transfer function $G(s)$ for a differential input signal has the following general form:

$$G(s) = -K(\tau_1 s + 1)/(\tau_2^2 s^2 + 2\zeta\tau_2 s + 1) \text{ where } \tau_1 \ll \tau_2 \quad (3)$$

τ_1 is a function of the feedback network elements, while τ_2 and ζ are functions of the input and feedback network elements and the amplifier open circuit DC gain and its first lag corner frequency.

Maximum differential gain in Equation (3) occurs approximately at the frequency where $\omega = 1/\tau_2$ and this gain is:

$$|G(j\omega)_{\max}| = K\tau_1/2\zeta\tau_2 \text{ with } \omega = 1/\tau_2 \quad (4)$$

Assuming an amplifier open circuit DC gain of 50,000 and a first lag corner frequency of 20 Hz, the transfer function for the preamplifier network elements is:

$$G(s) = -2.925 \frac{1.435s + 1}{[1.062(10^{-3})]^2 s^2 + 2(0.455)[1.062(10^{-3})]s + 1} \quad (5)$$

The maximum gain magnitude is from Equation (4): $|G_{\max}| = 4,350$, occurring at $\omega = 1/\tau_2 = 1/1.062(10^{-3}) = 942$ rad/sec or 150 Hz

If worst case tolerances for the amplifier open circuit DC gain (10,000) and first lag corner frequency (10 Hz) are applied to Equations (3) and (4), we would obtain:

$$G(s) = -2.925 \frac{1.435s + 1}{[3.36(10^{-3})]^2 s^2 + 2(0.229)[3.36(10^{-3})]s + 1} \quad (6)$$

and $|G_{\max}| = 2,730$ occurring at $f = 47.5$ Hz.

For comparison purposes, the gain magnitude as a function of frequency for Equation (5) (for DCOA open circuit gain of 50,000 and first lag corner frequency of 20 Hz) is plotted in FIG. 9. Also plotted is

the overall gain of the preamplifier 4 and detector and buffer amplifier unit 5.

Because different subjects will have different emg potential sensitivities, a gain control, potentiometer 78 is added to the preamplifier output circuit to permit adjustment of the signal level reaching the detector input.

The emg detector and buffer amplifier unit schematic is generally shown at 5 in FIG. 8. The detector is an absolute value type with a modification to make it frequency sensitive. Detection polarity is determined by the direction in which the diodes 79 and 80 are installed. To produce a positive DC rectified signal at the buffer amplifier output 81, the detector output voltage at the terminal 82, fed to the inverted input 83 of the buffer amplifier DCOA 84 must be negative with respect to buffer amplifier non-inverted input 85, since the buffer amplifier produces a sign inversion. The negative potential is produced by installing the diodes as shown. If the diodes are inverted, the detector and buffer amplifier outputs will be inverted. The function of the buffer amplifier is to provide isolation between the detector output and the load to be driven.

By adding capacitors 86 and 87 the detector is made frequency sensitive; its output will be essentially proportional to frequency for frequencies considerably below and above 60 Hz. The gain of the detector, i.e., the ratio of average value of the rectified DC output (E_o) appearing at the buffer amplifier differential inputs to the RMS signal input (E_1) appearing at the 10K gain adjustment potentiometer wiper, is as follows:

$$E_o/E_1 = 4 \sqrt{2} (R_F/R_1) (f R_F C) (\text{VDC/VRMS}) \quad (7)$$

subject to the restriction that:

$$f > > 1/4R_F C \quad (8)$$

where:

E_o = Average value of rectified DC output voltage (i.e., differential voltage measured across buffer amplifier input resistors)

E_1 = RMS voltage at detector input

R_F = Feedback resistance (resistor 92)

R_1 = Input resistance connected to DCOA inverted input (resistor 93)

C = Capacitance in detector output circuit (capacitors 86 or 87)

f = Frequency in Hz.

At frequencies much lower than $f = 1/4R_F C$, the gain is essentially:

$$E_o/E_1 = 2 \sqrt{2/\pi} (R_F/R_1) (\text{VDC/VRMS}) \text{ for } f < < 1/4R_F C \quad (9)$$

At DC, the gain is simply:

$$E_o/E_1 = R_F/R_1 (\text{VDC/VDC}) \quad (10)$$

A plot of the detector gain as a function of frequency is given in FIG. 10. Also shown for comparison purposes is a plot of the theoretical gain given by Equation (7) which holds for frequencies essentially above 30 Hz. Gain differences between the theoretical and ob-

served functions may be attributed in part to component tolerances, particularly to the tolerance on capacitors 86 and 87, and to the loading effect on the detector output caused by the buffer amplifier input resistances 88 and 89.

As the frequency increases, E_o would increase for a constant E_1 input, according to Equation (7) until either amplifier saturation occurred or the open loop gain roll off (of the DCOA) with frequency caused the output to fall off. To insure linear operation and since the emg frequency spectrum of interest lies between 30 and 500 Hz, an effective high frequency lag corner in the gain characteristic was introduced by inserting resistors 90 and 91 in series with capacitors 86 and 87.

Referring now to FIG. 11, there is shown a schematic diagram of the control unit 12, which receives from the buffer amplifier the sensed emg signal 6 and passes this signal to summing amplifier 7 and then to pulse width modulator 10. The output of the pulse width modulator is subsequently applied to the power unit (not shown) which ultimately actuates the prosthetic device (the hand). In order to minimize electrical power consumption within the power units a pulse width modulation system is employed. The pulse width modulator 10 utilizes the outputs 102 and 101, respectively, of the triangle wave generator 11 and the output of servo amplifier 120 to provide a pulse-width modulated signal 103 which regulates motor current.

In order to more clearly describe the operation of the control unit, any discussion of the lead-lag circuit 16 and the associated feedback loops will be temporarily omitted. The output signal 13 of summing amplifier 7 represents a DC signal that is proportional to muscle tension. This DC signal is applied to input 104 of pulse width modulator 10. Also, the output 102 of triangle wave generator 11 is received at input 104. The operation of the pulse width modulator 10 can best be explained by additional reference to FIG. 12. The DC output 13 of servo amplifier 120 is shown by waveform *a* of FIG. 12. In this condition, there is no sensed emg signal being received from the buffer amplifier 6. The DC output signal as shown in waveform *a* is obtained from a potentiometer 113 and applied to input 105 of operational amplifier 106. The output, $E\Delta$, from output terminal 99 of triangle wave generator 11, shown by waveform *b* of FIG. 12, is applied to negative input 104 of operational amplifier 106. Waveforms *a* and *b* are combined within amplifier 106 to produce the summed waveform *c*. As long as the summed waveform is less than zero volts, the output of the pulse width modulator 10 is as illustrated by waveform *d* which represents the control unit in the "off" condition. Thus, the prosthesis does not respond since there is no effective enabling signal. When an emg signal is sensed, there is applied to input 119 of summing amplifier 120 a positive DC signal. When combined with the threshold level signal as produced by potentiometer 113, the signal as represented by waveform *e* is thus applied to input 104 of amplifier 106. When waveform *e* is combined with the $E\Delta$ - waveform *f*, the resultant signal as shown by waveform *g* is produced. Whenever the upper peaks of waveform *g* exceed zero volts, the pulse width modulator 10 shifts from the non-enabling E_{p1} voltage level to the enabling E_{p2} level. In this manner does the prosthetic device actuate only when the sensed emg voltage level exceeds a predesignated (and variable) threshold. As the prosthesis is being actuated, its physical position

is indicated by the wiper arm of a position potentiometer (not shown). The feedback position signal 107 is applied to the lead-lag circuit 16 at terminals 108 and 109. The function of lead-lag circuit 16 is to prevent actuation of the prosthesis by a short term, high gain signal, e.g., noise. Via resistors 100, 110, 111 and 112 and capacitor 114, the prosthesis will only respond to long term signal, thereby preventing the prosthetic device from continually opening and closing upon every sensed signal.

Referring now to FIG. 13, there is shown a schematic diagram of the power unit. Emergent from pulse width modulator 10 is a series of enabling pulses as shown by lower portion of waveform g) of FIG. 12. These pulses are applied to the motor 8 after being amplified by power transistor 17. Transistors 121 and 122 serve as driver transistors for power transistor 17. Upon energization, the rotation of the armature of motor 8 causes like rotation in gear reduction means 123. After the necessary gear reduction is accomplished, the control cable 14 is connected to a pulley that is attached to the last gear element (not shown). Attached to the end of the control cable is the prosthetic device 9. In this manner, rotation of the motor shaft causes activation of the prosthetic device 9. Also connected to the gear means 123 is the wiper arm 124 of potentiometer 18. In this manner, position feedback signal 107 is provided for lead-lag circuit 16. A transient suppression circuit is provided with the power unit for inhibiting undesired interference with power transistor 17 and driver transistors 121 and 122 when motor 8 is switched on and off. The transient suppression circuit consists of resistor 125, zener diodes 126 and 127, and diode 128.

The gear box reduction means is shown in the cross-sectional drawing of FIG. 14. The output signal of power transistor 17 of FIG. 13 is applied to brush assembly 130 causing rotation of the rotor 131, and subsequently rotor pinion 132. Rotation of rotor pinion 132 induces rotation in spur gear 133 and pinion gear 134. Final gear reduction is accomplished by spur gear 135 to which is attached at its upper end pulley 136 and at its lower end potentiometer 18. Wrapped around pulley 136 is the control cable 14 which actuates the prosthetic device. The position feedback signal is provided by potentiometer 18 and wiper arm 124 as heretofore discussed.

Referring briefly to FIG. 15, there is shown one embodiment illustrating the physical arrangement of the subject invention in conjunction with an above-the-elbow amputee. The sensor unit 20 is held in intimate contact with the biceps muscle of the amputee. The output signal of the sensor unit is connected to the control unit 12 via signal cable 140. Control signals from the control unit 12 to the power unit 19 as well as position feedback signals from the power unit 19 to the control unit 12 are transmitted via signal cable 141. Upon receiving energization commands from the control unit 12, the motor causes the control cable 14 to retract (as described previously), consequently opening the fingers of the prosthetic hand. Since the operation of the prosthetic hand (or hook) is well-known in the art, only nominal attention will be directed thereto. Retraction of cable 14 will cause forearm 142 to raise if the elbow is unlocked. When the forearm is in the desired position, the amputee pulls on locking cable 143 by means of shoulder harness 144, thereby causing lever arm 145 to lock said forearm 142 into position.

Again, since this technique is well known in the art, no detailed discussion will be accorded thereto. When the forearm 142 is locked into position via lever arm 145, further retraction of the control cable 14 will cause the fingers of the prosthesis hand to open.

As was mentioned above, placement of the motor within the prosthetic device (hand) is but one embodiment of physical placement. Another embodiment that has been utilized is placement of the power unit around the waist of the amputee when there is not enough room for it in the prosthesis.

Other modifications, adaptations and embodiments of the present invention are of course possible in light of the above teachings. Therefore it should be understood that within the scope of the appended claims the invention may be practiced otherwise than as specifically described.

We claim:

1. A control system for actuating a prosthesis from myoelectric potentials, comprising:

a prosthesis;

electrode means for sensing myoelectric potentials developed by the body, said electrode means being adapted to engage over a single muscle area of the body;

means actuating said prosthesis in response to said sensed myoelectric potential;

means for providing a position feedback signal to said actuating means for identifying the degree of actuation imparted to said prosthesis; and,

means for delaying in time said position feedback signal.

2. The control system as claimed in claim 1 wherein said electrode sensing means comprise spaced first and second electrodes each formed of a metal biologically compatible with the skin and capable of making a low resistance contact therewith.

3. The control system of claim 2, wherein said means for minimizing said interference noise signals comprises

a guard ring assembly encircling said sensing electrodes, the assembly being connected to signal ground and adapted to contact the skin surface, a mounting plate adapted to be spaced from the skin surface and forming the base of the guard ring assembly and having the electrodes positioned on one side thereof, and

a preamplifier positioned on the mounting plate on the side thereof opposite the electrodes, the mounting plate serving as a shield between the preamplifier and the electrodes to prevent electrical feedback and to minimize direct pickup from stray electric fields.

4. The control system as claimed in claim 1 and further comprising:

means for converting said sensed myoelectric potential into a DC level signal; and

means for providing interfacing between said converting means and said actuating means.

5. The control system as claimed in claim 4, wherein said converting means is a detector and said interfacing means is a buffer amplifier.

6. The control system as recited in claim 5 wherein said means for providing said position feedback signal comprises a potentiometer whose wiper arm is mechanically actuated in direct proportion to the degree of actuation imparted to aid prosthesis, the voltage sensed

by said wiper arm being received by said actuating means.

7. The control system as claimed in claim 6 wherein said actuating means comprises:

means for determining the error difference signal between said position feedback signal and said DC level signal;

means for applying electrical energy to said prosthesis only when said error difference signal is of a predetermined amplitude; and

means for converting said electrical energy into mechanical energy for actuating said prosthesis.

8. The control system set forth in claim 7 wherein said means for determining said error difference signal is a summing amplifier comprising:

a first input terminal for receiving said DC level signal;

a second input terminal for receiving said position feedback signal;

means for providing a fixed threshold signal; and

means for obtaining the difference between said DC level signal and said fixed threshold signal and producing a difference error signal when the difference signal is of an amplitude greater or less than said position feedback signal.

9. The control system as claimed in claim 8, wherein said means for converting said electrical energy into mechanical energy for actuating said prosthetic device is a DC torque motor.

10. The control system of claim 1 and further comprising;

means for minimizing interference noise signals; means for amplifying said myoelectric potentials; and means for attaching said electrode means to the body.

11. A control system for actuating a prosthesis from muscle myoelectric potentials comprising:

a prosthesis,

means for sensing myoelectric potentials comprising electrode sensing means adapted to engage over a muscle area of the body,

means for amplifying said sensed myoelectric potentials,

means for converting amplified myoelectric potentials into a DC level signal having an amplitude proportional to the intensity of said generated myoelectric potentials,

means for identifying the exact physical position of said prosthetic device and for producing therefrom a position feedback signal,

means for actuating said prosthesis in response to said sensed myoelectric potentials, said actuation means providing an electrical signal sufficient to actuate said prosthesis and including a summing amplifier for producing such electrical actuation signal when the resultant threshold output amplitude of said means for converting the amplified myoelectric potentials into a DC level signal is

greater than the amplitude of said position feedback signal,

means for converting said electrical signal into mechanical energy; and,

means for producing a control signal effective to actuate said prosthetic device only when sensed myoelectric potentials are above a predetermined threshold level, said degree of prosthetic actuation being proportional to the amplitude of said sensed myoelectric potentials, said means for producing said control signal comprising a standard waveform generator having a predetermined amplitude, said electrical actuation signal being generated when said output amplitude of said summing amplifier is greater than said predetermined amplitude of the standard waveform.

12. The control system as claimed in claim 11, wherein said sensing means comprises electrode means adapted to a single muscle site.

13. The control system as claimed in claim 12, wherein said electrode means comprises first and second electrodes.

14. The control system as claimed in claim 11, wherein said means for converting said amplified myoelectric potential into a DC level signal is a DC detector.

15. The control system as claimed in claim 11, wherein said means for converting said electrical signal into mechanical energy comprises a DC torque motor.

16. The control system as claimed in claim 15, and further including gear means between said DC torque motor and said prosthesis, actuation of said DC torque motor causing said prosthesis to operate at a speed determined by the final ratio of said gear means.

17. The control system as claimed in claim 16, wherein said means for producing said position feedback signal comprises a potentiometer having a wiper arm, said wiper arm being operably connected to the output of said gear means.

18. The control system as claimed in claim 15, and further including means for suppressing spurious responses generated by the actuation and subsequent non-actuation of said DC torque motor.

19. The control system as claimed in claim 18, wherein said spurious response suppression means comprises a plurality of serially connected zener diodes all connected in parallel with said DC torque motor.

20. The control system as claimed in claim 11, wherein said means for generating said standard waveform is a triangular wave generator.

21. The control system as claimed in claim 20, wherein said means for generating said electrical actuation signal is a pulse width modulator.

22. The control system as claimed in claim 11, and further including means for delaying in time said position feedback signal.

* * * * *

UNITED STATES PATENT OFFICE
CERTIFICATE OF CORRECTION

Patent No. 3,735,425 Dated May 29, 1973

Inventor(s) Charles H. Hoshall, et al.

It is certified that error appears in the above-identified patent and that said Letters Patent are hereby corrected as shown below:

On the cover page and at Column 1, line 1, the title should read -- Myoelectrically Controlled Prosthesis --.

On the cover page, [75] "Robert L. Konigsbert" should read -- Robert L. Konigsberg --.

Column 8, line 29, change << to >>.

Column 12, line 67, change "aid" to -- said --.

Signed and sealed this 26th day of February 1974.

(SEAL)
Attest:

EDWARD M. FLETCHER, JR.
Attesting Officer

C. MARSHALL DANN
Commissioner of Patents