THE UNIVERSITY OF MICHIGAN
College of Engineering
Dept. of Electrical Engineering
Student Design Project:

ELECTROMYOGRAPHICALLY CONTROLLED ORTHOTIC HAND

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ABSTRACT

The project is concerned with the development of an orthotic device capable of restoring the grasping action of a paralyzed hand to a patient. The resulting device is controlled myoelectrically and actuated by an electromechanical system which provides translational motion for operation of a typical hand splint. Special features of the device include:

1) Proportional control over the device by the operator.
2) An electronic feedback system to restore the sense of touch and pressure to the hand.
3) Efficiency, quiet operation, and compact size of the actuator and actuator linkage.
4) Small requirements of physiological energy due to the patient's capability to select various system modes.

The device is applicable to patients with disorders ranging from total paralysis below the neck to paralysis of only the hand.

A history of ortheses, market evaluation, and technical description of design considerations and resulting solutions are included. It is concluded that the device represents a significant advancement in ortheses; suggestions for further research and testing are outlined.

COURSE DESCRIPTION

Development of the orthotic system was the subject of the Senior Project Course of the Department of Electrical Engineering of The University of Michigan College of Engineering. The researching, design, construction, and testing of the electronic circuits and mechanical actuator and the written account of the work were all carried out by the students during the four month semester. The orthotic hand splint used was designed and built by the staff of the Physical Medicine Department of The University of Michigan School of Medicine.

The fourteen students jointly formulated the design objectives and planned the overall system operation and then formed the following project groups with designated assignments:

1) Sensing and processing: electromyographic signal detection and subsequent amplification and integration; restoration of of the pressure sensation using a feedback system.

2) Control: conversion of the integrated signal into a control signal for the mechanical actuator.

3) Actuator: development of a mechanical actuator and linkage to operate the hand splint.

4) Market study: determination of the economic feasability of the orthotic device.

The Project Evaluation and Review Technique (PERT) was employed to formulate a time schedule for the semester's activities. An oral presentation was held for faculty and guests at the close of the semester.

ACKNOWLEDGEMENTS

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DESIGN OBJECTIVES

The project involved the researching, design, construction, and testing of an orthotic device capable of restoring to a patient the use of a paralyzed hand. The design objectives for the device were considered in terms of ideal and minimum levels of attainment and fall into four groups: restoration of function, avoidance of undesirable side effects, acceptability to the patient, and economic feasability.

Restoration of Function

Because the actual positioning of the hand is considered a design problem in upper extremity ortheses design, restoration of function generally means restoring the ability to grasp objects. Ideally, the patient should have absolutely natural control over his hand. He should be capable of initiating movement of his hand by normal nervous activity, and he should be made capable of altering his method of grasp in accordance with the size, shape, weight, fragility, and friction coefficient of the particular object to be grasped. These objectives require the following characteristicts, in the lower extremity device:

1) The patient should be capable of positioning fingers individually.
2) The maximum level of pinch force should be approximately 15 pounds. Assuming nominal coefficients of friction, the patient will then be able to lift objects of about five pounds weight by holding them between the fingers and the thumb.
3) The minimum time to close the hand should be about 0.4 seconds, and this time should be user variable.

4) The system should employ the central nervous system as a means of control. Interaction with the patient's nervous system should occur in two ways: a) the signal used to control the hand should be obtained from a physiological source; b) as the patient will probably have lost the sense of touch and pressure in his hand, a feedback system originating at a sensor on or near the hand and terminating at an input to the patient's nervous system should be included.

5) Control should be proportional to variations in the physiological source.

It is not possible to state accurately a minimum level of performance at this initial stage in development. A majority of patients must be satisfied with the restored function, when the degree of restoration is considered relative to undesirable side effects, cost, and cosmesis.

Avoidance of Physically Harmful Effects

The system must be designed so as to prevent the following problems:

1) Adverse effects on tissue due to irritation, excessive pressure, or changes in blood circulation at any interface between the device and the patient.

2) Adverse sensory stimulation such as strong light, loud or badly pitched sound, pain, or heat.

3) Excessive demands of energy from physiological sources.

4) Interference with functions otherwise available to the patient.

As a minimum, there should be no effect which prevents the patient from wearing and operating the device eight hours per day, every day.

Acceptability to the Patient

The following requirements should be met:

1) Operation should be without sound.

2) Operation should be easily learned and performed. The amount of concentration required for operation should not be so great as to detract from the ability of the patient to carry on concommitant activities.

3) The device should be cosmetically pleasing. There should be no lustrous or brightly colored components; it should be compace and inconspicuously worn.

4) Required maintenance should not be complex in nature or frequent in need.

5) The device must present no physical discomfort that often "reminds" the patient that he is wearing it.

6) Attachment of the device should not prevent use of upper-extremity ortheses.

As a minimum, the device should introduce no seriously disturbing psychological phenomena.

Economic Feasability

The capital required to carry out the extensive research and design necessary to produce a highly acceptable product can be obtained only if the foreseeable market value is encouraging. The profit-making capability can be enhanced in five ways:

1) The device must cover a wide enough range of patients to ensure enough buyers to make the product economically feasable. Extensive loss of function in many parts of the body should not seriously limit the possibility of restoring use of a hand.

2) The cost of the device must be within a reasonable range for purchase by a large fraction of possible patients.

3) Cost of operating the device due to power supplies and periodic replacement of parts should be minimal.

4) The reengineering required to adapt the device to a particular patient and a particular physiological source of control must be minimal.

5) The device must more closely meet the demands of patients than other systems currently available.

HISTORY

The human body is extremely complex. Simulating its functions is the most sophisticated engineering problem imaginable. Starting in the middle ages, men have replaced amputated limbs with prosthetic devices driven by the patient's muscles. In the early 1950's, Norbert Wiener suggested using myo-electric signals to control externally powered prostheses. An amputee could control his artificial limb with the muscles that had driven the same limb before amputation. Also by this method, patients with no available driving muscles could control prosthetic devices.

In 1955, Nightingale and Whillis showed that an APRL hook could be controlled by EMG signals. Geddes, Moore, Spencer, and Hoff developed a similar system in 1959. Both systems were binary, i.e. the EMG signals opened or closed the hook, but had no control over the force or the velocity of the motion.

Later, a group of Russian scientists (Kobrinsky, Bolkoivin, Voskabrinikova, Ioffe, Polyan, Slavitskii, Sysin, and Jakobson) developed an EMG controlled artificial hand with proportional control. EMG signals from two control muscles, a prime mover and its antagonist, are amplified, integrated, and converted into pulses of frequency proportional to the level of the EMG signals. Pulses from the two channels are fed, push-pull, into a stepping electric motor. The direction of rotation and the torque exerted are controlled. A group at the Rehabilitation Institute of Montreal improved the "Russian hand". They added an adjustable wrist for passive rotation, located the amplifier inside the surface of the socket, and

replaced worn parts with American hardware for increased reliability. They also designed new amplifiers and battery chargers.

A group at the University of New Brunswick developed a three state myoelectric control system. In this system, two levels of activity are controlled by a single control muscle. R. N. Scott, also of the University of New Brunswick, worked on implanting electrodes through the skin. He found that a patient could wear such electrodes up to ten weeks with no ill effects. Scott also completed an extensive study of human capabilities.

In Cleveland, Dr. Charles Long, working with the Highland View Hospital and Case Institute of Technology, developed a novel control system for a paralyzed patient. Paralyzed muscles are stimulated by EMG signals. The system is controlled by EMG signals from a control muscle.

The Human Systems Design Center at Rancho Los Amigos Hospital in Downey, California, has made an extensive study of methods of locating electrodes. They have also developed an orthotic hand splint.

SYSTEM DESCRIPTION

Introduction to the System Operation

The system is described in block diagram form in Fig. 1. It employs an electromechanical actuator, which provides a translational motion, to operate the orthotic hand splint (see Appendix for a description of the splint used). The control signal for the actuator is derived myoelectrically from a control muscle which the patient can operate by his own nervous activity (see Appendix for a description of the selection procedure for this muscle). The electromyographic signal (EMG signal) that the electrodes detect from the control muscle is amplified, filtered, and integrated by the sensing and processing components, and the resulting signal, the integrator signal, is a d.c. voltage proportional to the level of electrical activity in the control muscle.

The integrator signal is supplied to the control system, which determines the system mode (see Special Features) and converts the integrator signal into a control signal, also proportional to the control muscle activity, for use by the actuator. The actuator employs a torque motor which supplies a force to the actuator linkage that is proportional to the control signal. The linkage transmitts the force to the hand splint, resulting in the closing of the hand. The degree of closure is proportional to the control muscle activity. As the patient exerts greater force with the control muscle, the hand is caused to close "harder."

Movement of the splint caused by the actuator generates a signal from a splint mounted strain gauge. The signal is to be used in the pressure feedback system (see Special Features) and is proportional to the grasping force applied by the patient. The signal is amplified and used to operate a voltage controlled oscillator. The oscillator signal supplies an elec-

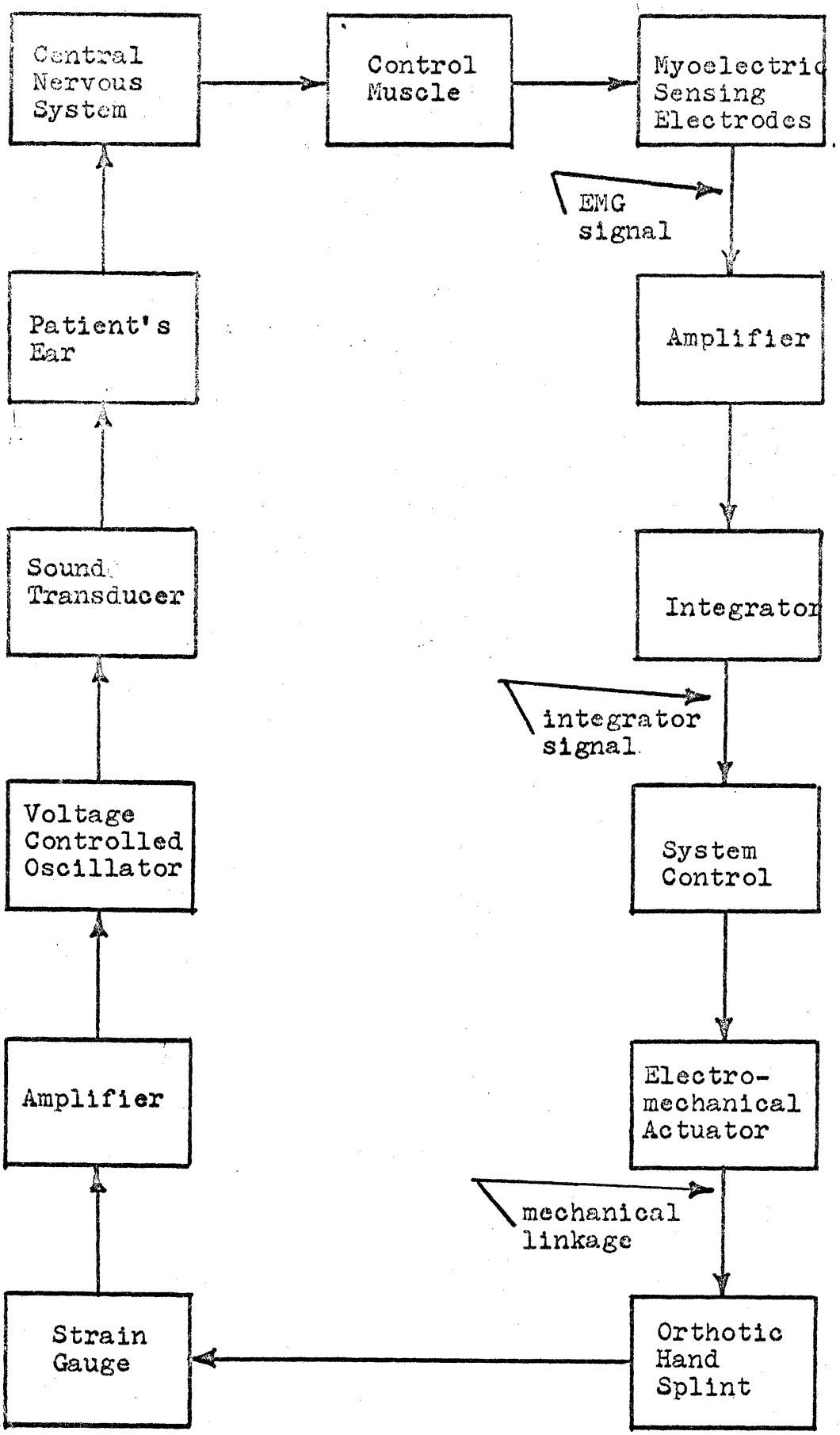

Fig. 1. System Block Diagram

tromechanical sound transducer which provides an audio display of magnitude of pressure being applyed to a grasped object.

Special Features

The orthotic system is a significant improvement over other available devices in four areas: restoration of proportional control of the hand, restoration of the sense of touch and pressure in the hand, low requirements of physiological energy due to patient selection of the system mode, and the efficiency, quiet operation, and compact size of the actuator.

1) Proportional Control: This feature concerns the design objectives of (a) restoring natural use of the hand by directly employing the patient's central nervous system for system operation, and (b) producing a device with a wide range of applicable patients. Normally, a person is able to control the grasping force he applies to an object by "mentally" determining the number of firing muscle fibers in specific hand muscles. The orthotic system restores this capability even though the specific muscles are not operative. A control muscle which the patient can control in a normal fashion is selected (see Appendix for selection criteria) and used as a source of control for the system. The patient is able to control the number of firing fibers in the selected control muscle, and as he increases the number, there is an increase in the electrical activity present in the muscle. This activity is sensed myoelectrically, and the resulting EMG signal is ultimately used to control the grasping force applied by the system. The patient has direct, proportional control over the system due to this intimate contact of the system and the nervous system.

Since many muscles in any part of the body can serve as a control muscle (with a minimum of re-engineering required), the range of patients is extended beyond those with only one paralyzed hand. Even if the patient

is totally paralyzed below the neck, a suitable muscle could still be found in the head region and used to control the system and restore use of the hand.

2) Restoration of the Sense of Touch and Pressure: This feature is one of the measures taken to meet the design objectives of giving the patient the ability to alter his method of grasp in accord with the object being moved and to employ the patient's control nervous system for control. The patient is provided with an electromechanical sound transducer located near or in his ear. When he causes the actuator to apply a force to the splint, he hears a constant frequency signal. An increase in force results in a proportionate increase in frequency. In this way, the patient can tell how much pressure is being applied to an object he has grasped by listening to the frequency of the signal change. The system is sensitive enough to also provide a sense of touch.

3) Reduced Requirement of Physiological Energy due to Mode Selection: In order to meet the design objective of minimum demand of energy from physiological sources (the control muscle), a quad-level control system was devised. Through this scheme, the integrator signal, which is proportional to the EMG signal, is used by the patient to select the desired system mode. There are three modes:

(a) Proportional forward: the system will close the hand to an extent proportional to the activity of the control muscle.

(b) Reversing: the system will open the hand at a preset rate not variable by the patient.

(c) Locking: the actuator will lock at a grasp pressure and position determined by the patient, and the system will require only a slight signal from the control muscle to hold that position.

The method by which the proportional signal is used to select the mode is ahown graphically in Fig. 2. The modes are determined by the level of the integrator signal. When the integrator signal amplitude is at a level between "c" and "b", the system will be in the proportional forward mode. After the desired amount of grasping force has been achieved in the forward mode, the force maintaining mechanism can be transformed from the motor to the actuator brake by quickly reducing the magnitude of the integrator signal to some level within the locking region. If the integrator signal is diminished below "d", the pressure feedback audio signal is interrupted, which in turn warns the patient of the proximity of the reverse region. This warning signal can also serve as a reference in the locking region since the operator could hold the integrator signal about this point, thus assuring himself that he is in the locking mode. In the reverse region, the orthesis opens at a predetermined speed until termination is desired. The device can be opened to its extremity, where opening is terminated by a limit switch in the actuator, or opened to some arbitrary position determined by the length of time the reversing signal remains in the reversing region. When the muscle is relaxed, low level involuntary muscle contractions will have no effect on the actuator since the splint is open when the muscle is relaxed and thus isolated from control by the actuator limit switch as long as the integrator signal level is less than "c". Therefore, there is also locking in the reverse region which is achieved when the splint is opened to the actuators limit and is abolished when the integrator signal level becomes greater than "c".

4) Special Features of the Actuator: The actuator meets many of the design objectives. It consists of a torque motor to be operated by the control signal, a fail-safe brake to provide the locking mode, and a ball screw to convert the rotational motion of the motor to translational motion

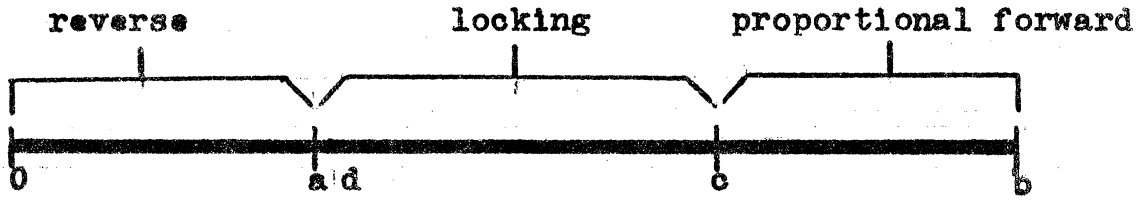

Fig. 2. Integrator signal levels for mode selection.

required to operate the splint. The actuator is encased in a nylon tube, and the actuator linkage is constructed of two concentric plastic cables. These components offer three specific improvements:

(a) The ball screw and motor are very efficient so that the power needed for splint operation is low. Because the brake does not consume power when it is in the locking mode, the power requirements of this mode are negligible.

(b) The noise level is low because the torque motor runs at a low speed, there are no gears, and the actuator is encased in nylon, a material with little capability for sound transmission.

(c) The actuator and linkage is cosmetically pleasing. The nylon and plastic have a neutral color, are without luster, and are warm to the touch. Because the motor, brake, and ball screw are small and are arranged in a linear fashion, the cylindrical package can be inconspicuously worn on the upper arm or the waistline.

CONCLUSION

The design group has developed an orthotic device to perform the work of incapacitated muscles, simulating the natural behavior of human muscles. The device replaces the muscles which close and open the hand. The conventional electromechanical device for this purpose consists of an orthotic splint matched to the hand, a motor, a coupling between the splint and the motor, and a control system. The conventional control system has sensed myo-electric signals and given digital commands to the motor, telling it to open or close the hand.

The newly developed device has three major improvements. The device has forward proportional control; the patient can control the force with which the device closes his hand. The device has an audio feedback system by which the patient is told how much force is being exerted through his fingers. The device has an actuating unit vastly improved in performance and appearance.

The proportional control enables the patient to control the rate at which the hand closes, and the force with which the hand is held closed. This adds a degree of freedom to users of orthotic devices. They can hold fragile as well as heavy objects in their fingers.

The audio feedback tells the patient with how much force he is closing his hand. A tone, of frequency proportional to the force

against the tip of the index finger, is sent to his ear. The audio signal is not unpleasant. It neither interferes with normal hearing nor requires undue attention of the patient. Audio feedback in a nonproportional system would be nearly worthless; in a proportional system, the feedback, or a substitute providing the same information, is absolutely necessary.

The conventional actuator consists of a motor and cable. The motor winds the cable around its shaft to close the hand. The actuator is T-shaped because the driving coil is necessarily perpendicular to the axis of the motor shaft. The actuator developed by the design group consists of a motor, ball screw, and cable. The ball screw converts the motor shaft rotation into a linear motion parallel to the axis of the motor. The force output of the actuator is proportional to the torque of the motor. The motor is active in both opening and closing the hand; no return spring is necessary. The ball screw is very efficient, and the actuator unit has excellent time response. The actuator is cylindrically shaped, making it much better looking than the conventional model. Also, its operation is quieter than that of existing actuators.

The product of the design group is a working prototype. This prototype has not had extensive testing. It has been tested neither for reliability nor for use outside of laboratory conditions. It works in the laboratory, however; it is not known

how long it will work, or if it will work where there are adverse weather conditions. Obviously, the next step in its developement is to make it reliable and manufacturable.

There are clear needs for future developements in this area. Better methods of sensing EMG signals are being developed. Proportional control systems with better time responses are needed. Better patterns of audio signals for feeding back information must be found. Quieter actuators with smaller motors and less friction are possible.

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Appendix A

SENSING

The sensing consists of amplifying and processing electromyographic (EMG) signals from the control muscle. In the muscle, electrical pulses accompany muscle activity (tension). Each pulse acts through a single motor unit. The motor unit is a group of muscle fibers attached at each end to other parts of the body, and actuated through a single nerve cell. An electrical pulse through the nerve cell causes the fibers to contract momentarily. A muscle contains many of these motor units. To maintain tension, the motor units must be continuously excited. Alter and Basmajian both mention that, at lower levels of muscle tension, individual EMG pulses can be observed. At this low operating level, an increase in tension is accompanied by a greater frequency of EMG pulses, with an insignificant variation in the magnitude of the signal. At higher levels of muscle activity, the pulses superimpose, resulting in an increase in the magnitude of the EMG signal to levels several times that of a single pulse.

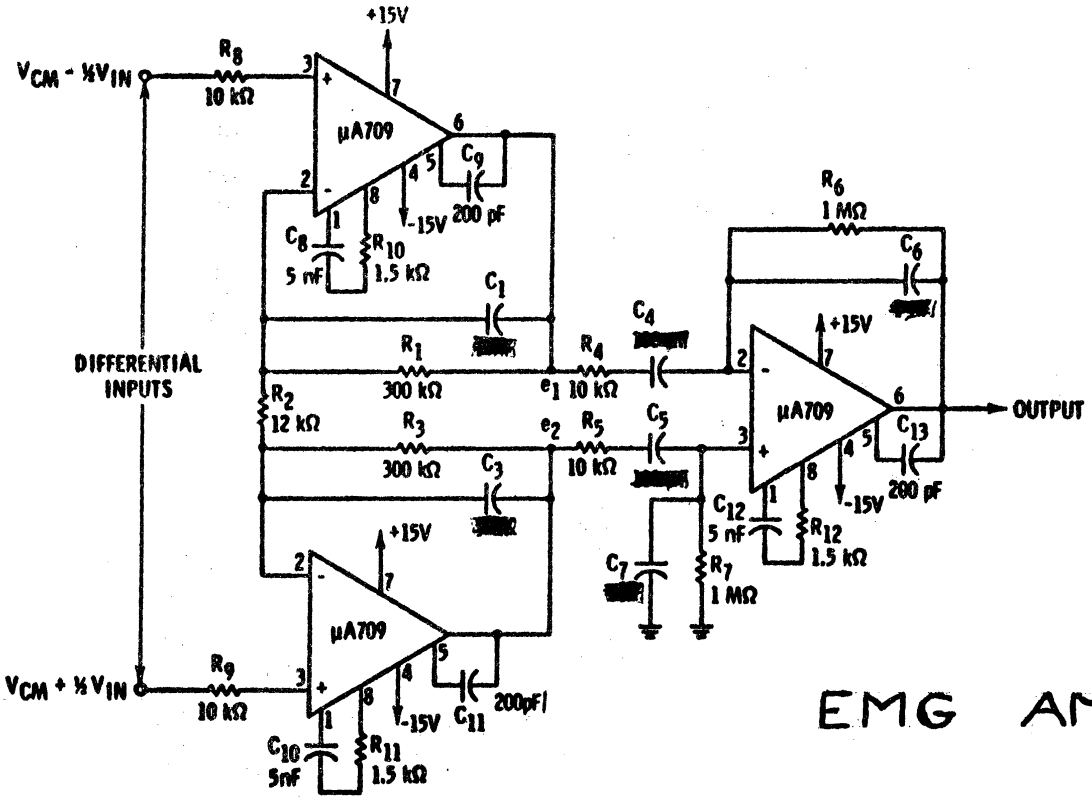
Amplifier

A. Development - The EMG amplifier both limits and amplifies the low level signals from the skin electrodes. Most of the energy in EMG signals is concentrated in the band of 30 to 300 cps. The amplifier ought to operate only in this bandwidth, in order to minimize noise in the output. Because the maximum amplitude of the EMG signal is near 1 millivolt, considerable care must be taken to keep 60 cps hum from the input circuit. The traditional method to accomplish this is to use a differential amplifier, on the assumption that the greater part of any hum detected will appear equally, in phase, at

both input electrodes. Also, a third electrode connection is commonly made between the skin and the amplifier ground in order to minimize the hum.

An early attempt at a practical amplifier was made using a Fairchild uA709 integrated circuit operational amplifier connected in the differential configuration. The circuit did work, however; it was severely plaqued by two problems. First, the input impedances were low, on the order of a few thousands ohms, with the consequence that a slight imbalance in electrode impedances would cause much hum in the output. This problem proved to be so severe that an earth ground was an absolute necessity. The second problem was that a slight D.C. offset voltage appearing between the skin electrodes would drive the amplifier into overload unless a balancing adjustment were touched up almost continuously.

This experience made it quite evident that both a very high imput impedance and the ability to tolerate a reasonably high D.C. offset voltage between the electrodes were highly desirable. A circuit found in the Fairchild Integrated Circuits Handbook looked very promising. This circuit features an ingenious arrangement of two uA709's as a differential input stage which has common mode and differential input impedances of about 100 megohms. A second stage composed of a single uA709 raises the overall gain to 5000 and converts to a fairly low impedance, single ended output. This circuit, when tested in our laboratory, was found to perform so well that operation completely floating from an earth ground causes no problems whatsoever. Also, the first stage is run at a sufficiently low gain that the unit can tolerate up to 100 millivolts of D.C. at the input without overdriving. This circuit was adopted.



EMG AMPLIFIER

B. Final Design - The only critical requirements for the circuit are that R4 be matched to R5, and that R6 be matched to R7. It appears that the use of a pair of small trimmers will be necessary here, to adapt the capacitor values to our necessary frequency responses. To adjust the trimmers, tie the input leads together and connect their junction to ground through a resistor of a few thousand ohms. Apply a signal of 60 cps and 100 millivolts across the resistor. Adjust the two trimmers for minimal signal in the output; this amounts merely to balancing the amplifier for best common mode rejection. It is necessitated by resistor tolerances. The low frequency limit is determined by the relationships of C4 to R4 and of C5 to R5. Use the following formula:

$$C_{4,5} = \frac{1}{20{,}000\,\pi f_L}$$

where the capacitances are in farads, and f_L is the desired lower cutoff frequency. Cutoff at the high end is adjusted by C1, C3, and C6:

$$C_{1,3} = \frac{1}{600{,}000\,\pi f_H}$$

$$C_6 = \frac{1}{2{,}000{,}000\,\pi f_H}$$

where f_H is the desired upper cutoff frequency.

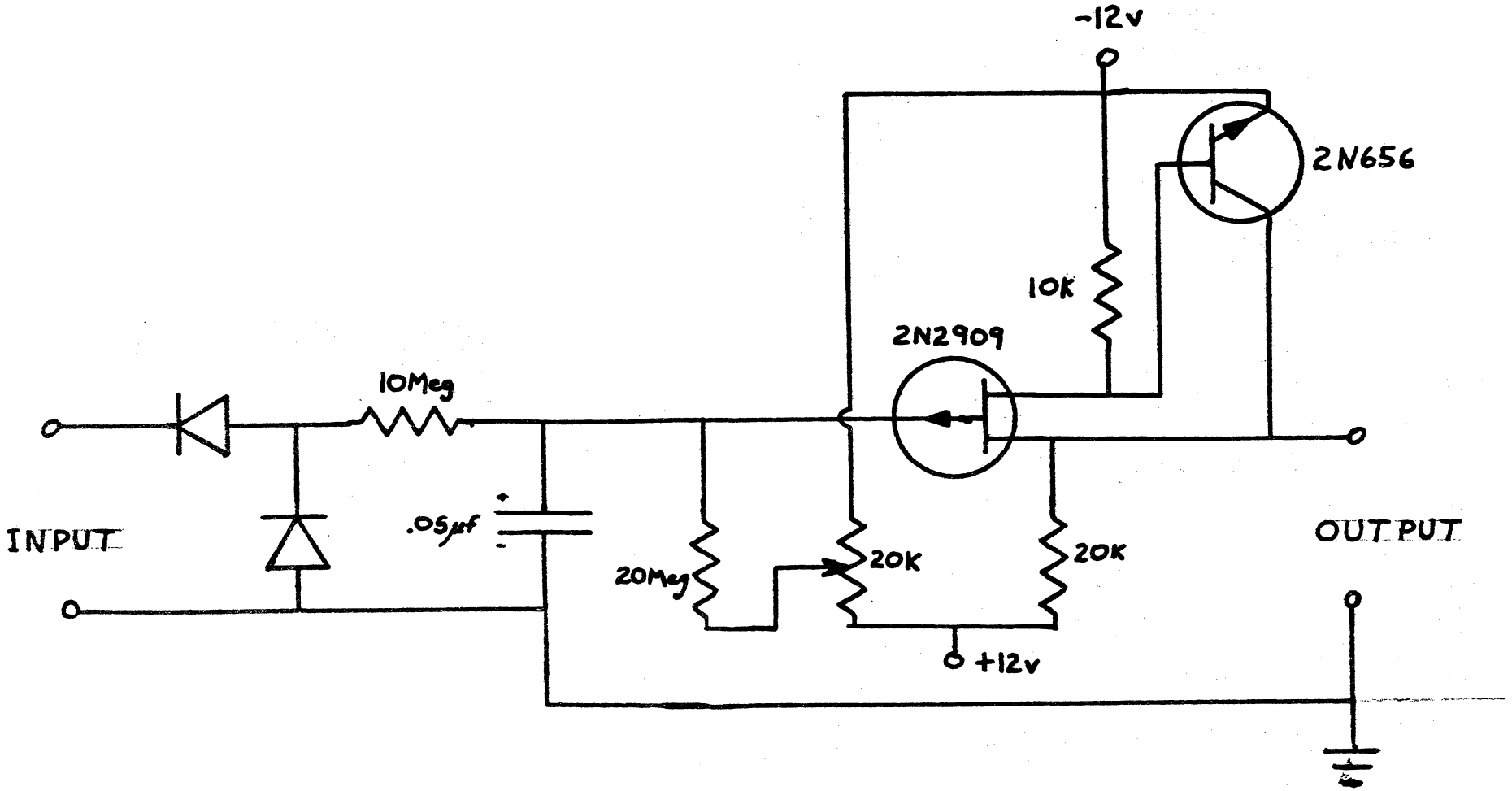
Signal Processor

A. Developement - At low levels of muscle activity, EMG pulse frequency is proportional to muscle tension; at higher levels of muscle activity, the amplitude of the EMG signal is significantly greater. At least at low

tension some form of frequency detection seemed possible. Unfortunately the body, a volume conductor, attenuated the pulses and yielded pulses of many different amplitudes which are inversely proportional to the distance from the motor unit to the sensing electrode. A similar situation is encountered by nuclear gamma ray activation analysis equipment. In that situation, the pulses are routed through a discriminator that produces a uniform output pulse for every input pulse which satisfies a certain amplitude requirement. This uniform output pulse is then fed to scalars or counting meters. Such a system approach could be used in our situation. An advantage would be the low level of muscle activity required. Disadvantages would include a very high muscular gain; a slight change in muscle tension causes a large change in pulse frequency. Even while the muscle is relaxed, some EMG pulses are generated. To accommodate this situation, we would have to narrow the usable bandwidth. Also, the electronics for such a system would be complex, and expensive. Such an approach has possible merit, and ought to be investigated, as should a combined frequency and amplitude analysis approach.

The output of the sensing processor was to be a voltage proportional to the level of the muscle activity. Three primary methods were proposed. The first was a simple bridge and filter. The desired output currents of 2 to 3 milliamps demanded capacitor values far too large, both from the standpoint of the physical size of the component and from the standpoint of response time; the time constants could not approach the required .2 to .5 seconds.

Next, a common collector transistor amplifier was attached to the output of the filter. This was a vast improvement but still did not offer


-12v
2N656
10K
2N2909
10Meg
INPUT
.05μf
20Meg
20K
20K
OUTPUT
+12v

INTEGRATOR

enough output isolation from the filter element, even with very high current gains. 1 to 2 per cent ripple was present at required current, voltage, and time constant.

B. Final Design - The third, and accepted, approach involves the use of a field effect transistor (FET) to sense voltage on the filtering capacitor. This allows very high current isolation from the components which determine the time constant. The high impedance also allows a smaller, in physical size and in capacitance rating, capacitor to be used. Time constants can easily be changed, and the imput-output characteristics can be modified with facility. The low current handling capability of the FET is bolstered by a current supplying transistor. Due to the offset voltage of the FET, a full wave bridge is no longer possible, and a half wave voltage doubler, has been substituted with no sacrifice in quality. Ripple on the output is .1 per cent at full output.

CONTROL SYSTEM

The control system receives a d.c. voltage signal, proportional in amplitude to the level of the EMG signal, from the sensing processor. It interprets this control signal, and accordingly controls the actuator. The control system also contains the power source for the actuator.

The level of the input signal to the control system is proportional to the EMG signal generated by the control muscle. Thus, the level of the input signal is controlled by the user. The control system must control the actuator according to the level of this input signal. System requirements for the orthetic device include three modes of operation: forward, reverse, and lock. According to the control signal, the control system determines the mode of operation.

The forward mode is that in which the actuator closes the hand. In order to simulate the natural function, the force with which the hand is closed should be proportional to the tension with which the user flexes the control muscle. A d.c. torque motor supplies the actuator force; this force is proportional to the torque developed by the motor. The torque developed is proportional to the torque developed by the motor. The torque developed is proportional to the current supplied to the motor. When the motor is stalled, i.e. when the hand can be closed no farther, the current is proportional to the voltage across the motor input terminals. Thus, by supplying a voltage proportional to the control signal, the control system causes the actuator to supply a force proportional to the tension in the control muscle. Note that the force if proportional only when the actuator is stalled. When the hand is closing, before it has encountered any resistant force, the rate of closing is proportional to the d.c. voltage across the motor input terminals.

Thus, the system with proportional control in the forward mode will close the hand at a rate proportional to the tension in the control muscle, and, after closing the hand, will apply force proportional to the tension in the control muscle.

The reverse mode is that in which the actuator opens the hand. There is no necessity for proportional control in the reverse mode; it is sufficient to open the hand at a predetermined rate. The control system can cause the actuator to open the hand by applying a negative d.c. voltage across the motor input terminals. The rate of ppening is proportional to the amplitude of that voltage. The control system enters the reverse mode when it receives a certain control signal; it then opens the hand at a predetermined rate.

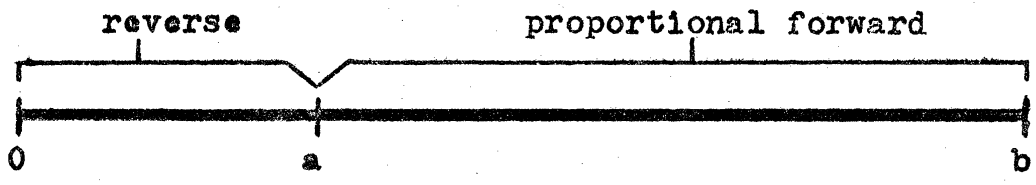
The lock mode is that in which the actuator maintains the hand in a certain position. When the user desires to maintain a certain force for an extended period of time, locking the position of the hand eliminates the necessity for a continuous supply of energy. There is a solenoid released brake in the actuator assembly; when the solinoid current is interrupted, the brake is applied. The control system inters the lock mode by interrupting the solinoid current and shutting off the motor supply. A possible danger with such a locking system is that the user could unknowingly harm his hand by applying pressure to long. A locking system which requires control muscle activity would discourage prolonged locking, just as muscle fatigue and physical pain discourage overloading in natural activities.

In the "off" position, the hand will be fully open. The control system will be inactive until a control signal calls for the forward mode. The forward mode control signal must be high enough so that involuntary EMG signals do not actuate the device.

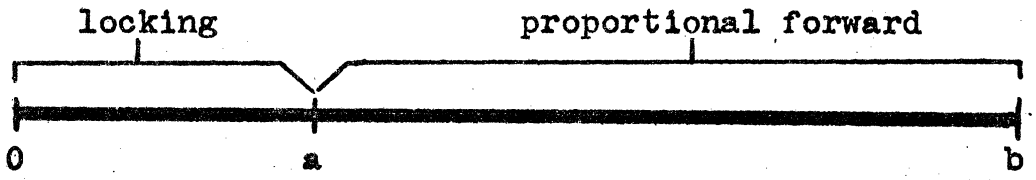
A. Development

One possible design (Fig. 3) would divide the control signal range into two regions. A control signal ranging from "o" to "a" would demand the reverse mode; a control signal above "a" would demand the forward mode, and; a control signal in the forward mode region, held nearly constant for a specified time, would call for the lock mode. The opening process, in the reverse mode, could be terminated in either of two ways: a limit switch in the actuator would turn the device "off" when the hand had been opened to its rest position, or an increase in the magnitude of the control signal would change the mode from reverse to forward. The lock mode would be abandoned whenever the control signal strayed a certain amount from its steady level. The circuitry required for the lock mode would be very complex, and expensive. Also, it would be very difficult for the operator to maintain high control muscle activity.

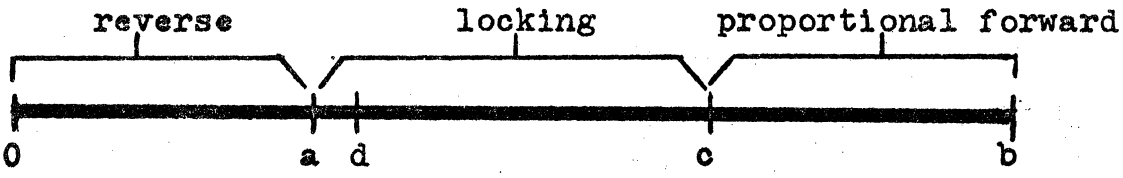
A second possible design (Fig. 3) would operate the lock mode when the control signal was between "o" and "a", and would enter the forward mode when the control signal was greater than "a". After the user had established his output force, in the forward mode, he could go into the lock mode by quickly dropping the control signal into the lock region. The brake would then lock the position of the device before the power supply to the motor was terminated. Conversely, by slowly dropping the control signal level within the forward mode range, the user could proportionally reduce the output force. The reverse mode would be entered whenever the user sharply tensed, the relaxed the control muscle. The resulting control signal would be distinguished by having a time derivative greater than a specified value. Again, the opening process would be terminated whenever the rest position was reached or the control signal entered the forward mode region. This design is not feasible; it would be impossible to distinguish the EMG pulse for reverse mode from normal EMG signal fluctuations.



First Proposal



Second Proposal



Final Scheme

Fig. 3. Development for the control modes.

A four-level control scheme (Fig. 3) employing many aspects of the above systems, was finally chosen. From "o" to "a" is the reverse mode region, from "a" to "c" is the locking mode region, and from "c" to "b" is the forward mode region. As in the second proposed design, the user would employ the locking mode by dropping the control signal from the forward mode region into the locking mode region. In the locking mode region above the level "d", the audio signal informs the user how much force is being applied through the hand. Below level "d", the audio signal is cut off. Thus, "d" is a reference level, easily distinguishable, by which the user can consciously maintain the locking mode. The limit of the hand's opening can be determined by a position switch in the actuator assembly. After the position switch is engaged, the device is "off".

B. Final Design

The control circuit includes the mode control and the power supply for the orthesis. The mentioned voltage levels are nominal. The control signals range from 0 to 5.0 volts. 0 to 0.5 volts is the reverse mode region, 0.5 to 1.0 volts is the locking mode region, and 1.0 to 5.0 volts is the forward mode region. Schmidt triggers are used to detect the level of the input signal (Fig. 4).

By adjusting R_1 and R_2, V_r can be set to the desired value for each of the Schmitt triggers. Integrated circuit differential amplifiers are used for each Schmitt trigger. Two triggers, ST1 and ST2, are used for mode control. V_r for ST2 is set at 0.5 volt, and V_r for ST1 is set at 1.0 volt.

When V_{in} 0.5 volt, the outputs from ST1 and ST2 are -12 volts. This causes Q9 and Q10 to conduct, which cause Q6, Q7, and Q8 to conduct. This, in turn, causes Q4 and Q1 to conduct, and Q13 to be open. Current flows from -12 v., through Q1, the motor, and Q4, to ground. Current in this

direction through the motor causes the hand to open. Also, current flows through D2 to the brake solenoid, releasing the brake.

When V_{in} is between 0.5 and 1.0, the output from ST2 is +12 v., and the output from ST1 is -12 v. This causes Q9 to conduct, and Q10 to be open. Then, Q1, Q3, and Q4 are off. No current is supplied to the power circuit, leaving the brake locked.

When V_{in} is between 1.0 and 5.0 volts, the outputs from ST1 and ST2 are +12 volts. This causes Q9 and Q10 to be open. Q9's being open causes Q13 and Q3 to conduct. Also, 1.0 volts will break down the .74 volt zener diode, turning on Q5 and Q2. Thus, there is a current path through Q3, the motor, Q2, and to ground. This current is proportional to V_{in}; it causes the splint to close. Also, current through D1 to the brake solenoid releases the brake.

Transistor Q7 was added to prevent Q6 and Q13 from turning on simultaneously. If they were to do so, Q3 and Q4 would burn out.

Schmitt triger ST3 is used as a warning device. V_R for ST3 is set one-tenth volt avove V_R for ST2 by means of R_3. When V_{in} drops to .6 v., the output from ST3 switches from -12 to +12 volts. This is used to warn the patient that he is nearing the bottom of the range of voltages which lock the splint, and is about to go into reverse.

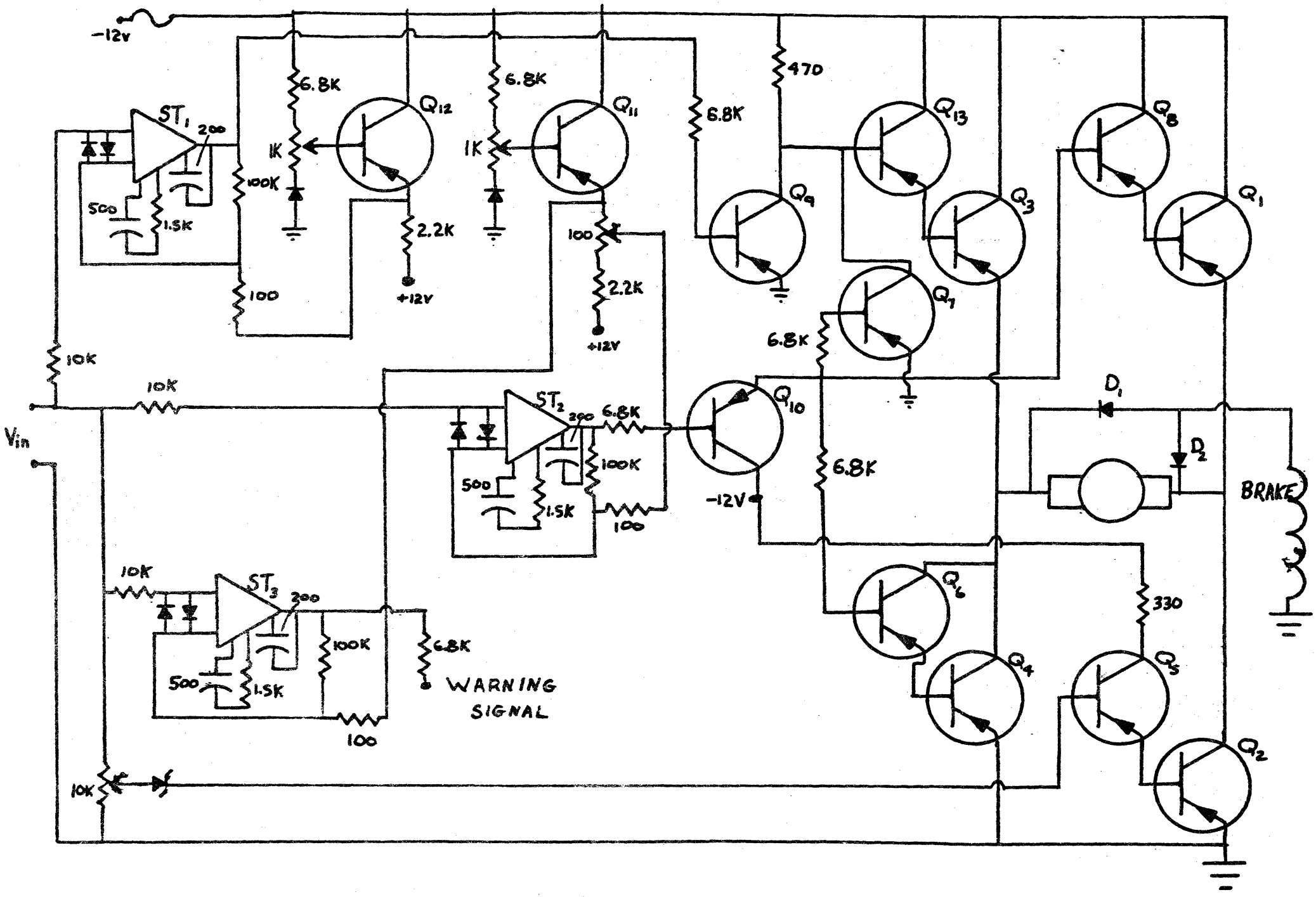
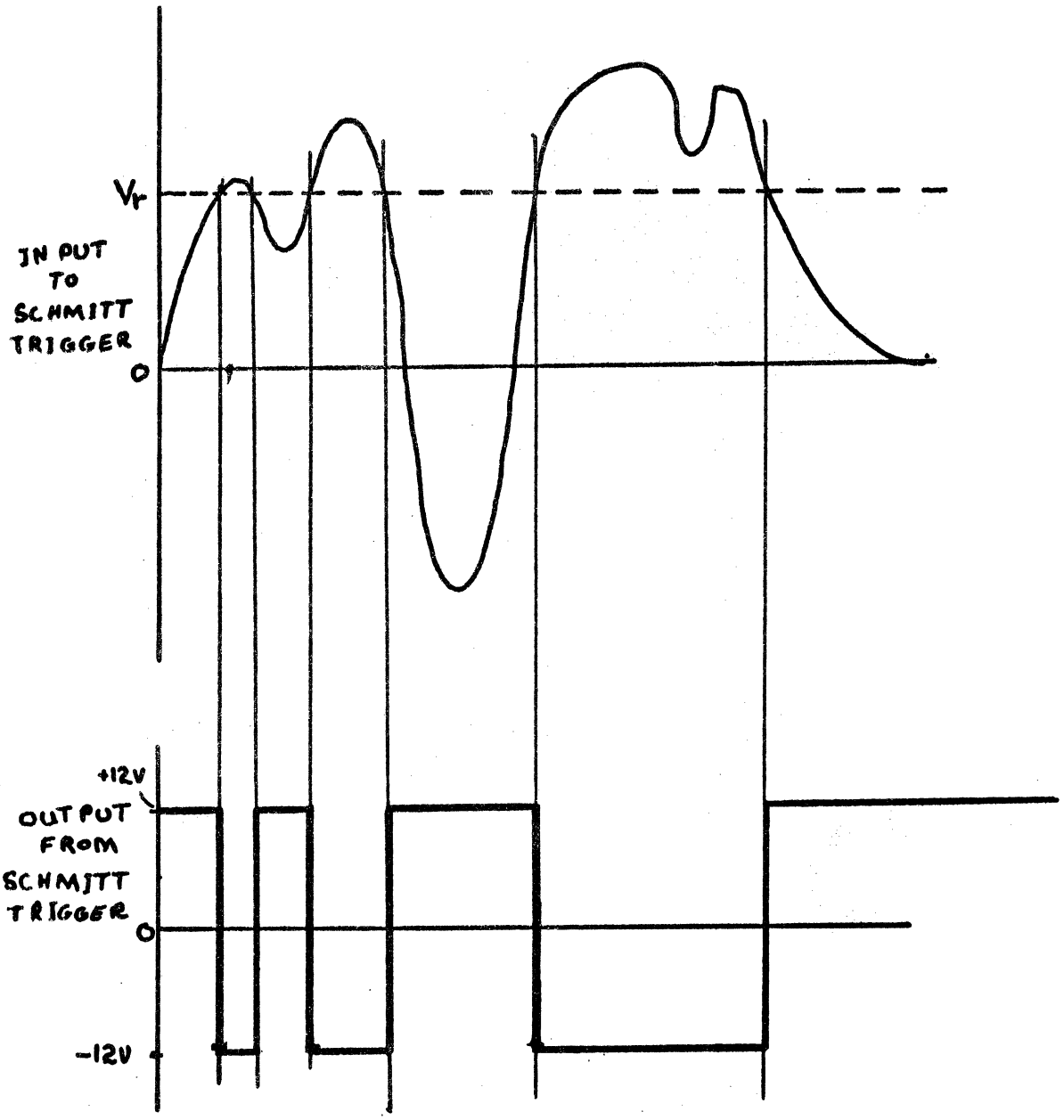


FIG. 4 CONTROL CIRCUIT



TRANSFER FUNCTION FOR SCHMITT TRIGGER

TRANSISTORS FOR CONTROL CIRCUIT

Q_1	2N178	Q_8	2N1303
Q_2	2N178	Q_9	2N1303
Q_3	2N178	Q_{10}	2N1303
Q_4	2N178	Q_{11}	2N1303
Q_5	2N1303	Q_{12}	2N1303
Q_6	2N1303	Q_{13}	2N1303
Q_7	2N1303		

DEVELOPMENT OF THE ACTUATOR

Mechanical simulation of one degree movement such as prehension involves problems ranging from physiology to cosmetics. The patient must be made mentally as well as physically confortable with his appliance; a review of currently available mechanical actuators reveals dissatisfaction. In general, present methods involve large noisy units which are essentially on-off, fixed force level devices.

It was decided to first view the actuator as a black box whose input was of necessity electrical and whose output was a translational motion of the output cable with a force level proportional to the electrical input. Further, this black box was to be as free as possible from the problems inherent in present systems. Analysis showed that in order to have this desired result, the actuator should have the following description:

1) Size - Although the existing unit was designed for an orthetic device, we would like the size to be such that it could be inserted in a prosthetic appliance.

2) Light weight - While designed as a "belt-pack", the unit, possibly with its control device, could be, as previously mentioned, inserted in a prosthesis which should approximate the weight of the limb it replaces.

3) Low noise - Users took great displeasure with the "robot" sound.

4) High force level - Some of the units studied had trouble in firmly grasping a telephone receiver.

5) Speed - The actuator should close the hand at a speed simulating the human muscle counterpart.

6) Efficient - Reducing dissipated power reduces size, weight, and cost of the entire unit.

During the calculations of these requirements the group concluded that they wanted a cylindrically shaped device whose axis was collinear with force direction. Also, it should weigh less than half of a pound, and should deliver ten pounds or better over a nominal one and a half inches in less than one half of a second. Furthermore, the force level should be directly proportional to the applied input.

The requirements dictated the investigation of hydraulic, pneumatic, and electromechanical linear actuators. Hydraulic linear actuators are inherently quiet, fast, powerful, and efficient. But for the force levels involved, the pumping losses would be too great, not to mention the extremely high cost for the subminiature size components desired.

As an alternative to a recirculating system, a system using a charged tank under pressure and a bleed tank was considered. This would eliminate the noise and cost of constant pumping. However, the number of operations would be limited by the tank size, since the output force requires constant flow.

Pneumatic systems, either pumped or compressed gas types, were discarded for the same reasons as above, plus the effects of compressibility on force level, and the interfacing required.

Thus the group turned to electromechanics since in this system input electrical interfacing problems are reduced, and with electrical motor efficiency being fairly high the major problems seemed to be mechanical. This greatly simplified the design problem.

Again, we reviewed the existing systems to pinpoint specific problems. The typical system consists of a high speed (8000 rpm and up) series wound d.c. motor turning a planetary, rack and pinion, or other gear train, which

in turn pulls a Bowden cable against a spring loaded splint. This system has inherent problems. First, the high speed motor has excessive commutating noise in addition to the noise of the high speed gears. The high speed is necessary to gain torque multiplication in the gear train, since a series motor will burn up if it is required to supply power near stall speed. Additionally, this type of motor has a non-linear current-torque characteristic, thus making it undesirable for a proportional system. From a packaging standpoint, this system is bad since force is transmitted perpendicular to the motor shaft. Finally, the force is applied in one direction only, requiring a restoring force in the opposite direction supplied by a spring.

The first solution suggested involved a pair of differential solenoids acting in opposition and operating the splint through a cable. The primary disadvantage of this system is the amount of power necessary in the coils at all times.

Next we considered an a.c. hysteresis-synchronous motor. This motor has a high starting torque, is efficient in the stall mode, and, having no brushes, is considered to be very reliable and silent.

Concurrently the conversion of torque to force was studied. Since the group was considering high torque, low speed motors (around 3600 rpm) high gear ratios would not be necessary, thereby reducing friction and inertia in the system. At this time, the advantages of recirculating ball screws manufactured by Beaver Precision Products were brought to our attention. These devices are normally better than ninety percent efficient. They are quiet and fast. With the sixteen turns-per-inch screw, thirty-two revolutions of the motor produces a two inch translation. Ball screws transmit a force parallel to the axis of rotation, consequently cylindrical shape is possible.

As this point three new problems were presented. First, the control and dc to ac conversion circuitry for the ac motor appeared to be unduly complicated. Next, the ball screw, being extremely efficient, would be reversible unless power to the motor or some other type of position locking mechanism was provided. Finally, a method was needed for symmetrically applying load to the ball nut which also had to be constrained from rotation. Eccentric loading of the ball nut would cause decreased efficiency and somewhat increased wear. Whatever method was provided for constraining rotation of the ball nut could not prevent symmetrical loading of the nut by the cable, which exits parallel to the ball screw.

We were again fortunate in having brought to our attention a commercially available item well suited to our needs. "Inland Motor of Virginia" builds a one inch cube d.c. permanent magnet servo motor, which has several desirable characteristics. These motors, available in a variety of voltage ratings, have no-load speed and stall torque versus voltage curves which are essentially linear. Thus, in operation, given an applied voltage the motor will run in at a speed proportional to applied voltage until it stalls at a torque also linearly proportional to the input. The motor has little inertia, is very efficient, can deliver up to seven inch ounces, and can be run in stall mode safely. With the torque output available at low speeds, we were able to utilize relatively course pitch ball screws to produce forces in excess of our requirements. We found that with this motor and a representative sixteen pitch ball screw, forty pounds could be developed, but hand closure was too fast. We decided to run this motor at less than rated voltage to obtain reasonable closing rates and yet still produce better than twelve pounds force.

Rather than continuously dissipating power in the motor, the group decided to utilize some mechanical method of locking the ball screw. The first proposed solution involved a solenoid ratchet device, such as an escapement for model airplanes. But an escapement, or a pawl type brake proved unsatisfactory. The discrete steps available would not permit fine enough force increments. An attempt was made to adapt a model plane electric disc brake, but its operation was exactly opposite to that desired. As envisioned, power would be cut simultaneously to motor and brake, which, being spring loaded, would lock the system with no power consumption.

American Precision Industries in New York was able to supply us with a twelve volt d.c. version of their BFR-5 spring-loaded, friction type failsafe mechanism which mounts to the motor with a machined adapter plate. Flats ground on both the motor and ball screw shafts mate in a brass sleeve coupling which rides on these shafts within the brake body.

It appeared convenient to combine the symmetrical loading of the ball nut with a means of rotational constraint such as a yoke and trunion assembly guided by slots on the inside of the case. However, modifying the ball nut to accept two or more trunions proved to be difficult. In addition, a thrust bearing was needed to prevent motor bearing damage, and eliminate binding of the ball nut due to shaft deflection. The combination of these two functions could be accomplished in one of two ways. It was possible to attach the yoke to the ball nut with a clamp, passing the yoke arms through locating slots in the case end plate which would double as the bearing support. Alternately, the case might be extended to contain all movement of the yoke, mounting the thrust bearing along the inside of the case. This method was not only cosmetically preferable but would also prevent external interference with yoke motion.

While receiving a hypodermic injection, one of the members of the group was reminded of the ball screw by the plunger motion in the syringe. It was found that a 10 c.c. plastic disposable syringe would fit over the ball screw and was close to the diameter of the ball nut. An appropriately scaled brass replica of the syringe was fabricated, slotted to clear the bearing support, and clamped to the ball nut. The opposed slots ride on the bearing support thus constraining rotation of the ball nut. The "needle end" of the replica was tapped to accept the output cable fitting (see actuator assembly drawing). The 10 c.c. itself is being tested to determine its wear characteristics and may be utilized.

While investigating the model airplane parts previously mentioned an interesting radio control cable was found. This cable consists of a 3/16" outer nylon cable fitted with a 3/32 inner nylon cable. Both ends of the inner cable were threaded to accept 3-48 screw stock. The cable has been tested at fifteen pounds pull and six pounds push with strength left to spare.

Packaging was the next concern. The brake dissipates four watts while energized. The motor is operated at less than half of its rated maximum power and, therefore, only gets slightly warm. Nevertheless, these elevated temperatures transmitted through a metallic case might tend to cause uncomfortable hot spots on the actuator case.

To this end, a nylon case was selected. The prototype has been machined. The possibility of casting later models is being investigated. The case is a rigid tube housing the entire actuator assembly, including a travel limiting switch fashioned from Switchcraft contacts. It is fitted with two end plates sealed with O-rings and screws. The actuator could be made waterproff by dipping in silastic.

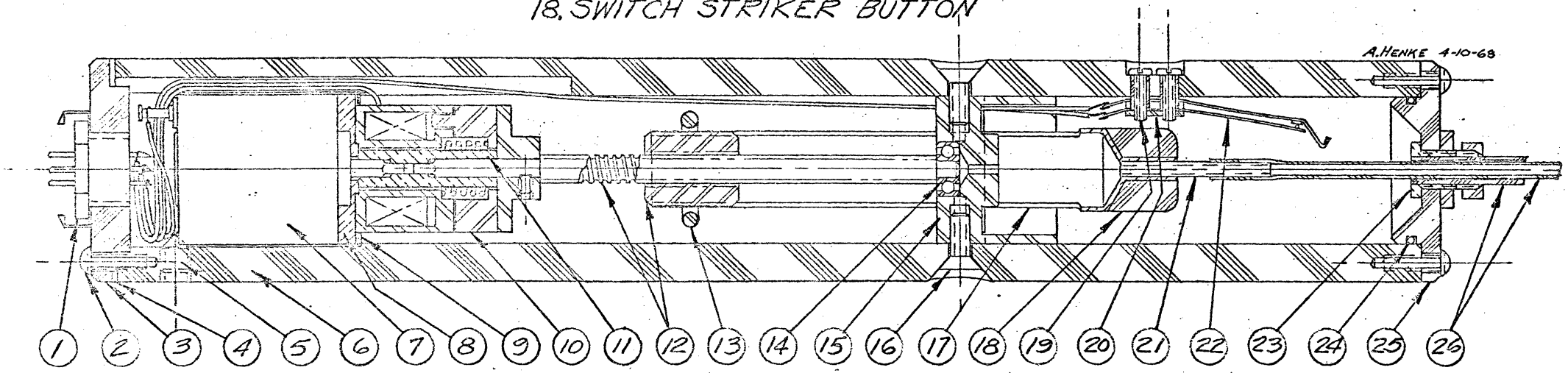
The rear end plate houses a miniature seven-pin connector carrying all the power and control leads. The front end plate is fitted with a suitably modified 1/8 inch shaft lock for restraining the outer cable; the inner cable is connected internally to the syringe through a piece of 3-56 threaded stock.

Final Design

The completed actuator is 8.93 inches long, weight 7.5 ounces, is 1.5 inches in diameter, takes a 0 to 12 d.c. volt input at the back and emits a 0 to 12 pound force at the front. Noise is below present systems and may further be reduced by plastic bearing supports and the use of the plastic syringe. The device will pick up a twelve pound weight two inches off a table in one half of a second. It seems to have met or exceeded all initial requirements.

Further experimentation is planned in the area of materials to cut both cost and noise below already acceptable levels. And with the announcement of a new smaller servo motor by Inland, we believe we can shrink size without degrading performance.

1. CONNECTOR - MIN. 7 PIN MALE
2. SCREW #3-48 × 3/8 LG. RD. HD STAINLESS ST'L - 8 REQ'D
3. TAIL END CAP
4. GASKET
5. SET SCREW #10-32 × 3/16 LG. CONE PT. S'K'T - 2 REQ'D
6. CASE
7. MOTOR (REFER TO TEXT)
8. ADAPTER
9. SHIM (NO. REQ'D VARIES)
10. BRAKE (REFER TO TEXT)
11. COUPLING
12. BALLSCREW & NUT 3/16 DIA. 16 PITCH (SEE TEXT)
13. CLAMP - 1/2" RING HOSE
14. BALL B'R'G - NEW DEP. #77R2
15. BEARING SUPP. & GUIDE
16. SCREW #6-32 × 3/8 FL'T. SKT. HD.
17. SLEEVE
18. SWITCH STRIKER BUTTON
19. SCREW #2-56 × 5/16 PAN HD. STAINL'S
20. CLAMP PLATE
21. ROD SCR. #3-48 × 1" STAINLESS
22. LIMIT SWITCH - MAKE FROM SWITCHCRAFT # H83P
23. CABLE COLLET - INCL. NUTS MAKE FROM 1/8" MIL. POT SHAFT L'K
24. O-RING SEAL
25. HEAD END CAP
26. ACTUATING CABLE



ACTUATOR

SCALE 1=1

FEEDBACK

A. Developement

Existing orthetic devices do not inform the user how much pressure or force is being applied. It would be manifestly desirable to incorporate a "feedback" system into the device, in order to better simulate the natural function.

The feedback information ought to be easy to interpret and should interfere with other capacities as little as possible. Ideally, the feedback system would be of high enough sensitivity to provide a "sense of touch," as well as to distinguish magnitudes of force up to fifteen pounds. Also, the cosmesis of the feedback system is a consideration.

The first design problem was the method of measuring the force applied by the hand. The appropriate location for the measurement would be near the hand. Measurements at other locations (eg. actuatro linkage, EMG level, etc.) would be less direct, and would consequently yield lower sensitivity. A semiconductor strain gauge mounted on the index finger brace of the hand splint was selected as the information source.

The nature of the information transmitted to the user had to be determined with a good deal of care. Optical, electrical impulse, or skin transducer systems would require conscious attention, they would be insensitive, and they would interfere with other functions. A system employing an audio signal could provide high sensitivity, while requiring little conscious effort from the user. In this system, the strain gauge varies the frequency of a voltage controlled oscillator. The oscillator output voltage is converted to an audio signal of the same frequency by a crystal earphone, and the resulting audio signal is transmitted by low

cross section tubing. The frequency of the audio signal is proportional to the force in the hand. The size of the tubing permits normal hearing. The device is similar in appearance to certain types of hearing aid transducers. The ideal waveform for the audio signal must be determined by testing.

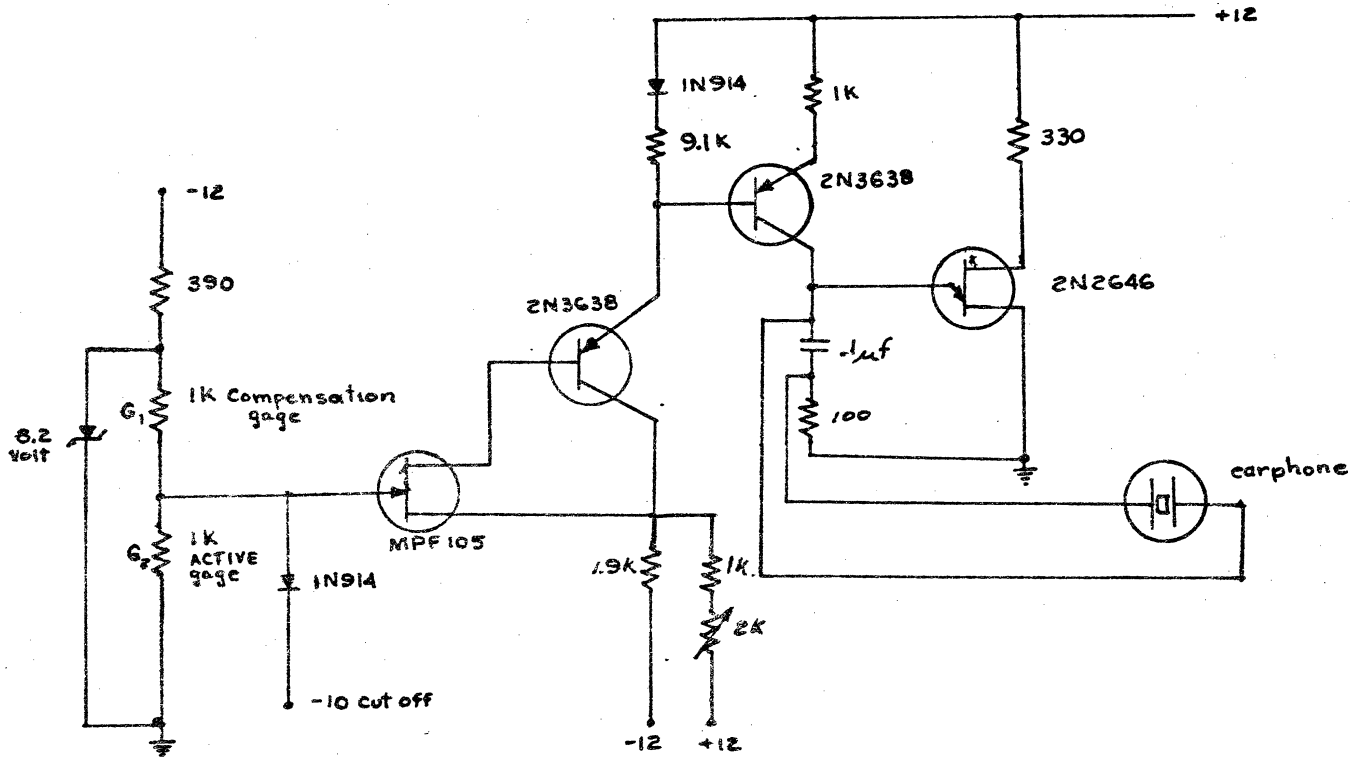
B. Final Design

The circuit involves a regulated voltage bridge comprised of two silicon strain gauges. One gauge is active; the second is for temperature compensation. Both are epoxied to the stainless steel finger brace of the hand splint.

The center of the bridge controls the gate of a field effect transistor (FET). A -10 volt signal from the control system at the gate will turn off the oscillator. The amplified drain-source current of the FET causes a voltage drop across R_3, which controls the current to a unijunction relaxation oscillator. The oscillator operates from 1 hertz to 4 kilohertz.

R_4 is used to bias T_1, and thereby determine the lowest frequency. R_3 controls the sensitivity of the circuit; an increase in R_3 increases the sensitivity. The output waveform is a sawtooth.

Feedback Oscillator



SELECTION OF A METHOD FOR EMG SIGNAL DETECTION

Three possibilities for detecting the level of electrical activity in the control muscle were considered: the implanted electrode, the implanted transducer, and the skin surface electrode.

The Implanted Electrode

This method involves the surgical implantation of an electrode into the control muscle. The electrode generally consists of a very thin platinum, gold, or stainless steel wire. One end is embedded in the muscle while the other is available at the skin surface for connection to sensing apparatus. The advantages of such a method are:

1) Extremely low impedence values are obtained. With the electrode implanted in the muscle, there is very little insulating tissue between electrode and muscle fibers.

2) There is a very low level of artifact signals. The principle source of these extraneous signals is movement of an electrode relative to the tissue. After healing of the original wound, this movement is negligible.

3) There is little noise from muscles near the control muscle. The implanted electrode detects a signal from only the specified muscle.

Disadvantages of this method are:

1) The entry site of the electrode into the skin is essentially an open would even after healing. There is a strong possibility of irritation to the skin and infection.

2) Experimenters have reported a problem of lead and electrode breakage due to work hardening. This results in a need for frequent replacement by surgery. A new method of construction of these electrodes

has been introduced by Case Western Reserve University. Their Transcutaneous Electrode consists of a wire wound into a coil, which is then filled with Silastic. This construction significantly decreases the possibility of fracture, and the electrode should prove extremely useful in short-term experiments. Use in an orthotic device designed for many years of dependable operation would not be possible since replacement would be required at least twice a year.

3) The requirement for connections to sensing equipment restricts the patient in freedom of movement.

4) Most patients would, for solely psychological reasons, object to the idea of a semi-permanently implanted electrode protruding from the surface of the skin.

The disadvantages of using an implanted electrode far outweigh the advantages. It is not probable that his method will be improved upon to the point that it can be employed in the future.

The Implanted Transducer

This procedure consists of placing an electronic transmitter in or on the control muscle. Electrodes inserted into the muscle detect the signal and pass it to the telemetering unit, which transmits a signal to an externally located receiver. The signal received is proportional to the electrical activity of the muscle. Devices of this type are being developed at Case Western Reserve University. The advantages are:

1) Extremely low impedence values due to proximity of the muscle to the sensing electrodes.

2) Prevention of device movement within the tissue and subsequent absence of artifact signals.

3) There are no open wounds and little danger of infection.

4) There are no physical connections to external sensing and processing equipment, which allows the patient greater freedom of movement.

5) A signal from only the specified control muscle is received; there is no noise from adjacent muscles such as the antagonist muscle of the control muscle.

It is possible that such devices will eventually be used in such long-term applications as in orthotic devices. The advantages are substantial. The microcircuitry industry is advancing rapidly, and the size of the telemetering unit will undoubtedly be no problem. Significant advances have been made in utilizing biological sources of power and such a source could possibly be used to operate a transducer. Thus, a reliable, long-term unit is foreseeable... Work must be done in the following areas:

1) Electrode materials must be found that will not effect, or be affected by body tissues and fluids.

2) The problem posed by the body's tendency to reject foreigh objects must be solved.

3) The problem of material fractures due to stress and hardening must be solved.

Although the prospects for the transducer are promising, the device has not been developed to the point that it can be used in current orthotic devices.

Skin Surface Electrodes

This method involves an electrode applied at the surface of the skin and leads for attachment to sensing equipment. The EMG signal passes through the tissue between the electrode and muscle and is detected by the potential set up at the skin surface. The electrode generally consists of a stainless steel or silver coated disk or a silver chloride pellet. A compound is usually applied to the skin at the site of the electrode; it dissolves the oils exuded by the skin and thereby reduces the impedence between the skin and the electrode. The compound conducts the signal from the skin to the

surface of the electrode.

For both the surface and implanted electrodes, a three electrode scheme must be employed to avoid electrical interference detected on the body surface and in the leads. Two of the electrodes are active and placed approximately four inches apart (depending on the size and shape of the control muscle), while the third is grounded and situated between the active electrodes.

It can be expected that the magnitude of the detected EMG signal will increase with an increase in surface area of the electrode. Unfortunately, effects from adjacent muscles also increase, so that a compromise must be reached. Small electrodes introduce less noise and larger ones provide a stronger signal.

Advantages of the surface electrode are as follows:

1) There is no need for surgical implantation. The devise can be easily attached to the skin with the proper adhesive.

2) There is no danger of infection or body rejection.

Disadvantages include:

1) Possible irritation to the skin by the signal conducting compound.

2) Movement of the device relative to the skin, producing artifact signals. Although methods of electrode attachment have been developed, movement is more troublesome than with implanted devices.

3) Relative distance from the control muscle. This has two bad effects. First, the tissue between the electrode and the control muscle is of considerable thickness and the impedence between the two is increased. Second, the noise level is increased since the surface electrode can detect signals from muscles other than the control muscle.

4) Connections to sensing devices restricts the patient from full freedom of movement.

A decision was made in favor of using the surface electrodes. Although this type of device is not problem-free, it presents the simplest method for signal detection and is therefore most appropriate for initial developmental work. The signal to noise ratio is sufficient to attain delicate control since most interference introduced by the electrodes is masked out in the signal processing units. The most attractive attribute is that surgical implantation is not required.

Selection Criteria for a Surface Electrode

The following considerations must be made:

1) The electrode must be lightweight.

2) The cosmetic value must be acceptable to the patient.

3) The compound used to conduct the signal to the electrode must not cause irritation even if the electrodes are worn approximately eight hours per day, every day.

4) The adhesive used to attach the electrode must not irritate the skin. It should be able to hold the electrode firmly enough that artifact signals are not introduced; it should hold the electrode for at least ten hours even if the patient is perspiring. The method of attachment should be quick and simple.

5) The electrodes must have the smallest possible values for the following characteristics:

a) Offset voltage: electrodes face to face with compound in between.

b) Electrode-to-skin offset voltage: depends on how the skin is preconditioned.

c) Polarization voltage: electrodes face to face with compound in between; with 0.1 microamp. flowing across electrode interface.

d) Electrode impedence: electrodes face to face with compound in between.

e) Electrode-to-skin impedence: depending on how the skin is preconditioned.

Characteristics of the Selected Electrode

A survey of available surface electrodes has shown that the Beckman Bioptential Skin Electrode (Spinco Division of Beckman Instruments, Inc., Palo Alto, Calif.) most closely meets the selection criteria. A disposable adhesive colar is used to apply the electrode to the subject. The adhesive collar is a transparent plastic disc with an adhesive coating on both sides and a hole in the middle. The adhesive coating on one side bonds the electrode to the collar. The adhesive coating on the other side bonds the collar to the skin of the subject. There is an electrolyte reservoir space between the back of the electrode face and a silver/silver chloride pellet inside the electrode. The hold in the middle of the adhesive collar exposes the skin of the subject to four reservoir access holes in the face of the electrode. When electrolyte gel has been squeezed into the reservoir through the access holes, it forms an interface between the skin and the pellet. When the electrode and collar are pressed into place, the collar hermetically seals the electrolyte into the reservoir, retarding evaporation and preventing contamination of the electrolyte. The electrolyte provides an ion flow from the skin to the pellet.

The specifications are as follows:

Weight	2.3 grams
Diameter	19 mm
Thickness	5 mm
Offset voltage	less than 350 microvolts
Electrode-to-skin offset voltage	as low as 500 microvolts
Polarization voltage	less than 12 microvolts
Electrode impedence	less than 100 ohms
Electrode-to-skin impedence	as low as 300 ohms

The electrodes are available in flesh colors and the adhesive collars allow virtually no movement of the electrode relative to the skin. There

is, however, a problem concerning the electrolyte compound. Experiments have shown that wearing the electrode with the compound for ten hours per day, for one week, produced significant irritation to the skin at the point of application of the gel and moderate irritation at the point of contact of the collar.

SELECTION OF BATTERY FOR +12 V. SUPPLY

Power requirements of the electronic circuitry for signal sensing, control, and feedback necessitate a -12 V. to +12 V. power source. Half of this power, the -12 V. to 0 V. portion, is available from the lead-acid storage batteries used to power the patient's electric wheelchair. The 0 to +12 V. portion is to be obtained from a separate power source.

A survey was made of available power sources, taking into consideration energy density, practicality and availability, and the field rapidly was narrowed to rechargeable batteries. Latest literature (Annual Power Sources Conference Proceedings) indicated that three types now account for most usage: nickel-cadmium, cadmium-silver, and silver-zinc. Some comparative characteristics of these were given, as follows:

	Ni-CD	CD-AgO	Zn-AgO
Open-cir. Voltage	1.3	1.4	1.8
Discharge Voltage (5 hr. rate, 80°F)	1.2	1.1	1.5
Watt-Hr/Lb (5 hr. rate, 80°F)	10-15	20-35	40-55
Watt-Hr/In^3 (5 hr. rate, 80°F)	0.7-1.1	1.2-2.7	2.0-3.7
Charge Retention (50% cap., 80°F)	1 yr.	>2 yrs.	>2 yrs.
No. of Recharge Cycles in Lifetime	>2000	300-500	100-250

Examination of cataloges of industrial electronic supply houses revealed that only Ni-CD batteries are stocked on a regular basis, which would require that CD-AgO and Zn-AgO types would have to be purchased directly from the manufacturers, possible involving some problems in obtaining these types. It was this availability factor which lead to selection of a Ni-Cd battery (G.E. #10GB100, 12.5 V), although the longer cycle life of the Ni-CD was also of

value. Since the particular patient who will use this device is confined to a wheelchair, the higher energy density of the Cd-AgO and Zn-AgO batteries was not a deciding factor.

Appendix B

METHODS FOR DISPLAYING THE PRESSURE FEEDBACK SIGNAL

Several possibilities by which the patient could sense the signal of the pressure feedback system were proposed. The patient's sense of temperature could be employed by placing an electrically heated element somewhere on the body, so that the patient would sense a slight increase in warmth as he caused an increase in grip pressure. This method was rejected due to a lack of precision and possible irritation to the skin. Mechanical sensing employing a vibrating device to blow air on the skin is a possibility. Studies in human factors suggest employment of a mechanical display of the signal since a mechanical quantity, the grip pressure, is being measured; physiologically, it is the most natural method. The proposal was rejected, however, due to a lack of sensitivity and cosmetic value.

Visual sensing was proposed, in which the patient would observe the change in brightness of a light. As the grip pressure is increased, the light would become brighter. The proposal was rejected because the light would not be acceptable to the patient. It reduces the cosmetic value of the system and, if the patient had to watch a light he would not be able to carry on activities concurrent with operating the splint.

The audio sense was determined to be the best suited for the purpose. The method employs an electromechanical sound transducer placed near the ear. The changing grip pressure changes, by use of strain gauges on the splint, a particular audio variable. As will be shown, the method presents acceptable cosmetic value, provides excellent capability for distinguishing various grip pressures, provides a minimum of distraction from concurrent activities, and employs a code to which the patient can

become accustomed with reasonable practice.

Selection of the Audio Variable

The three possible variables are:

1) intensity: as the grip pressure increases, the loudness increases.
2) frequency of occurrence of monotonous clicks: as the grip pressure increases, the frequency increases.
3) frequency of a periodic signal: as the grip pressure increases, the frequency increases.

Of the three choices, sound intensity is the least attractive because the human ear's capability to discriminate levels of intensity is very poor. The range of intensity employed would have to be extended beyond a value acceptable to the patient in order to get a sufficient number of distinguishable levels.

Variation of the frequency of monotonous clicks is a great improvement over intensity. The intensity level and frequency of the pulses can be set at values acceptable to the patient. Studies have shown that the patient could be able to distinguish changes in frequency of occurrence of the pulses to within three-four pulses per second. This variable would have to be used by a tone deaf patient.

The change in frequency of a periodic waveform was selected to be used in the prototype. It is not yet clear that this variable is more advantageous than the variation in frequency of clicks. For instance, the patient can detect changes in frequency of one-two cycles per second, which is not a significant improvement over the three-four pulses per second that is detectable with monotonous pulses. Studies must be run to determine which variable is preferred by the patient and whether one

variable is more useful than another. The frequency of a periodic waveform was used because the circuitry required to employ this variable was simpler than that required for the alternative. A sawtooth waveform was chosen because it too required the simplest circuitry. Studies must be carried out to determine the waveform that is preferred by most patients.

Human Factors

It is desired that the patient be able to use the pressure feedback system in two ways. First, he would like to use it to sense changes in the pressure he is applying; he would like to compare various pressures by comparing frequencies. Second, the patient would like to remember the frequency that is associated with a particular grip pressure. For instance, he would like to remember the frequency required for holding a drinking glass. In summary, the patient requires both absolute and comparative judgement of frequency.

The requirement of comparative judgement presents no problem. The oscillator used provides a side frequency range, and the patient can detect changes of one-two cycles per second. The patient will easily be able to select a pressure value and vary the pressure with respect to that value.

The patient will not have equally good absolute judgement. Studies in human factors have shown that subjects can identify no more than nine frequencies selected from a wide frequency range. As the experimenter increases the number of frequencies to be identified from one to six or seven, the subject is able to handle the information. After seven, the subject begins to make errors. This limit on capacity for absolute judgements due in part to memory capabilities since the main difference

between absolute and comparative judgement is the presence of a reference frequency in the latter, and its absence in the former. Little success has been obtained in improving absolute judgement by practice.

These factors imply that the patient will be able to remember no more than seven to eight frequencies after reasonable practice. The fact that the feedback system presents a continuous signal and that the cited human factors studies employ distinct frequencies presented one at a time may mean that the studies do not actually apply. No work has, as yet, been done to determine the limit of absolute judgement when a continuous signal of varying frequency is presented to a subject.

Appendix C

ECONOMIC FEASABILITY

Meeting the criteria for economic feasability (see Design Objectives) was considered to be of major importance in all design considerations.

Range of Patients: The range of patients who would have use for the device extends from those who have incurred only one paralyzed hand to those who are totally paralyzed below the neck. If a patient does not have use of a hand but still has control over virtually any other muscle in his body, the system could be employed. A patient with only a paralyzed hand could again pick up a pencil and write. A patient with extreme paralysis throughout his body could operate the control lever of an electronically controlled wheel chair.

The University Hospital of The University of Michigan School of Medicine receives approximately twenty-five patients per year that fall in the above category. No surveys have, as yet, been run to obtain a number for the entire country. It is expected, however, that there are enough patients per year to make manufacture of a large number of devices practical, and thereby reduce the cost per unit.

Associated with the objective of securing enough patients to make manufacture practical is the fact that the entire system could be adapted for use as a prosthetic device. The system could, in the same way that it operates an orthotic splint, be used to open and close a prosthtic hand. This factor significantly increases the number of potential purchasers.

Manufacturing Costs: It is not possible to present an exact cost for producing the device since, to the date of this report, only a prototype has been built. If ten units were built, the approximate costs, per unit, would be as shown below.

1) Sensing	
EMG amplifier	$ 45.00
integrator	10.00
2) Control circuits	50.00
3) Actuator	
motor	90.00
brake	30.00
ball screw	55.00
miscellaneous	125.00
4) Feedback circuits	15.00
5) Electrodes	30.00
6) Battery	30.00
7) Battery charger	10.00
8) Orthotic Hand Splint	200.00
Total Cost	$690.00

No studies have been made concerning the average income of patients who would have use for the device. Many patients incur paralyzed hands due to automobile accidents, industrial accidents, and war. It is expected that such cases would be covered, at least in part, by accident insurance. Since the unit is designed to provide many years of reliable operation, a cost of $600 to $700 should be manageable.

Cost of Operating the Device: The power supply battery will provide reliable service for approximately 15 years. The actuator will require overhauling after about 3 years of operation at a cost of approximately $75 for a new brake and motor brushes.

Adaptation for a Particular Patient: The patient requires adaptation only to the extent that various muscles used as control muscles will present different degrees of electrical activity, that is, different maximum levels for the EMG signal. The effects of these variations can be suppressed by changing the values of certain components in the sensing and control circuitry, and the cost of adaptation to the patient would be negligible.

Appendix D

SPLINT

The hand splint was provided by the Department of Physical Medicine at the University of Michigan Hospital. It is a modified version of the splint made at the Rancho Los Amigos Hospital in Downey, California.

The flexor hinge splint is designed to provide a functional "three-jaw chuck" type prehension for the patient who has lost the normal ability to grasp with his fingers. In this type of prehension, the index and long fingers contact the thumb. It is finger driven; the hinged finger brace harnesses the fingers together, and the thumb brace holds the thumb in a position of opposition to the fingers.

The splint is made of stainless steel, with a skin colored plastic coating on all surfaces except the finger brace. The outer cable of the driving linkage is firmly fastened to the splint near the base of the thumb.