END POINT CONTROL USING HEAD ORIENTATION

A.G. Potter and J.H. Friedman

Summary

The orthotic brace control system described in this paper is designed to enable a quadriplegic patient to operate an externally powered, upper extremity brace using head orientation information. This type of control scheme, sometimes reffered to as end point control, involves measuring head orientation coordinates, computing the desired coordinates of the patient's hand in space, and driving the orthotic brace to achieve this position. At present, two gimbals interconected by a small rod are used to obtain head orientation coordinates with one gimbal strapped to the back of the patient's head and the other mounted on the wheelchair. This transducer is designed so that restraint of head movements is minimal. The equations that express the relationship between the coordinates derived from head orientation measurements and the actual brace angles are relatively complicated. To keep cost down, approximate linear control equations are being used.

The Problem

In recent years an increasing number of people are surviving accidents or neuromuscular disease with extensive paralysis. Typically such disability occurs in persons who have suffered poliomyelitis, muscular distrophy, celebral palsy, or lesions in the fourth or fifth cervical spinal cord region. Many of the spinal injuries are the result of automobile and driving accidents which unfortunately do not appear to be on the decline. Quadriplegics, those experiencing paralysis of all four limbs, are normally bedridden, but may spend some portion of their day wheelchair bound. Generally C-4 patients, while lacking function in their upper-extremities do retain normal muscular control from the shoulder up including, in some cases, the ability to raise and lower the shoulder girdle. For this type of patient the problem of restoring limited function by means of external mechanical devices, termed upper-extremity orthotics, is difficult because such mechanisms must be built to follow the anatomical joints and support the flail extremity while at the same time generating nearly normal upper-extremity motion. The control of such an orthotic device is particulary difficult for severely handicapped patients requiring multi-degree of freedom assistive braces and possessing few functional residuals for control signal sources. This paper presents a simple engineering approach to this important problem in the medical rehabilitation field.

The Objective

The rehabilitation of upper-extremity function through orthotic devices is doubly challenging for any solution must be both technologically sound and psychologically acceptable to the patient. One approach to establishing the design criteria for an orthotic control system is to investigate the natural system's function and control mechanisms. Simply stated, the overall purpose of upper-extremity control is to provide a limb with a controllable movement and an ability to exert useful energies against external forces. The design of any assistive device for patients lacking normal arm function must include consideration of the mechanical constraints among different parts of the body, the position and velocity characteristics of the limbs, the influence of inertia of the limbs, and the influence of external forces.

Realistically, one must accept the fact that the mechanical device will never satisfactorily substitute for a normally functioning limb. However, the objective of rehabilitation of a quadriplegic patient is not to enable him to perform tasks as efficently as can be done by an attendant, but is to provide some degree of functional independence and associated personal satisfaction. It is of psychological advantage to allow the patient contlnous voluntary control over the system rather than merely initiating a fully automated motion sequence even though its performance must be superior. From a purely mechanical standpoint, it would be much easier to design a manipulator which would execute a programmed routine, but it is generally agreed that mobilizing an existing arm and actively involving the quadriplegic in the control system are beneficial in minimizing the feeling of being a "mechanical man" and encouraging any possible increase in residual limb function.

The goal of the research described in this paper is the development of an upper-extremity orthotic system which will partiatlly compensate for the functional losses inherent in upper-extremity paralysis. Included in the specifications of the device are an ability of the patient to perform grooming, self-feeding, and light vocational tasks. The design is based on providing the C-4 quadriplegic with an orthotic brace that can be positively positioned in a natural manner and that requires minimal patient training, concentration, and effort to operate. The particular control scheme developed is chosen for its cosmetic acceptability, direct patient involvement, and ease of operation.

State of the Art

During the past decade researchers have developed numerous upper-extremity orthosis to provide partial return of an arm function to severaly paralysed patients. Although designers have shown awareness of control and feedback, their primary attentions have been directed toward the powering and fitting of assistive devices. Investigations have been conducted in many areas including studies regarding brace configuration, actuator types, modes of control, and suitability of various control sites. Present state of the art is such that the necessary hardware can be built, but there are serious problems involved in designing effective control systems.

Arm function is extremely complex; in fact, there are eleven degrees of freedom in the arm not including the hand. The trend in the development of orthotic brace configurations has been to increse the number of degrees of freedom in hope of providing a more flexible and functional brace. One such device, the Rancho Electric Arm, has seven degrees of freedom which is thought by some investigators to be the minimum number required to restore reasonable arm movement. The seven degrees of freedom include two at the shoulder, two at the elbow (one flexion/extension and the other humeral rotation), forearm rotation, wrist flexion/extension, and hand prehension. However, generally associated with an increase in the number of degrees of freedom is an undesireable increase in the bulk of the brace and complication of the control system.

Studies in the past have concentrated on two basic approaches: First, to operate an orthotic brace completely by direct patient control, and second, to make brace control fully automatic. In direct control scheme, the patient exercises continous control over the motion of the assistive device. Automatic control, once initiated allows a movement to progress to its completion without further conscious attention. There are obvious problems with both methods.

Direct patient control is difficult because of the number of degrees of freedom which must be controlled. Devices in this category presently require separate sites or switches to control each joint of the brace. The disadvantages of this type of control are that coordinated motion of the brace is difficult since multiple sites must be activated simultaneously in order to achieve smooth positioning of the brace. In general, presently developed systems of this type require a degree of mental effort and attention that is excessive, particularly in terms of the frequently unnatural motion that results.

On the other side of the spectrum is the completely automatic device in which the patient simply selects which one of several programmed motions will be performed. The problems associated with this approach include reduced adaptibility due to a limited number of movement sequences, the expense of peripheral equipment normally associated with such a system, and substantially reduced patients participation. Presently, several studies are being conducted which combine automatic and direct control by allowing the patient to interrupt an automated motion at any time and assume direct control.

In recent years, one of the most active areas of interest has been in discovering anatomical sites which are suitable for generating control signals. In general, C-4 level quadriplegics have only the following control sites available: relative motions of body parts above the shoulders, electromyographic signals (EMG), electroencephalographic signals (EEG), and sound of speech.

Investigators have considered several schemes for transforming relative motion of parts of the body into usable control signals. Including among these studies have been systems using head or chin motion to actuate switches, a light source attached to eyeglasses that may be directed to activate approptiate photocells, and switches activated by eyebrow motion. The use of the tongue to operate a switch or strain gage array has also been used. One of the latest approaches has been an attempt to use eyeball motion as the signal source. This method utilizes the fact that light shining on the eye is reflected back toward the source in varying amounts depending on eye orientation. Electromyographic signals, the electrical activity associated with muscle activity, have been used in various control schemes. The main problem with such systems has been the excessive amount of mental efforts required to activate multiple sites in a coordinated fashion. Electroencephalographic signals, the potential activity of the central nervous system, have been proposed as a signal sources, but present techniques of signal pattern recognition are not sufficently advanced to make a practical system. The use of sound or speech to activate electrical circuits by using acoustical filters appears feasible, but again the control of several actuators by this method would be difficult and its operation would limit communication during brace operation.

Control System Design

Based on the limitations of existing upper-extremity orthotic systems, the design of an improved assistive device requires the selection of a more suitable control site. The study described in this paper is an attempt to develop a more functional control system through the use of a set of transducers which allow the patient to directly specify the desired position or end point of the brace. Such a system has the capability of greately reducing the requirement for signal processinq and simplifying the control task to one that is similar to the conventional rendezvous problems in guidance engineering.

The design criteria is based on providing a control system which allows a patient to smoothly coordinate and control an orthotic brace without having to separately actuate each joint. In this way the effort, concentration and training required by the patient are minimized while attempting to maintain reasonable functional utility. Throughout this study several compromises were made in order to limit the cost of the final product.

Control Site

The desire to initiate movement of an orthotic device originates at some conscious level in the central nervous system and takes the form of some voluntary physical action. The simplest of such actions and the most easely detected are those associated with musculoskeletal movement or gross body motion. For patients suffering extensive paralysis and capable of spending a portion of their day in a wheelchair, the logical choice for a control site is one which involves movement of a functional body part relative to the wheelchair. For this reason, head orientation was considered as a possible signal source.

Head orientation is particularly suited as a control site, since the head has its own vertical sensing element as smooth control over a wide dynamic range is possible. However, the main advantage of using head orientation is the fact that it is a natural site of independent motion in azimuth and elevation and as such can possibly simplify the control test. Although eye orientation is also a natural site for genearting azimuth and elevation signals it was considered to be less desireable that using head orientation. Head motion offers the capability to specify an additional coordinate, the desired range or radius of action, and permits the complete use of visual feedback in performing tasks requiring precision.

Transducer Design

In order to provide a continous measurements of the angular orientation of the head together with a simulated range signal based on head position a transducer consisting of two interconnected gimbals was constructed. This device, shown in Figure 1, includes two plexiglass gimbals with potentiometers on each axis connected by a small aluminium rod. One gimbal is strapped to the patient's head and the second gimbal is attached to a mounting on the back of the wheelchair. The device allows the patient to rotate his head approximately 100 degrees in the vertical plane and 80 degrees in the horizontal plane. The only significant restriction is in the forward-backward motion of the head which is limited to a control range of approximately two inches. This could be increased by making the control range a small portion of a larger total motion if desired.

It is necessary to use two gimbals to measure the actual angles of head azimuth and elevation as referenced to the wheelchair. As shown in Figure 2, the true elevation angle is obtained by adding the corresponding angles of both gimbals. This same relationship holds true for measuring azimuth. Thus the gimbal mounted on the wheelchair measures the necessary correction to account for the fact that the gimbal strapped to the patient measures angles with respect to the interconnecting shaft rather than to a fixed set of axes.

Radius of action or desired range is simulated by the patient's relative forward-backward position of his head which, as shown in

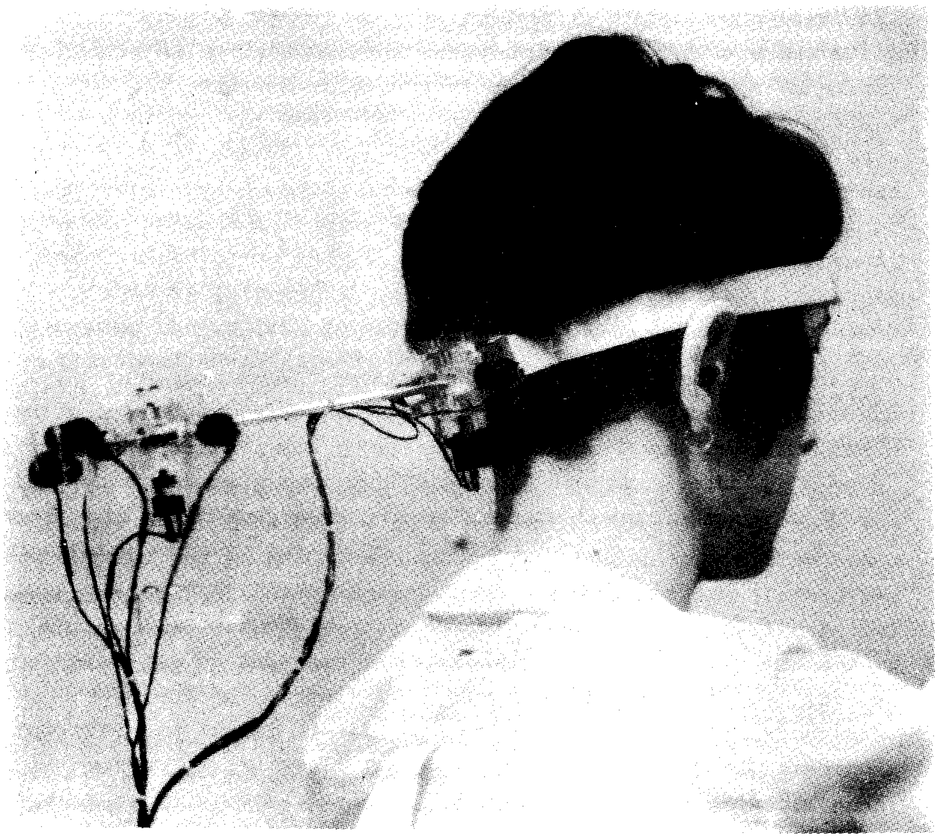

Fig. 1. Head orientation transducer

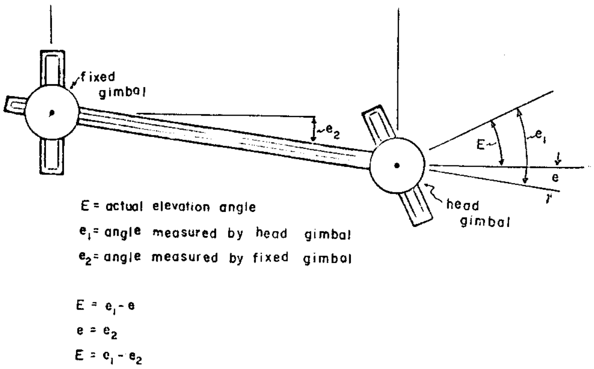

Fig. 2. Measurement of elevation angle

Figure 1, is converted from **linear** displacement of the interconnecting shaft to the **rotation** of a potentiometer. Minimum range is selected by the patient moving his head to the most forward position, a somewhat natural eating posture. Range is increased by the patient moving his head backward.

Head orientation is well suited as a control site **because of** ease in measuring a set of coordinates which fully specify the desired end-point of an orthotic brace. It is also a natural signal source which requires minimal concentration, effort, and training to activate.

Orthotic Brace

The objective of this study, most simply stated, is to maneuver a flail arm to a position where a hand splint can perform some useful function. Since the patients who can benefit from this system are wheelchair bound, the wheelchair is utilized as a convenient support for the necessary transducers, electronics, power source, and orthitic brace. The brace shown in Figure 3, was originally developed by Engen as a pneumatically actuated feeder, but has been converted to electric motor drive. Electrical actuators were selected primarily **because** of the readily available source of battery power in electric wheelchairs and the ease of control circuit design.

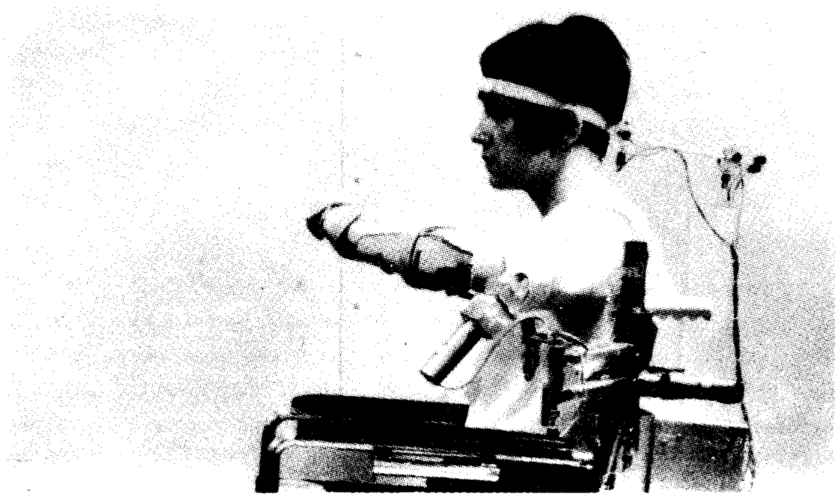

Fig. 3. Orthotic brace

This orthotic brace allows for three powered motions; a horizontal displacement, a vertical displacement, and an elbow flexion/extension. The horizontal and vertical displacements are completely independent motions and together contribute to the abduction/adduction and flexion/extension at the shoulder. An extension spring is used to lessen the effects of gravity by assisting in the vertical support of the brace and the arm. A telescopic rod and tube connected to the elbow flexion/extension unit serves as an attachement for the hand support. A molded elbow and forearm trough is attached to this unit and acts as a support for the forearm which is held in the trough by a velcro strap. There are several other sites for adjustment of the orthosis to assist in the fitting of patients.

The exact power requirements of upper-extremity orthotics are difficult to define, not only because of the wide age span of patients but also because of variations in loading and limb weight. To allow for a wide range in torque requirements permanent magnet, DC motors were used with adjustable gain drive circuitry. The 24 volt DC motors used have planetary gear heads with a 639:1 gear reduction and provide the capability of 288 oz. in. torque under continous load conditions. This type of motors is particularly desirable because of its relative compactness and light weight. Although there is some noise associated with their operation, it is not distracting and may provide some useful function as an audible feedback.

All joints of the orthotic brace, three driven and one free, are continously monitored by potentiometers. These single turn potentiometers provide measurement of the brace angles and are mounted with couplings which allow easy adjustment for proper reference.

Control System

The control system was designed to be volitional, proportional, and vectorial. Volitional means the patient can start, stop, or modify the course of action. Proportional control means that by varying his motion the patient can control the rate of action or the force exerted. Finally, vectorial control means that a particular motion can be achieved in a smooth direct fashion rather than in a sequence of motions about different axes.

The overall scheme behind this design involves the theory of end-point control in which the parameter to be specified is position and the control signal is in terms of a desired end-point. To fulfill the requirements of end-point control it is necessary to solve control equations which express the relationship between the head oriented coordinated and those of the orthotic brace.

Figure 4 is a diagrammatic representation of the brace along with the head and brace coordinate systems. The brace angles include: F, a motor driven brace angle producing humeral flexion/extension; G, a motor driven brace angle causing humeral abduction; and J, a motor driven angle giving elbow flexion/extension. The axis of rotation of angle J is offset 40 degrees from the vertical thereby giving this motion both a vertical and horizontal component. The angle H is a free brace angle.

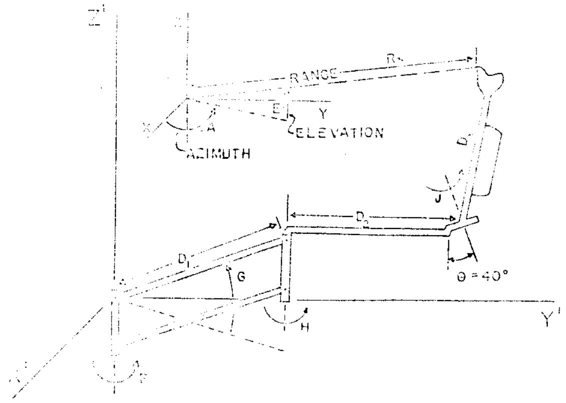

Fig. 4. Brace coordinate systems

The equations expessing the realtionship between the head oriented coordinate system and the orthotic coordinate system in terms of a desired end-point (x, y, z) are as follows:

$$x: \quad R \cos E \cos A = d_1 \cos G \cos F + d_2 \cos (F + H) + [d_3 \cos \theta + (d_3 - d_3 \cos \theta) \sin J] \cos (F + H + J) + k_x$$

$$y: \quad R \cos E \sin A = d_1 \cos G \sin F + d_2 \sin (F + H)$$

$$+ [d_3 \cos \Theta + (d_3 - d_3 \cos \Theta) \sin J] \sin (F + H + J)$$

$$+ k_y$$

$$z: \quad R \sin E = d_1 \sin G + d_3 \sin \Theta \cos J + k_z$$

where d_1 = 7.5 in., d_2 = 7.5 in., d_3 = 18 in. and θ = 40 degrees.

The constants k_x, k_y, and k_z account for the respective x, y, and z displacements between the two sets of axes and are dependent upon the adjustment of the brace in fitting it to a particular patient.

These exact equations are relatively complex trigonometric expressions and their solution requires considerable computational equipment or a costly series of resolver chains in place of the potentiometers. At the sacrifice of some accuracy, but with considerable cost reduction the simplified set of control equations given below are being used.

$$A \cong K_1F + K_2H$$

$$E \cong K_3G$$

$$R \cong K_4F + K_5H + K_6G - K_7J$$

Where all K terms are experimentally determined weighting factors. These equations are based on the geometry of the brace along with some consideration for the natural motion that is being simualted.

The horizontal and vertical displacements of this assistive device are completely independent motions, but actuation of the elbow flexion/extension unit results in both horizontal and vertical components. A straightforward way of relating the head generated signals of azimuth, elevation, and range to the motorized angles of the brace is to assume that the elbow flexion/extension actuator is primarily involved in changing the range variable.

This assumption is reasonably valid in that the actuators controlling horizontal and vertical displacements by themselves have minor effects in changing the radius of action of the hand. Based on these approximations, the actuator for horizontal displacement, angle F, and the actuator for vertical displacement, angle G, are coupled to the range signal to compensate for the fact that elbow flexion/extension has components besides range associated with its motion.

A basic assumption in this research is that the test control technique would be one where the hand is driven to a point in space which lies on a ray defined by the intersection of the mid-sagittal plane and mouth level transverse plane both of which are referenced to the head. The range from mouth to hand on this ray is determined by the distance from the back of the head to a reference point behind the patient. This design depends to some extent on visual feedback for correcting errors in the positioning of the brace which are a result of simplifying the control equations and in the fine positioning required for performing precision tasks.

Figure 5 is a block diagram of the overall control system for one motorized component. Azimuth, elevation, and range and appropriate brace angles are weighted together by a resistive network and the result, a computed angle, is compared to the actual brace angle. This is accomplished by the differential amplifier arrangement shown in Figure 6. The two outputs of this circuitry are directly related to the magnitude of the error existing between the computed and the actual brace angle. The next stage of the electronics consists of a pair of pulse width modulator circuits, shown in Figure 7. If an output from the comparator stage exceeds a prescribed level, which is adjustable, the pulse width modulator circuits will generate a signal with a duty cycle which is a function of the error. Both the dead zone and gain of this circuit are adjustable, thereby allowing the control of sensitivity, motor speed, and dead zone to be matched to the limitations and requirements for a particular patient. The motor drive circuits, shown in Figure 8, control the direction and speed of the motor so that the error angle is reduced to within a given range. Each of these circuits is mounted on an individual printed circuit board and inserted in a rack fastened to the back of the wheelchair. Two 12 volt batteries are used as power source for both the motors and electronics.

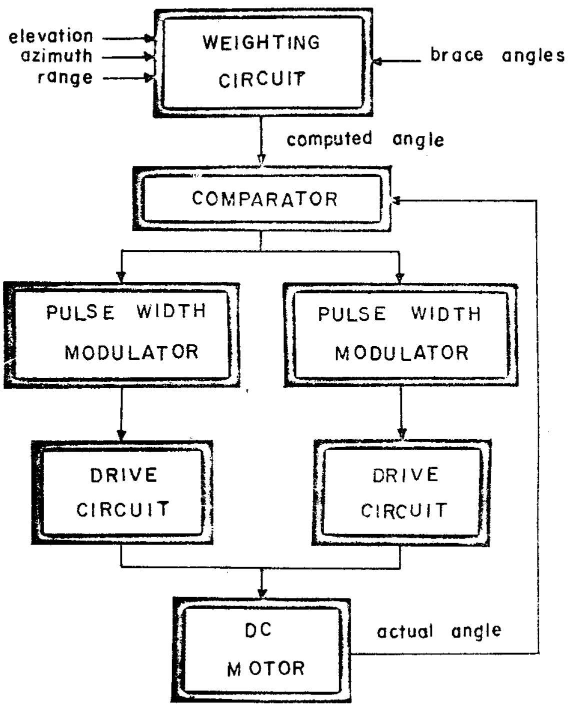

Fig. 5. Diagram of control system

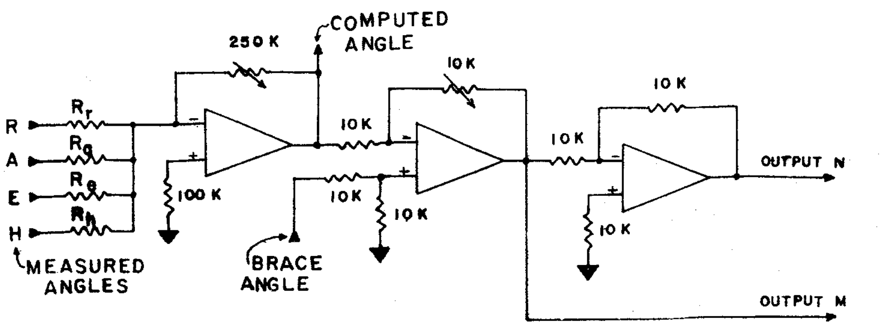

Fig. 6. Weighting circuit and difference amplifier

Conclusion

This system provides the severely paralyzed patient with a simple, low cost, arm brace which can be operated with minimal effort, concentration and training. The key feature in this design

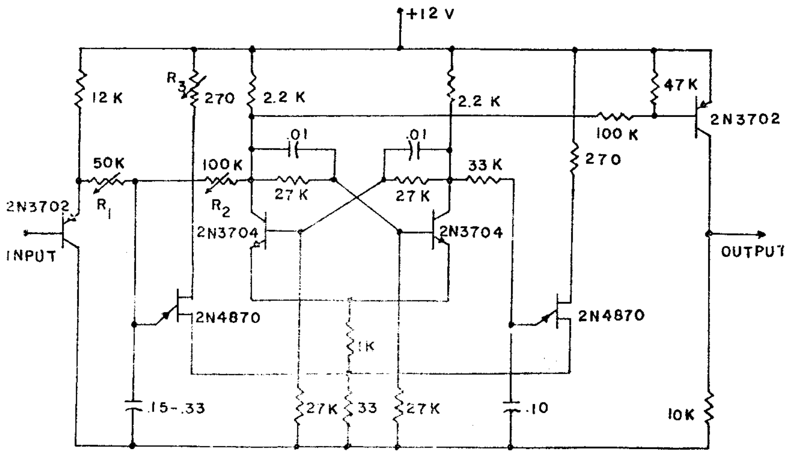

Fig. 7. Pulse width modulator circuit

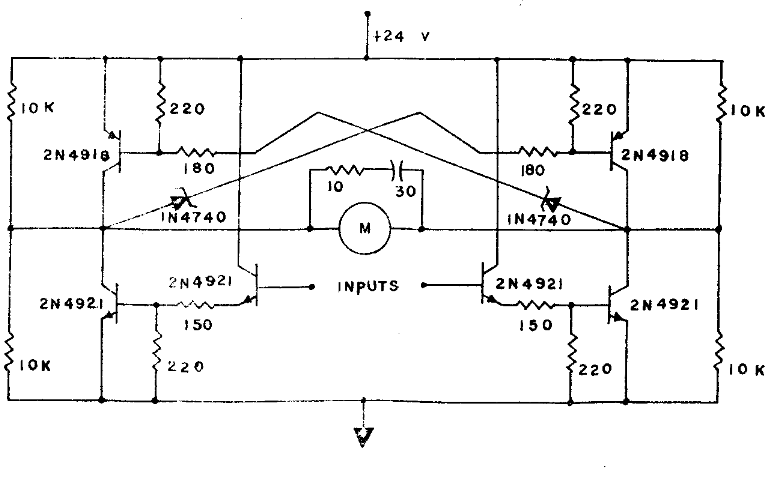

Fig. 8. Motor drive circuit

is the use of head orientation as the controlling signal. This control site provides independent motion in azimuth and elevation, is cosmetically acceptable, and allows the patient to exercise direct control over the orthotic brace. To simplify the control problem the end-point coordinates are automatically computed from head orientation so that the brace will move the hand to this point without requiring any further mental effort.

Evaluation of this design was performed by filming a normal subject executing a serious of prescribed motions. The subject was instructed to perform a sequence of head motions which generated independent signals in azimuth, elevation, and range. During this experiment, films were taken to evaluate the accuracy and controllability associated with head orientation. Throughout the study, the subject was able to maintain positive control of the brace with little effort and the resulting motions were smooth and reasonably natural. The subject was then given the task of performing a set of coordinated maneuvers which involved taking a glass from the experimenter, lifting the glass to the mouth, and returning the glass to the experimenter. The first two attempts to perform this task required considerable concentration, but on succeeding trials the subject completed the maneuver with relative ease.

Future Research

Although a control system based on head orientation appears to be a highly desireable approach to the problem of restoring limited function to severely paralyzed patients there are still many improvements which can be made. The following summary lists various problems presently being investigated.

1. A new way of sensing head orientation using ultrasound so that no physical connection between the head the wheelchair will be needed.
2. Incorporation of hand prehension control and wheelchair drive control into the basic end-point control system that has been developed.

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