Application of External Power in Orthotics*

by

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The application of external power in orthotics has been explored for a number of years and has, in certain instances, progressed to routine clinical applications. This paper represents an attempt to delineate the practicability of external power versus conventional power and control sources. Both modes of power have been employed on a research and clinical level at the Institute of Rehabilitation Medicine, New York University Medical Center. The results of this experience are presented in this paper.

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In choosing one power source over the other, one must bear in mind that external power in itself does not necessarily bring about improved function. It is generally agreed by most investigators that, if at all feasible, and if there is no appreciable difference in the functional end-result, body power should be chosen in favor of external power. The reasons for this choice are obvious. At the present state of the art, body powered devices are less complex and require, therefore, less maintenance and are likely to be of lighter weight. More important, the basic design of a prehension orthosis, for example, is identical in the type of pinch provided, whether external or body power is used. This means that there is no inherent improvement in the terminal function of externally powered devices. External power is, however, indicated when body power is insufficient to activate the orthosis. Control of an externally powered device requires little force and range of motion at the control site, but does not provide feedback in an anesthetic limb. Control of body powered orthoses, on the other hand, provides some means of feedback because of a certain proportionality of excursion and force between the control site and the actuating device. So far, most of the research in externally powered devices and their routine clinical application has concentrated on the upper extremity. There are, however, a number of interesting lower-extremity developments which offer some functional improvements over conventional braces.

LOWER EXTREMITY

External power applications in lower-extremity orthotics have been confined to a relatively small number of designs. A likely reason for this lack of sophistication is that a paralytic lower extremity can be stabilized by very simple means, i.e., limiting or eliminating ankle, knee, or hip motion with a conventional brace. Although this enables the patient to ambulate, most conventional braces produce an abnormal gait pattern and an increase in energy consumption by blocking joint motions. This may not be of great consequence in unilateral involvements but imposes severe limitations in locomotion when the patient is paraplegic.

This problem has been attacked by Dr. Liberson by motorizing the hip joints of bilateral long leg braces and pelvic belts. The electric torque motors are designed to alternately drive the braces in opposite directions, thus producing hip flexion on one side while producing hip extension (push-off) on the opposite side. Control of hip motion is obtained through switches placed in the patient's shoes. This development is still in the experimental phase, but holds great promise in future developments in providing the paraplegic with greater mobility.

Another approach eliminating the brace entirely has been proposed by Liberson, et al., as well as Moe and Post. It is a muscle stimulator used in drop foot conditions, due to upper motor neuron disorders. An electrode placed over the peroneal nerve area provides a stimulus to pull the foot into dorsiflexion and eversion. A switch placed in the heel portion of the shoe is used to interrupt the muscle activating pulse in the stance phase of gait. The power pack of the muscle stimulator is carried around the waist.

An alternate solution to providing more nearly normal locomotion with a brace is the incorporation of a hydraulic stance and swing phase system between the ankle and knee joints, rather than the application of external power (Figure 1). Developed at the Institute of Rehabilitation Medicine, this system is in the experimental fitting phase at the present time. It is designed to provide stability at the knee during the critical period from heel strike to mid-stance and at the

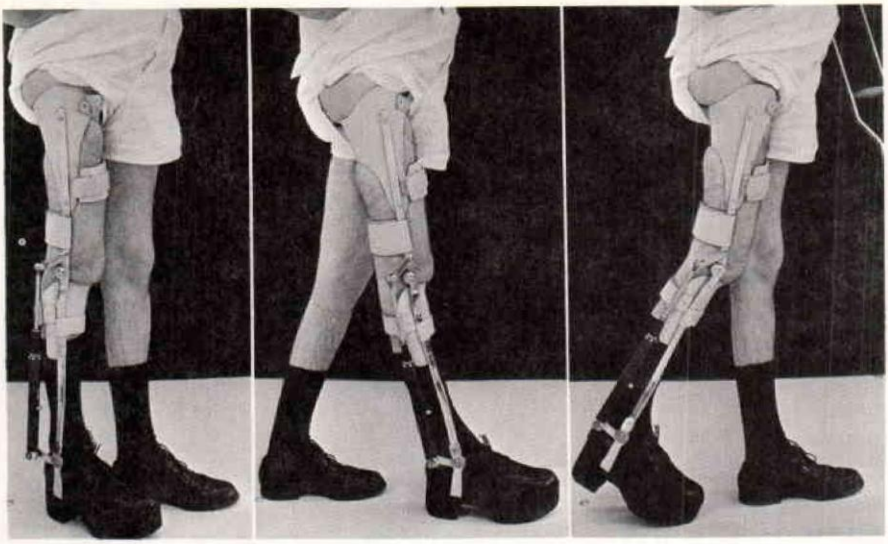

FIGURE 1—Above knee brace with Hydra-Nu-Matic cylinder for coordinated knee-ankle control.
FIGURE 2—The Hydra-Nu-Matic cylinder offers controlled fluid resistance to plantar flexion.
FIGURE 3—Controlled knee flexion offers a more nearly normal pattern of gait.

same time offers controlled fluid resistance to plantar flexion (Figure 2). Plantar flexion causes the hydraulic fluid in the cylinder to be displaced upward, resulting in an extension moment about the knee joint. This reciprocating action also comes into play in the swing phase, where knee flexion produces dorsiflexion of the foot. A 90 degree dorsiflexion stop is used for standing stability and to substitute for push-off. In allowing controlled knee flexion, a more nearly normal pattern of gait as well as a reduction in energy consumption is achieved (Figure 3). This design is still in the early stages of development and more clinical applications are needed to determine the practicability of such a design in terms of mechanical wear and maintenance.

UPPER EXTREMITY

The task of providing useful hand and arm functions is much more complex than that of providing ambulation by orthotic means. Hand and arm functions are generally much more important in the activities of daily living and vocational pursuits than lower extremity function. One may, in fact, consider the hand an extension of the brain, as we employ our hands not only for physical activities but also to lend greater expression to the spoken word and to enhance the effectiveness of speech.

The application of external power usually depends on the patient's residual motor power, i.e., if it is insufficient to activate a body-powered orthosis, external power is indicated. There are, however, conceivable exceptions to this rule. The use of external power should be explored in certain applications even when sufficient body power is present. For example, a wrist-driven prehension orthosis is commonly used to provide pinch when the patient has residual wrist extensor strength. Although adequate body power is available, activation of the device involves motions not only at the wrist but also compensatory motions at the elbow and shoulder joints in order to maintain the hand over the object to be grasped. It would seem that with further development of external power a more efficient mode of finger prehension could be obtained. On the basis of clinical experience at the Institute of Rehabilitation Medicine, patients fitted with gas or electrically-driven prehension orthoses tend to use their devices more regularly than patients fitted with wrist-driven prehension orthoses. This tendency may not only be ascribed to the fact that externally powered orthoses are fitted to patients who sustained a cervical cord lesion above the C 6-7 level and are, therefore, more dependent on orthotic devices. Rather, it is very likely that functional performance is improved because compensatory motions, needed with wrist-driven prehension orthoses, are not necessary for prehensile activities.

Over the past three years electrically-driven prehension orthoses have been used for patients who lack wrist extensor strength (Figure 4). The power pack consists of a nickel cadmium battery, a permanent magnet 12 volt motor, and a charger, permitting the patient to recharge the battery from a regular household outlet. An adjustable slip clutch (Figure 5) is used to selectively adjust the pinch force as well as to provide a safety mechanism in case of switch failure. The orthosis is activated by an unidirectional microswitch usually

ELECTRICALLY – DRIVEN PREHENSION ORTHOSIS

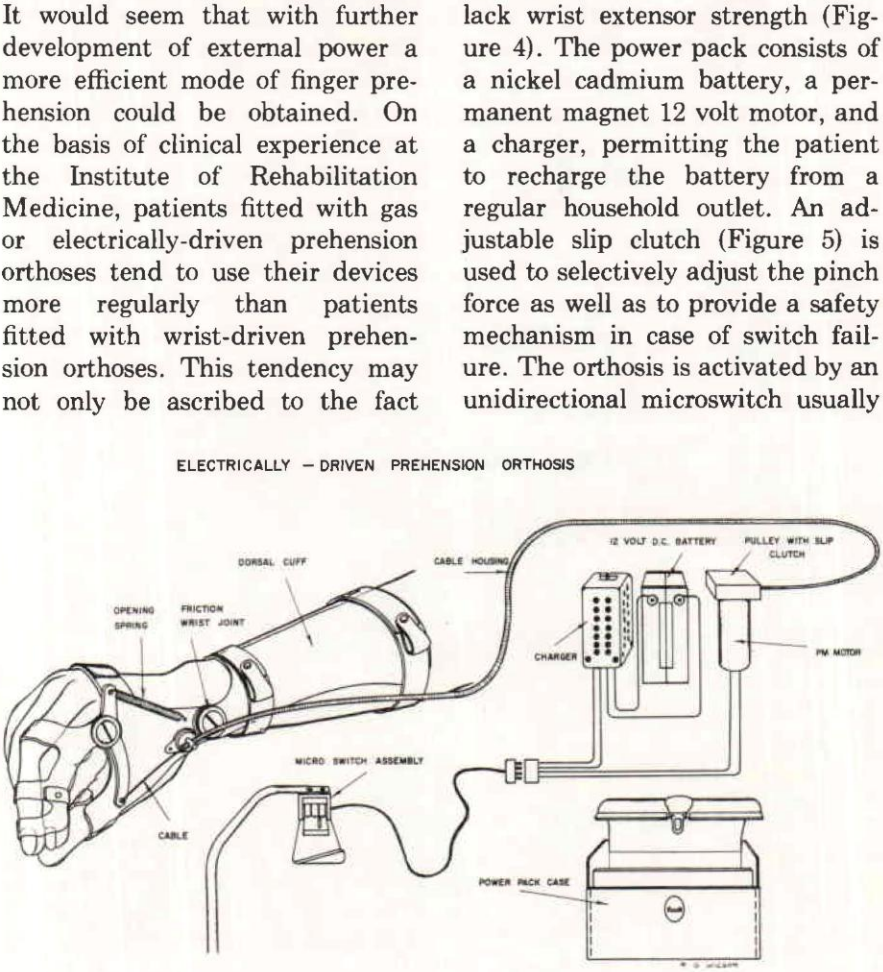

FIGURE 4—Electrically-driven prehension orthosis.

placed superior and posterior to the contralateral shoulder (Figure 6). This control site has been found to be most effective in leaving the fitted extremity free to move without inadvertently operating the orthosis. It provides a more nearly synergistic control motion since one normally uses a certain amount of "body English" when reaching for objects. Thus, a sound kinematic mode of control is achieved which requires little or no patient training.

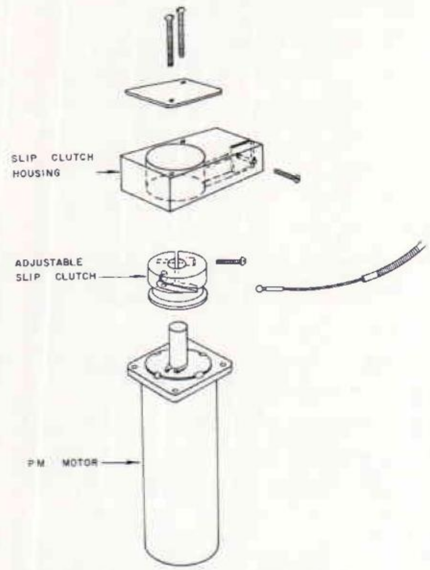

FIGURE 5—Exploded view of adjustable slip clutch for electrically-driven prehension orthosis.

The application of external power would seem useful even in patients who have good wrist extensors and who would conventionally be fitted with a wrist-driven prehension orthosis. Thus, there would be no need for compensatory motions of the other arm joints for terminal device operation. The orthosis would not have to extend above the wrist, leaving the upper extremity with the optimum degree of freedom. Although such fittings are possible, they are not practicable at the time with the conventional mode of orthotic control. Further development of

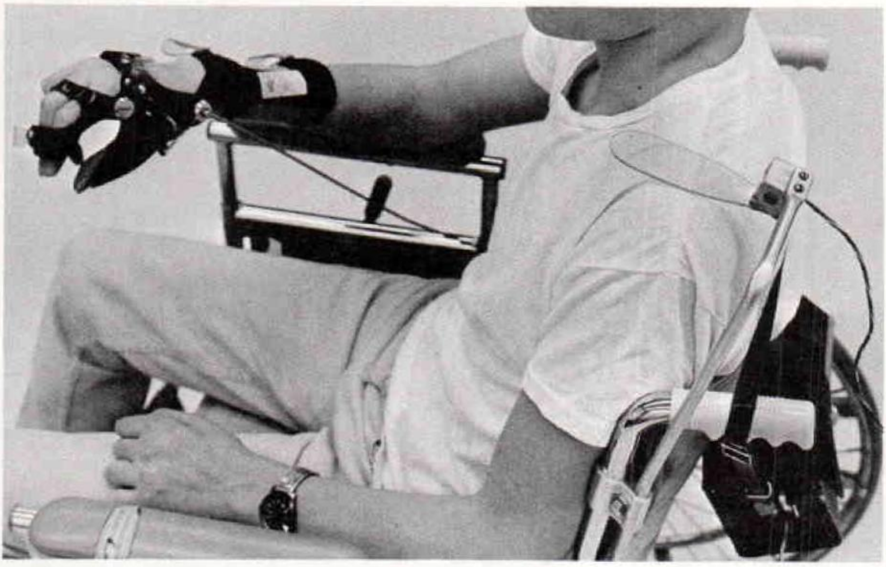

FIGURE 6—Unidirectional microswitch is placed superior and posterior to contralateral shoulder.

FIGURE 7—Three state myoelectric trainer unit.

myoelectric control systems may, however, lead to development of a prehension orthosis for the C 6-7 quadriplegic patient which would permit freedom of the wrist. This is possible by picking up myoelectric control signals from suitable forearm musculature. The three state myoelectric trainer developed at the University of New Brunswick is utilized to test the feasibility of such a system (Figure 7).

The results of external power applications in ambulatory patients has not proved to be as satisfactory as it has for wheelchair-bound patients. The additional weight of the power source and actuator which the patient has to carry are difficult problems to overcome. Furthermore, ambulatory patients possess greater mobility which may increase the frequency of inadvertent operation. It has been our experience that for ambulatory patients body powered and/or manually controlled orthoses are preferred because of their simplicity and lighter weight. Patients who have no hand function can be successfully fitted with a shoulder-driven prehension orthosis (Figure 8), providing voluntary opening and spring closing (Figure 9). The spring closing feature necessitates the incorporation of a pressure relief mechanism to avoid skin breakdown of the thumb, index and middle finger pads during periods when the orthosis is not actively used. This is of utmost importance in patients with a brachial plexus or similar lesion, resulting in both motor and sensory losses. The pressure relief control consists of a spring-loaded push-button below the MP joint

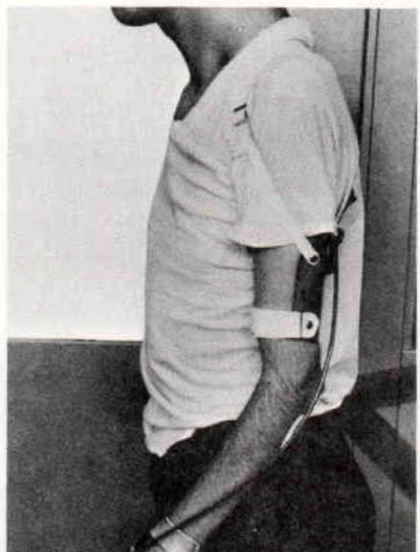

FIGURE 8—Shoulder-driven, cable controlled prehension orthosis.

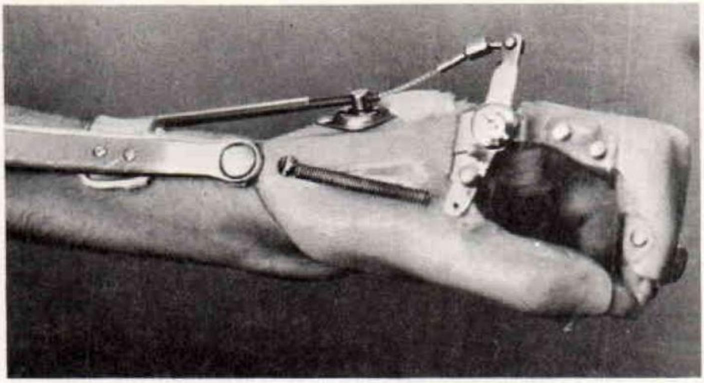

FIGURE 9—Voluntary opening-spring closing to three jawed chuck prehension.

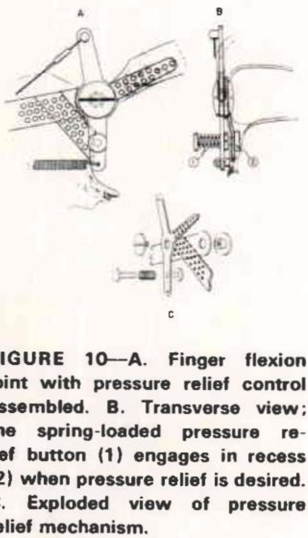

FIGURE 10—A. Finger flexion joint with pressure relief control assembled. B. Transverse view; the spring-loaded pressure relief button (1) engages in recess (2) when pressure relief is desired. C. Exploded view of pressure relief mechanism.

which is pushed into a semicircular recess. This prevents full finger closing. Increased cable tension causes the pushbutton to automatically retract, unlocking the pressure relief (Figure 10).

Manual control of arm braces for ambulatory patients, i.e., positioning and locking of the elbow at the desired angle of flexion, has proved more satisfactory than both external power or cable-controlled body power. Obviously, the force requirements for body-powered arm orthoses are infinitely greater than for an equivalent prosthesis because, in addition to the weight of the orthotic device, the weight of the patient's arm must be considered. While for unilateral arm braces manual control is most practical, in patients with bilateral arm involvement, external power is the only alternative to providing useful arm and hand functions. Here, CO_2 as a power source seems to have advantages over electric power because of the greater simplicity of the system. A CO_2 piston and cylinder elbow actuator obviates the mechanical elbow lock needed with the McKibben muscle substitute (Figures 11A and 11B). In the piston, the carbon dioxide is confined in a rigid-walled container, which affords sufficient resistance against elbow extension when forearm loads are applied. Activation of the control valve requires little force which can usually be obtained by harnessing residual arm motions (Figure 12). The wheelchair-bound patient requiring an arm orthosis poses again another problem. In this application electric power sources have been found more useful than carbon dioxide. Electric power actuators provide a more definite control when compared to CO_2 actuators, especially the McKibben muscle substitute. In the latter, the problem of rebound is difficult to overcome when loads of various magnitude are placed in the terminal device. The electric arm orthosis developed at the Institute of Rehabilitation Medicine is designed to provide the high level quadriplegic patient (C 4-5) with voluntary arm and hand functions (Figure 13). It allows the patient a total of five degrees of freedom, four of which are motorized. The power actuators are 12 volt permanent magnet motors located at the back of the wheelchair. A twelve volt battery serves as the power source for both the arm orthosis and the wheelchair. Power transmission from the electric motors to the moving orthotic arm segments is provided through Bowden cables (Figure 14). The motor used for powered prehension is equipped with an adjustable slip clutch designed to vary the force of prehension and to act as a safety device. The other motors are provided with limit switches. Motions motorized are:

1. Finger opening and closing to a jaw chuck type of pinch (Figure 15).
2. Pronation and supination obtained through a spiral shaft running in a nylon sleeve (Figure 16). A linear pull on the sleeve causes forearm rotation (Figure 17).
3. Elbow flexion and extension

FIGURE 11 (A & B)—Arm brace with CO_2 piston actuator to provide elbow flexion.

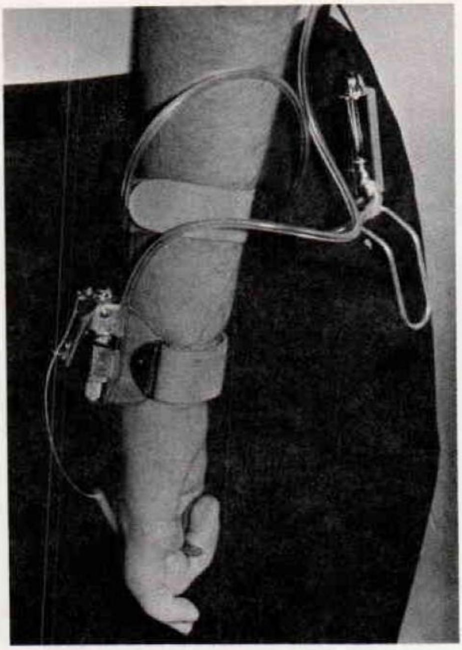

FIGURE 12—Residual wrist and finger flexion is harnessed to activate CO_2 control valve.

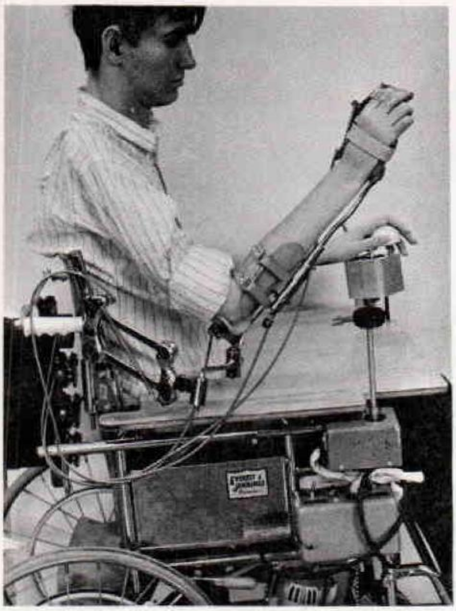

FIGURE 13—IRM electric arm orthosis.

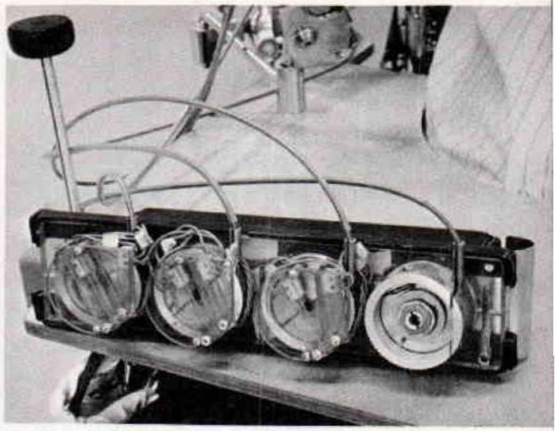

FIGURE 14—Power transmission from the electric motors to the orthotic arm segments is provided through bowden cables.

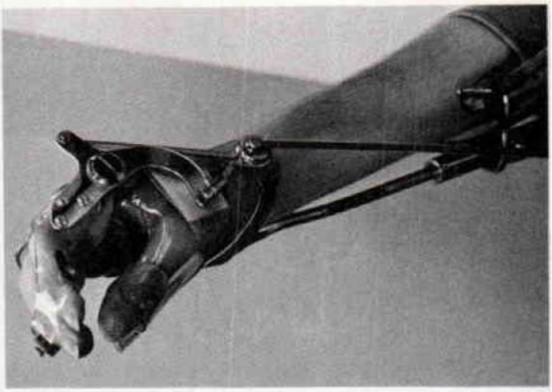

FIGURE 15—Finger closing to a three jawed chuck type of pinch.

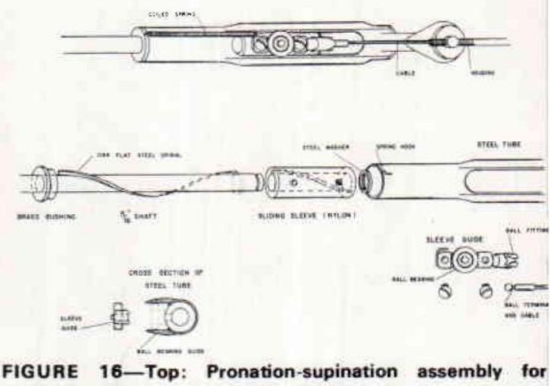

FIGURE 16—Top: Pronation-supination assembly for electric arm orthosis. Bottom: Exploded view.

FIGURE 17—Forearm rotation is caused by a linear pull on the sliding sleeve of pronation-supination unit.

FIGURE 18—Motorized elbow flexion unit.

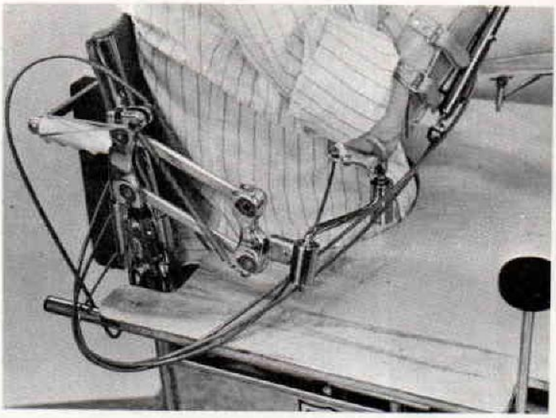

FIGURE 19—Parallel linkage lateral to humerus provides combined flexion-abduction and extension-adduction.

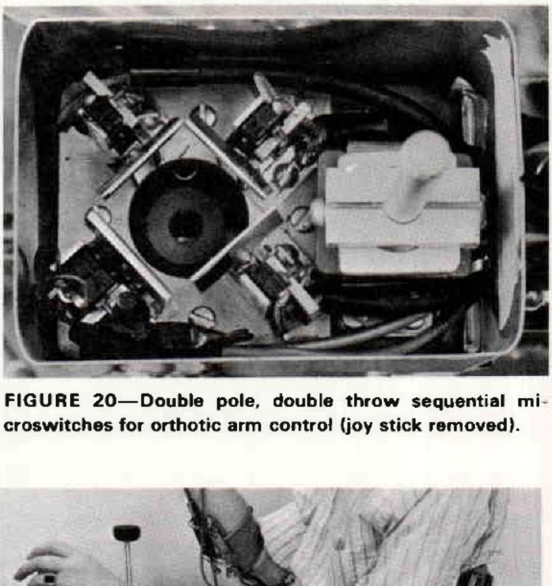

FIGURE 20—Double pole, double throw sequential microswitches for orthotic arm control (joy stick removed).

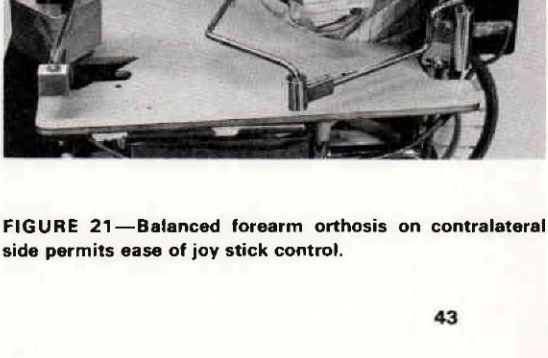

FIGURE 21—Balanced forearm orthosis on contralateral side permits ease of joy stick control.

through a spring-loaded pulley located medial to the elbow (Figure 18).

4. Combined humeral flexion-abduction and humeral extension-adduction through a parallel linkage, lateral to the humerus (Figure 19).

Horizontal adduction and abduction are not motorized since the type of patient requiring an electric arm orthosis is likely to have sufficient residual shoulder girdle control to produce such desired motion, once the effects of gravity are eliminated in a properly balanced linkage system. The electric motors are activated through specially designed double-pole, double-throw sequential microswitches (Figure 20). The patient's contralateral arm is supported in a balanced forearm orthosis to permit ease of switch control (Figure 21). This is possible by a shift of the center of gravity, induced by head motion, in combination with residual shoulder and arm motions.

SUMMARY

An attempt was made to delineate the practicability of external power in orthotic applications. In lower extremity orthotics the application of external power is, at the present time, restricted to relatively few, mostly experimental devices. Developments in this area which hold promise are motorized hip joints for paraplegic patients, electrical stimulation to evoke muscle contraction in upper motor neuron disorders, and as an alternate to external power, hydraulic controls to coordinate knee and ankle motions in the above knee brace. In upper extremity orthotics the application of external power depends, to a great extent, on whether the patient is ambulatory or wheelchair-bound and whether he is unilaterally or bilaterally involved. In the ambulatory, unilaterally involved patient, shoulder-driven, body powered and/or manually controlled orthoses have found greater patient acceptance and are of greater practicability in terms of weight, wear, and maintenance than externally powered devices. If, however, the patient is wheelchair-bound or an ambulatory patient with bilateral arm involvement, the indications for external power are definitely within the realm of practicability. It provides, in most cases, the only means of obtaining useful hand and arm functions.

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