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Modular Externally-Powered System

IN CONVENTIONAL PROSTHETIC SYSTEMS, intact portions of the amputee's body are harnessed to provide a means of operating upper extremity prostheses. By motions such as hunching the shoulders and thrusting the arm forward, the amputee operates control cables. Like tendons in the body, the cables transmit mechanical power that operates terminal devices or joint articulations of the prosthesis. Harnesses designed to develop mechanical forces are, of necessity, tight and restrictive and, consequently, are uncomfortable. Amputations at higher levels proportionally reduce the ability of the amputees to produce the cable forces (15 to 30 pounds) and displacements (2 to 4 inches) required to operate conventional devices. The "work envelopes" of body-powered prostheses are limited as a result of restrictions imposed by the harness. Harnessing techniques that facilitate good control of more than two functions have not been developed; the harness becomes too complex for the patient to don and operate. One possible solution to these problems is to supply power from an external source rather than from the body, so that the

amputee need supply only the control input.

In a collaborative effort, the Applied Physics Laboratory and The Johns Hopkins Medical Institutions have been examining the feasibility and applicability of a powered prosthetic system for upper limb amputees. The program, sponsored by the Prosthetic and Sensory Aids Service of the Veterans Administration, is two years old and is now at the stage where experimental units are being evaluated on eight amputees.

System Concept

The basic concept employed in the APL/JHMI effort uses a proportional control power pack that can either be located remotely from the prosthesis (worn on the belt) or placed within the prosthesis (for elbow or higher amputation). Such a powered control system can be used with many of the components that are standard in the prosthetics industry. In such applications, a single power unit can control more than one function.

The system elements in schematic form are shown in Fig. 1. The input to the system is a

Many upper limb amputees, especially those who have sustained high level amputations or injury to remaining portions of the body, are unable to use conventional prosthetic devices effectively. In an effort to help the more severely handicapped amputees and those who may for other reasons require capabilities that standard prostheses cannot provide, the Applied Physics Laboratory, in conjunction with The Johns Hopkins Medical Institutions, has developed a prosthetic system which is powered by rechargeable batteries. The amputee need supply only a control "signal." This article describes the concept which is now being evaluated with the aid of amputee subjects, and discusses results obtained to date in field tests.

for Limb Prostheses

myoelectric signal (EMG) generated by muscles over which electrode sensors have been placed. This signal is amplified from its inherent low level of 10 to 1000 microvolts RMS, and is then detected and filtered to provide the command signal to the power servo. In order to minimize electrical power consumption, particularly when the control voltage is not commanding the motor to pull the control cable, a pulse-width modulation system is used to control the motor output torque. The output of a 25.0 kHz triangle wave generator and the output of the servo amplifier stage are combined to provide a pulse-width modulated signal which controls motor current. The motor is driven in one direction only. If the error signal (which corresponds to the difference between control signal amplitude and the position of the cable drive pulley) is small, motor current flows for a small part of the duration of each cycle of the 25.0 kHz waveform; the "on" time of the pulse is a function of the magnitude of the error signal. By operating the output transistor in a switching mode, relatively little power is dissipated in the electronics package. When the electrodes are not

sensing an EMG signal, standby power consumption of the servo and sensor electronics is less than 300 milliwatts. No mechanical switches or special

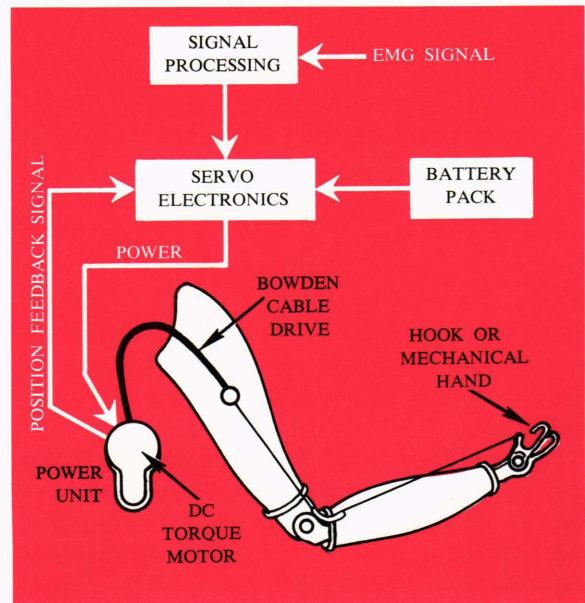


Fig. 1—Block diagram of prosthetic control system.

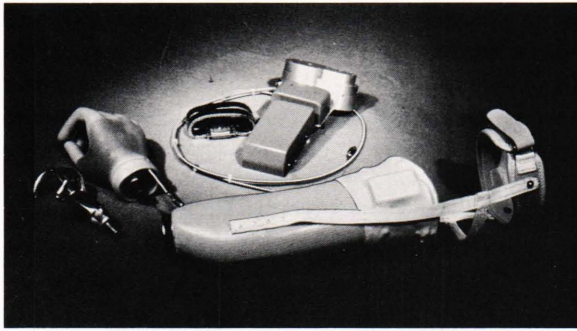


Fig. 2—Components of wrist-disarticulation prosthesis.

power cutoff relays or circuits are required to switch from “standby” to “operate” condition.

A potentiometer in the power pack assembly provides a position signal to the servo amplifier. The feedback shaping provides high gain at low frequencies and less gain as the frequency is increased. Such signal processing makes the opening of the terminal device very easy to control at all elbow flexion positions with or without an object in it.

The power unit employs a low-speed, high-torque DC torque motor to control the terminal device or elbow motion by means of a Bowden cable drive. Since this cable drive is similar to that employed on conventional body-powered prostheses, many conventional, readily available prosthetic components may be utilized with this system.

The adaptability of this system concept is demonstrated by Figs. 2 and 3. Figure 2 shows all the components of a wrist-disarticulation prosthesis. Myoelectric control signals are obtained at one control location on the forearm. Mechanical power is transmitted through a Bowden cable from the motor unit, which is worn on the belt, to the terminal device. A variety of conventional terminal devices, such as the hook and anthropomorphic hand shown in Fig. 2, can be used without modification. The battery pack, which permits in excess of 500 operations without being recharged, and the servo-control unit are also worn on the belt.

Figure 3 shows a prosthesis in which a motor unit (identical in basic design in that shown in Fig. 2) has been modified to control both elbow and terminal device operation. This prosthesis is controlled by signals obtained at a single control site on the upper arm. The motor assembly is fitted in the upper part of the prosthesis, and the

electronic control unit is located in the forearm. Only the battery pack is worn at the waist. When the elbow is locked, the terminal device is proportionally controlled. With the elbow unlocked, the forearm can be raised to any position the amputee wishes in less than one second. Operation is smooth and quiet, and more than 600 operations of the prosthesis can be provided with one battery charge.

Electromechanical Design Considerations

The philosophy followed in designing the experimental systems was based on the premise that a rugged mechanical power unit which could control more than one function was essential. The system must require minimal control effort on the part of the amputee and little or no maintenance. Some of the rationale which went into design of the basic modules is given in the following paragraphs.

Myoelectric Sensor Input—The single-site myoelectric signal sensor system was selected as the primary control input. The system was de-



Fig. 3—Above-elbow prosthesis with power unit in the elbow and battery pack on the belt.

signed to operate from signals obtained from only one location on the amputee's body to relieve the difficulties attendant with applying multiple electrodes each time the amputee dons the system. Many additional advantages accrue as a result of this choice. The first is the ease with which amputees can learn to use these systems—they need to learn to exercise control of only one set of muscles. The system is also designed so that it is nonsequential, i.e., there is no requirement that the amputee remember to where, in a sequence of operations, the system has been advanced.

Amputee subjects have worn these sensors as part of their externally-powered prosthetic systems for periods ranging from a few weeks to over a year. Signal levels of all of the amputee subjects have remained consistently high, and interference from 60 Hz domestic power sources has been found to be a negligible problem even when operating the system in high 60 Hz environments. This result is believed to be due to the careful attention paid to this problem in the electronic signal processing circuit design and the use of the ground guard ring that encircles the electrodes.

For one below-elbow amputee there was a dramatic improvement (by a factor of two or three) in EMG signal level within one week after the prosthesis was fitted. The signal level then remained at this high level. To date, the experience of all of the amputee subjects has been that the units will operate satisfactorily without attention all day or until the prosthesis is taken off. The skin need be dampened only once a day when the system is first put on.

Power Unit—The direct-drive DC torque motors which are used in these units are very rugged and provide high torque when stalled and when operating at low speeds. Their high mechanical power output capability makes possible fast actuation (less than one second) at high force levels (up to 30 pounds cable force) with a simple, two-stage gear reduction system. With this low gear ratio, noise is sufficiently low so that the system can scarcely be heard when it is worn by the amputee. The power units have no clutches, brakes, or mechanical stops, and are very reliable. There have been no malfunctions or mechanical failures of any of the motor drive units that have been used in the amputee evaluation program.

The motors can be held stalled at full power without internal damage of any kind. No current limiting circuits are required, therefore allowing the utilization of a motor drive circuit which conserves battery power.

Electronic Unit—The electronic control units are very conservatively designed, using quality commercial grade electronic components. Discrete components and hermetically sealed operational amplifiers are mounted on printed circuit boards. There has been no electronic circuit failure over the past year in any of the systems that are being evaluated. The systems have been worn several months without any adjustment of the controls that have been provided to permit initial adjustment for differences in myoelectric signal levels of different amputee subjects.

Interface Wiring—Interface wiring was recognized early in the program as a potential source of trouble and, in fact, has been a source of difficulty on two units. Wires which interconnect the sensor, motor unit, electronics unit, and battery pack are subjected to strain and twisting in normal wear and especially when the amputee dons the system. As it is assembled, each system is analyzed to determine potential trouble sources. Several precautions have been taken to reduce the possibility of failure. Electrical connectors have been eliminated where possible; electrical disconnects are used only at the battery pack and at the electronics unit. The special flexible wire used by the telephone company to connect desk sets to wall boxes is the most satisfactory wire tested in these experiments. This wire is used for all signal interconnections in the system. To relieve the strain on the electrical wiring, $\frac{3}{64}$ inch stainless steel prosthetic cable is run along with the electrical wires inside an outer cable jacket where heavy units are interconnected. This has proven to be very effective in reducing wire breakage. Wherever possible, units of the system are mounted together on rigid back-plates and wiring is concealed.

Battery Pack and Charger—All of these systems utilize twelve rechargeable nickel-cadmium AA size (penlight) cells with 500 milliampere hour capacity. A diode is placed across each cell to minimize reverse polarity effects when the cells run down. The battery pack can be fully charged

in four hours using conventional charging circuits. Each amputee is provided with two battery packs and a charger.

To date, there have been no electrical failures of any battery packs. Vacuum-formed polyvinylchloride cases were used for the first battery packs, but were subject to breakage when the battery packs were dropped. Another material (Kydex) is now being used and has been satisfactory; there has been no breakage of the Kydex covers even when the battery packs were dropped.

Amputee Evaluation Results

Eight amputees were selected to help provide evaluation data on the Johns Hopkins unit. In order to determine the range of applicability of this powered system, amputees were selected which represented a broad range of the possible upper limb amputation levels. Included in this program were one wrist-disarticulation amputee, two below-elbow amputees, two above-elbow amputees, two elbow-disarticulation amputees, and one shoulder-disarticulation amputee. Following is a summary of the results of the evaluation to date on these amputees.

Unit #1—This unit was fitted to a left-wrist-disarticulation amputee in February 1970. It has a laminate forearm socket suspended only by a supracondylar strap. The motor, electronic control unit, and battery pack are worn on the waist. The EMG sensor is built into the forearm stump socket in a floating arrangement and is located over the junction of the common extensor tendon and the extensor muscle bellies. Proportional opening of the terminal device is controlled by varying the EMG signal in the desired manner. The unit was evaluated in three phases.

During the break-in phase, which lasted for several weeks, the amputee wore the prosthesis for graduated periods while performing increasingly demanding manual skills. Design features which caused wire breakage and other mechanical and functional problems were identified and corrected. By practicing various manipulations the amputee attempted to develop maximum dexterity.

In the comparison phase, various physical and functional measurements were made comparing the new prosthesis to the amputee's pre-existing body-powered (BP) prosthesis. These include

the span, force, velocity, and fine control characteristics of the terminal device grasp, the range of terminal device placement, the bimanual work envelope, and grasp-placement coordinations, as well as stability, weight, comfort, and speed of application. Test activities were recorded on movie film for review and further analysis.

In the final phase, arrangements were made for the amputee to wear the new appliance as his primary prosthesis for all of his activities for several months in order to reveal undiscovered problems and to determine mechanical endurance and ultimate amputee preference.

After wearing the unit for one year, the amputee reported the following: For delicate work the body-powered prosthesis is superior to the externally-powered (EP) one owing to (a) shoulder muscle feedback, (b) the capability of setting the shoulders for steady force, (c) lack of lag, and (d) higher speed. Because of the time lag and lower velocity, he is inclined to use his EP prosthesis primarily in a "bang-bang" mode. He dislikes battery and motor bulk and weight, especially at the end of the day. He has reported that the EP prosthesis is ideal for working above his head or within closely restricted space, such as when lubricating the underside of an automobile, or for conditions which render shoulder motion undesirable or difficult, such as when propped up to read in bed. The absence of a shoulder harness and lack of compression force on the end of the amputation stump are definite advantages of the EP system if and when the end of the amputation stump is tender. This amputee has been able to use a snugger socket with the EP system than with the BP system, because of this feature. He has no sensor site skin sensitivity problems despite a known sensitivity to nickel, the metal rivets in his lower limb prosthetic socket, and his wrist watch. The net result of all factors is that he likes to have his externally-powered system for special uses and for relief from his body-powered prosthesis, especially in hot weather. The average EP use is now two or three days per week.

Unit #2—This unit, shown in Fig. 3, was fitted to a right-above-elbow (AE) amputee in August 1970. One year prior to this time he had been fitted with a body-powered AE prosthesis with internal locking elbow, AE figure-eight harness, dual control quick-disconnect wrist, volun-

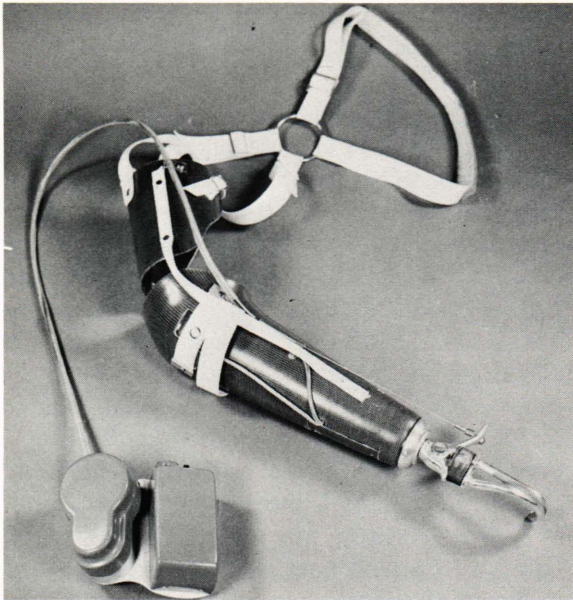


Fig. 4—Below-elbow prosthesis with belt-mounted power unit and battery pack.

tary opening (VO), hook, and an Army Prosthetics Research Laboratory (APRL) hand. He developed moderate facility with this equipment but never used the hand. The EP power unit is located in the elbow space and controls either the terminal device (voluntary opening function) or elbow flexion, using a routine external locking cable and strap to the shoulder saddle in order to select terminal device vs. elbow function. The battery for this system is located on the belt. The single site EMG sensor is mounted in the stump socket over the biceps muscle.

The amputee clearly prefers the externally-powered prosthesis to his body-powered system and uses it all the time. Some harness adjustment, repair of the elbow locking cable attachment, and replacement of a broken rubber band (which closes the hook) have been necessary, but these were unrelated to the power system. There have been no breakdowns or adjustment problems in the EP system.

Although he states he had actually preferred the additional weight of his EP prosthesis compared with his BP one, he also states that the weight becomes objectionable if his activities require him to be standing or walking and without additional support for his prosthesis for durations of time exceeding three hours. For this reason he feels that his type of limb might be too tiresome

for certain types of outdoor employment. On the other hand, if he is able to sit down and thereby rest the prosthesis in his lap or on a table for a few minutes every couple of hours, the externally-powered prosthesis causes him less fatigue than the body-powered one; therefore, he feels that the limb is especially suited to persons doing office work or to persons whose work activities entail occasional periods of sitting, as in driving a motor vehicle.

Preliminary analysis of experience with his EP system indicates that its advantages include ease of operation and an enlarged work envelope.

Unit #3—This system, shown in Fig. 4, was fitted to a right, mid-below-elbow amputee during October 1970. His power unit and battery were initially designed to be clipped onto the belt with the power being transmitted to the prosthesis by a Bowden cable. A more permanent type of belt mounting such as is illustrated in Fig. 5, with the battery and motor on opposite sides, was subsequently found to be much more comfortable, secure, and less obstructive.

Prior to the fitting with his externally-powered prosthesis he had had one year of experience with his body-powered limb. Reports of his EP limb use time have varied from "most of the time" to "almost every day." This individual was fitted very snugly in order to obtain maximum stability of the socket on the stump, especially, to facilitate playing the piano. It is not possible to fit the socket for his BP prosthesis as snugly as that for his EP without causing discomfort in the end of

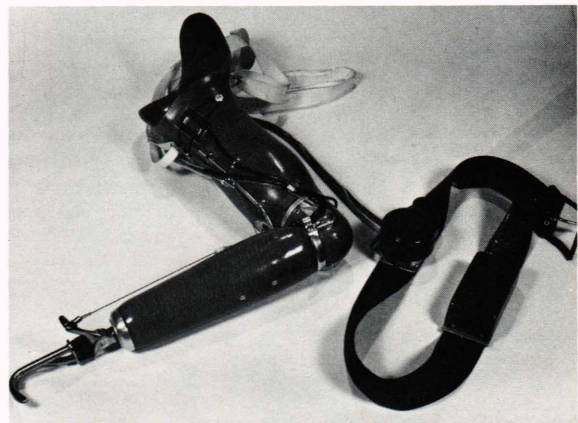


Fig. 5—Elbow-disarticulation prosthesis with power unit and battery pack on belt.

the stump. This is believed to be due to the development of compression forces in the end of the stump where the socket compresses it in reaction to the tension forces developed in the cable each time the terminal device is opened. In the EP prosthesis these compression forces are developed in the cable housing rather than in the soft tissues in the end of the amputation stump. This amputee has said that a significant reason for not wearing his EP prosthesis even more than he does is the extra time and effort required to apply the tightly fitting socket.

This amputee had difficulty on some occasions with slow operation of the terminal device. He reported that it seemed to be dragging. The problem was not reproduced when the prosthesis was examined and then reapplied to him. Ultimately, it was discovered that he was inadvertently twisting the cable as many as several revolutions in the process of donning the belt and prosthesis. This problem was solved by teaching him to use the colored wires which are attached to the cable as an indicator of twisting.

He had a very weak EMG signal when he was first fitted. After the first week of wear, his signal improved by a factor of 3, necessitating a readjustment in electronic system gain. His EMG output has remained constant at this high level since that time. This amputee is a piano student. He has been able to continue his piano lessons, and is developing a good facility for the control of his



Fig. 6—Special-purpose terminal device used with below-elbow prosthesis.

EMG signal while playing the piano, using the specially designed terminal device shown in Fig. 6.

Unit #4—This unit, shown in Fig. 5, is fitted to a right-elbow-disarticulation amputee who had only two months experience with his body-powered prosthesis prior to being fitted with the externally-powered unit. A conventional Hosmer internal locking elbow was used, the lock being activated in the traditional manner by force transmitted from the shoulder through an external cable connected to the elbow lock control strap. The power unit and battery are permanently mounted on the belt. The battery mount is designed to permit convenient battery replacement. Force transmission between motor and terminal device is by Bowden cable. The terminal device is of the voluntary opening type. The limb was delivered in December 1970.

This amputee's original injury was incurred at work in the process of operating machinery used for processing soap. In January 1971 he returned to the same full-time job with the same employer. He uses his externally-powered prosthesis from the time he gets up in the morning at least until he returns from work in the afternoon. Usually he takes the limb off at home, but puts it back on for any bimanual activities or to go out in the evening. He has worn the limb continuously as long as 14 hours. On one occasion he dropped and broke a battery case. Since delivery of his EP prosthesis, the only times he has worn his body-powered prosthesis was while this battery case was being replaced and on two other occasions when his EP prosthesis was in the laboratory for checkout. The case was replaced with one of a more durable material and no further breakage has occurred. He has reported inadvertent openings of his terminal device in the washroom at his job when trying to hold wet toweling in order to wash and dry his remaining hand. In this situation he has resorted to switching off the battery in order to maintain a grip on the toweling. The sensor case was modified to improve electrode contact. He has had no other malfunction or mechanical or electrical breakdown. In reply to inquiry about speed he said that ideally he would like to have it a little faster. In reply to inquiry about the weight of his prosthesis he stated, "No problem. I guess I am used to it." Regarding the components on his belt he answered, "A little bulky, but

they don't bother me none." He stated that, "The arm is working beautifully. It will do anything you try to do with it."

Unit #5—This unit, shown in Fig. 7, was fitted to a short below-elbow amputee who had already learned to use a body-powered prosthesis with Muenster socket and APRL hook or hand. The externally-powered prosthesis has a Muenster socket. There is no additional suspension or harness. The power unit and battery are permanently belt mounted, but the battery mount permits convenient battery replacement. Force transmission between motor and terminal device is by Bowden cable; hook and hand are of the voluntary opening type. Prior to fitting with the EP prosthesis the amputee had stated that he was bitterly disappointed with the functional capabilities of his BP prosthesis. Although he usually wore this prosthesis, he did so almost entirely for appearance alone and rarely used the hook.

This amputee adapted immediately to the EP prosthesis without any special training. Since delivery of this system, he has used it exclusively, full-time, every day except on two occasions when he removed it for a few hours because of some skin irritation at the socket sensor port. It was felt that this problem was due at least in part to over-zealous wetting of the skin in this area at the time of application of the prosthesis and to excessive pressure of the sensor. The condition improved with corrective action and no longer interferes with the use of the prosthesis. It is evident that this amputee is cycling his EP prosthesis more frequently than he formerly cycled his BP prosthesis. He reports that his battery lasts him about nine hours, and, in order to permit continuing function of his limb, he carries an extra battery with him when away from his house for the whole day. He is a part-time farmer and mechanic. He likes to use the prosthetic hand in a work glove when working with farm hand tools. He prefers the hook when working with smaller metal objects. He reports that although the belt-mounted components seemed annoying at first, he is no longer aware of them. When sitting, he rotates his belt slightly, thereby bringing the motor around to his side where it does not press against his back. The speed of operation is satisfactory, but he would like for the hand to close a little faster. There have been no equipment breakdowns

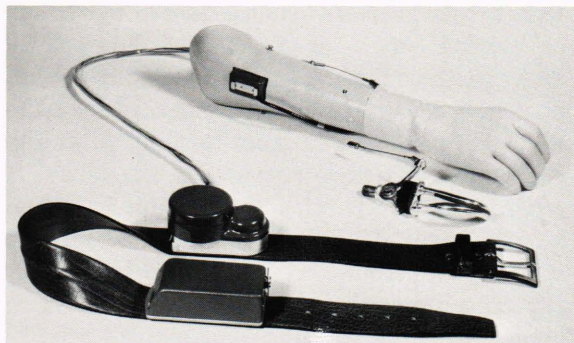


Fig. 7—Below-elbow prosthesis with anthropomorphic terminal device.

or malfunctions. He denies having any trouble with inadvertent openings but says that occasionally when applying high force to an object, he switches off his battery to maintain the grip. He states that it will be terrible for him when the system is retrieved when the evaluation is completed, and he complains that he does not know how he will get along when he no longer has it.

Unit #6—This unit is fitted to an above-elbow amputee who lost his left upper limb in an industrial accident six years ago and has been using a body-powered prosthesis for approximately five years. He is a full-time employee of an industrial contracting company where he works as a high-pressure boiler welder. He is also a private horse farmer. His BP prosthesis is of heavy duty construction and has an internal locking elbow and farmer's hook. He is very skillful with his BP prosthesis, subjects it to heavy use, and has been wearing it full-time, all the time.

Since he depends heavily on the humeral rotation turntable in his work and since his stump is fairly long, a special turntable was designed and constructed that occupies minimal longitudinal space and still permits placing the motor in the elbow region without undesirable lowering of the elbow center. The sensor was placed in a socket port over the biceps muscle; the remaining electronic components were placed in the forearm unit and the battery is worn on the belt. One week after the externally-powered limb was delivered, the amputee reported that the above-elbow figure-eight harness was uncomfortable, owing to the increased weight of the prosthesis. The harness was then converted to a shoulder saddle with cross chest strap. The patient reports that this new harness system is more comfortable. He also

reports that his work requires him to be standing or walking almost all of the time. He finds that under these circumstances the externally-powered prosthesis is very tiring and that he must switch to his BP prosthesis in order to continue working efficiently for periods longer than five hours. He reports relief from tiring if he is able to sit, and he states that the prosthesis would be ideal for work which permitted resting its weight occasionally on a supporting surface such as a desk top or in his lap. He believes that part of his fatigue might be due to unaccustomed use of his biceps muscle. He has had no malfunctions or breakdowns and denies inadvertent openings. A fresh battery pack lasts him five hours of work time. He states that the response of the limb is good and he hopes to develop an increased fatigue tolerance. Until the weight problem is solved, this amputee shows a clear preference for his BP prosthesis.

Unit #7—This unit was fitted to a 56-year-old left-elbow-disarticulation amputee about one week after delivery of his first body-powered prosthesis and nine months after his amputation. The original injury to his left upper limb also resulted in severe permanent limitation of motion in his left shoulder. Due to the limited muscle power and excursion available, the harness selected for the BP prosthesis consists of a shoulder saddle and cross chest strap for suspension and a separate axillary loop for transmission force from the right shoulder to the main cable. This loop is designed to encircle the upper arm near the belly of the deltoid in order to exploit some sound side shoulder motion to obtain maximum cable excursion. In spite of these special considerations and daily special training efforts, the amputee is having great difficulty developing any useful function with his (BP) prosthesis.

The externally-powered unit has a socket, forearm unit, wrist unit, and voluntary opening terminal device which are identical to those of the BP unit. An identical shoulder saddle and cross chest harness are used, but the axillary loop was omitted from the EP unit. In the EP unit the main cable is routed around a pulley in the forearm unit. The power unit and battery are mounted on the belt as in Fig. 5.

The subject was able to function both the elbow and terminal device with good control immediately

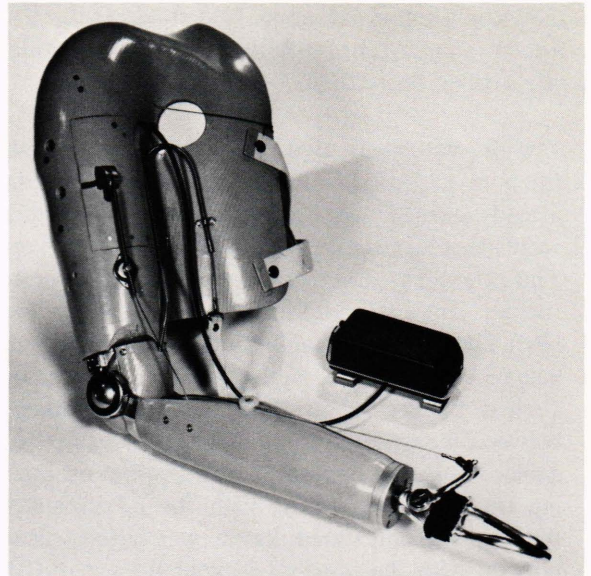


Fig. 8—Shoulder disarticulation prosthesis.

after delivery of the limb and without special training. Since delivery, no further adjustments have been necessary. There have been no malfunctions or breakdowns. Thus far, the amputee reports that he is delighted with all aspects of his EP limb, but that he is very discouraged about his BP limb.

Unit #8—This prosthesis, shown in Fig. 8, was fitted to an 18-year-old right-shoulder-disarticulation amputee to replace the limb lost in a traumatic amputation secondary to a corn picker accident.

The amputee was fitted with a conventional body-powered prosthesis approximately 3½ months after the amputation. This device provided negligible function and he wore it (for appearance only) on special occasions. Prior to being fitted with the externally-powered prosthesis he had, for all practical purposes, totally rejected the conventional system.

No myoelectric signals suitable for control of the prosthesis were found when a thorough exploration was made of the muscles of the injured side of the amputee's body. However, more than one inch of transverse motion of the scar results when the pectoralis muscle is contracted. The patient has excellent voluntary control of the motion of the skin.

A special transducer was developed to utilize this skin motion. This transducer utilizes a mov-

able magnet and stationary semiconductor elements which respond to changes in magnetic field strength. Approximately $\frac{3}{8}$ inch of motion of the small string, which can be seen in Fig. 8 emerging from the front of the shoulder of the prosthesis, controls the arm and terminal device. When the prosthesis is in use, the end of the string is passed through the large hole close to where it emerges from the prosthesis, and is attached to a button which is, in turn, attached to the skin with double-sided adhesive tape.

After using this prosthesis for approximately one month, the amputee reports that he uses it 50 to 75% of the time. He uses it for bimanual operations, and for tasks like carrying empty buckets, but does not use it when he can do his work without it, such as when he drives a tractor. The adhesive attachment to the skin has remained intact up to two weeks without coming loose, and appears to be satisfactory for this evaluation. The battery is providing up to nine hours of operation on a single charge. The amputee has expressed a need for a "reach" capability, and is of the opinion that the prosthesis would be much more useful if the shoulder were movable.

Conclusions

Preliminary results of the evaluation tests of experimental models of the externally-powered system on eight amputees indicated that this system has merit over body-powered systems in those cases where the following conditions exist:

1. Amputee unable or unwilling to furnish the high force level and large excursions required to operate the body-powered prosthesis. Specific examples are: (a) high level of amputation; (b) tenderness in the end of the stump; (c) restriction of joint motion, especially in the shoulders; and (d) shoulder muscle weakness.

2. Amputee needs a wider work envelope than is possible with a conventional body-powered prosthesis.

3. Amputee desires minimal harnessing for the prosthesis. As a tradeoff for this characteristic he must be willing to accommodate the additional weight/bulk of the electrical/mechanical subsystem of the powered system.

4. Training time must be minimized.

Compared to other externally powered systems, the Johns Hopkins prosthetic system offers advantages in that (a) it utilizes many standard,

available prosthetic components, (b) it provides a powered elbow and/or powered terminal device capability with a single motor and single EMG site, (c) it has demonstrated that proportional control is very "natural" and easy for the amputee to learn to use, (d) the system is versatile since powered components may be located either on the prosthesis or worn on the belt to optimize the system to individually suit the amputee's needs, and (e) donning is simplified by attachment of the sensor to the wall of the socket.

Item (d), versatility of component location, is a particularly significant parameter if the amputee is to be fitted in a manner consistent with his projected utilization of his prosthesis to maximize his rehabilitation capability. The equipment location criterion appears to be one of the critical factors in the ultimate acceptance or rejection of the powered system by amputees. The Johns Hopkins concept allows maximum flexibility for equipment location.

The above amputee evaluation was based on the original experimental design to get basic data on the practicality of the basic powered concept. This initial design was limited in scope and did not include significant effort on miniaturization of components. Further design refinements in packaging are planned to bring this design to a stage suitable for more extensive clinical testing and possible future availability to amputees.

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