EVALUATION OF THE NORTHWESTERN UNIVERSITY ATTITUDINALLY CONTROLLED ELECTRIC ELBOW[a]

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I. INTRODUCTION

This report presents the results of a performance assessment study of the Northwestern University attitudinally controlled electric elbow (hereafter referred to as the experimental prosthesis or electric elbow) conducted on one above-elbow amputee at the Biotechnology Laboratory, University of California at Los Angeles. The evaluation study employs the "semi-case study" approach as outlined and employed at UCLA (1, 2, 3).

The body of the report includes three major sections. In Section II the electric elbow is described. Section III describes subjects and fitting proce-

FIGURE 1. — Flexion control motion. Amputee demonstrates approximate angle of humeral flexion necessary to initiate and complete elbow flexion motion. (Photo courtesy of Northwestern University Prosthetic Research Center.)

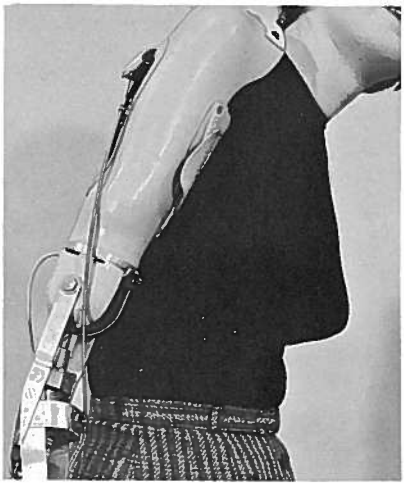

FIGURE 2. — Extension control motion. (Photo courtesy of Northwestern University Prosthetic Research Center.)

[a] Based on work performed under VA Contract V1005P–9779.

dures and summarizes performance tests conducted on one above-elbow amputee subject. Section IV presents conclusions and recommendations.

II. DESCRIPTION OF EXPERIMENTAL PROSTHESIS

The electric elbow is an externally powered elbow unit which is designed to be incorporated in a standard Hosmer above-elbow turntable (Part No. E401). The elbow is powered by a d.c. motor located within the forearm shell. A battery carried by the amputee provides energy for the motor. Movement of the elbow is effected by a power screw linked directly to the motor gearhead. Operation of the motor is regulated in an on-off fashion by a set of mercury switches mounted on the elbow unit. The switches are attitudinally controlled; that is, the state of each switch (open or closed) is determined by the position of the elbow unit with reference to gravity. With the amputee in the usual orientation, sitting or standing, the reference is the vertical plane. The amputee controls elbow position by movement of his humeral stump.[b] The assembled prosthesis is shown with forearm uncovered in Figures 1 and 2, and with forearm covered in Figures 3 and 4.

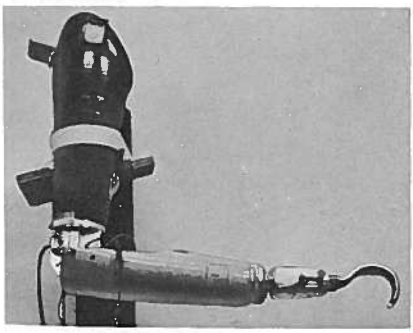

FIGURE 3. — Experimental prosthesis during bench testing.

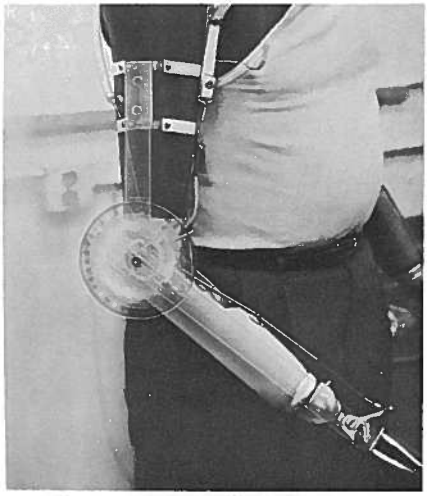

FIGURE 4. — Experimental prosthesis during performance testing with goniometer in place.

[b]The notation used by Santschi (*Manual of Upper Extremity Prosthetics, 2nd ed.*, 1958, p. 4) to describe forearm and humeral movements will be used in this paper. Specifically, the terms "extension" and "flexion" will be used to describe backward or forward movements, respectively, of either the forearm or the humerus. Humerus and forearm locations will also be labeled according to this notation. The range of humeral movement backward past the vertical plane will be considered degrees of extension, and the range forward from the vertical will be considered degrees of flexion.

A. Mechanical Characteristics and Components

The weight of the elbow unit and forearm shell, exclusive of terminal device or upper arm socket, is 2.2 lb. The length, taken from elbow axis to end of the forearm shell, is 10.8 in. The center of gravity varies from a position 4.9 in. distal to the elbow axis when fully extended to 3.0 in. when fully flexed. The weight of the entire prosthetic system, without harness straps, is 3.8 lb.

The elbow mechanism is composed of four interrelated units: 1. the forearm frame, 2. the motor with its gearhead, 3. the power screw, and 4. an elbow turntable (Fig. 5). The complete prosthetic system also includes an upper arm socket and harness, a forearm shell, and a terminal device with cable control system.

1. *Upper Arm.* Since the electric elbow is designed to be incorporated in a standard elbow turntable, it was possible to attach it directly to the upper-arm socket of a spare prosthesis provided by the amputee. This attachment necessitated modifications of the harnessing and cable system, but not of the socket.

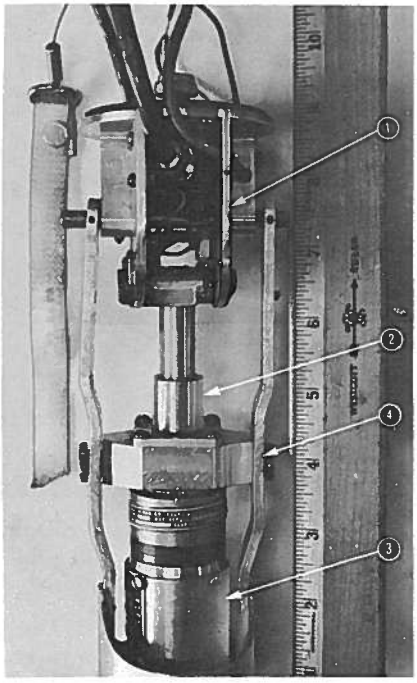

FIGURE 5. — Elbow mechanism. The four major parts of this system are the elbow turntable (1), the power screw system (2), the motor (3), and the forearm frame (4).

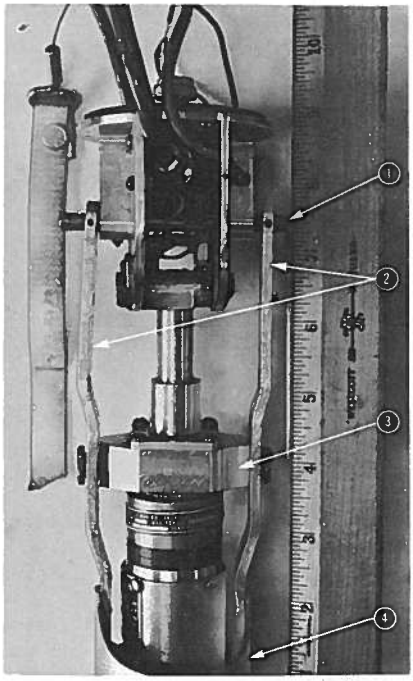

FIGURE 6. — Forearm frame. This unit includes a pin (1), side rails (2), motor attachment frame (3), and a cylindrical guard (4).

2. *Turntable Structure.* A circular plate with two parallel projecting legs provides the basic turntable structure. One end of the structure is designed to fit into the Hosmer E401 turntable at the distal end of the upper-arm socket. The legs of the "U" serve as attachment points for the forearm frame and for the power screw housing. The turntable structure also serves as a mounting platform for the various control switches.

3. *Forearm Frame.* This unit consists of two side rails, a pin, a cylindrical guard, and an attachment frame for the motor (Fig. 6). The two side rails are joined at the turntable structure by a hollow pin which runs perpendicularly through the rails. This pin is the axis of the elbow rotation. The cylindrical guard is connected to these rails; it protects the distal end of the motor system. Midway along the length of the side rails an octagonal motor attachment frame is mounted between the rails.

4. *Motor.* The motor is located in the distal end of the forearm frame; it incorporates a gearhead with a 32.8:1 reduction ratio. Its output shaft connects directly to a power screw. The motor can be operated in either of two directions. The amputee selects the direction of rotation of the motor, and thus the direction of forearm movement (flexion or extension), by activating one of two control circuits.

5. *Power Screw and Cam.* The power screw delivers torque from the motor unit through a shaft to the turntable structure. The proximal end of the shaft serves as a cam and pivots about the point of attachment to the elbow turntable structure. It is notched to contact two microswitches positioned on the turntable structure. The setting of these microswitches determines the limits of elbow rotation.

The driving force for the elbow, then, is provided by rotation of the power screw which acts against a fixed pivot point on the turntable structure; thus, the resultant motion is a rotation of the forearm frame around the pin connecting the proximal ends of the side rails. This system provides for motion in two directions and for automatic locking of the elbow when inactive.

6. *Forearm Shell.* The forearm shell, housing the forearm frame, motor, and power screw system, was provided by the developer of the arm. It is constructed of conventional plastic materials. The shell was trimmed at UCLA to increase access to the motor system. The distal end of the shell contains a mounting ring (wrist unit) which provides the means of attachment for the terminal device.

7. *Terminal Device.* The terminal device used with the experimental arm was the amputee's own hook and wrist flexion unit from his conventional prosthesis. These components are described in the Section entitled "Fitting." The experimental arm was designed to receive any standard prosthetic terminal device. In Figures 1, 2, and 4, a Dorrance hook is in place on the prosthesis, while in Figure 3 an APRL-Sierra hook has been attached.

B. Electrical Components

In addition to an energy source, the electrical system consists of the motor, six mercury switches (MS1–MS6), two limiting microswitches (LS1, LS2) and one alternating, pull-type switch (S1). These components are described in Table 1.

TABLE 1. — *Electrical Component Specifications*

Motor:	
Manufacturer	Barber-Colman Co. (Part No. BLYM 42900–50)
Type	DC, permanent magnet
Rated voltage	12 v.
Rated current	1.5 amp.
Rated speed	9800 r.p.m.
Rated power	0.100 hp.
Rated torque	0.165 in.-lb.
Gearhead:	
Manufacturer	Barber-Colman Co. (Part No. BLYH 923–6)
Gear ratio	32.8:1
Maximum torque output	2.5 in.-lb.
Limit switches:	
Manufacturer	Honeywell (Microswitch Part No. 1SX1–T)
Type	Lever, normally closed
Mercury switches:	
Manufacturer	Honeywell (Microswitch Part No. AS419A1)
Type	Single pole, single throw
Mounting clip:	
Manufacturer	Honeywell (Microswitch Part No. 1MR11–F)

The electrical system controlling the operation of the elbow is presented schematically in Figure 7. The location of all switches except S1 is shown in Figure 8. For flexion, switches MS3 and MS4 must be open and all others closed. For extension, switches MS1 and MS2 must be open with all others closed.

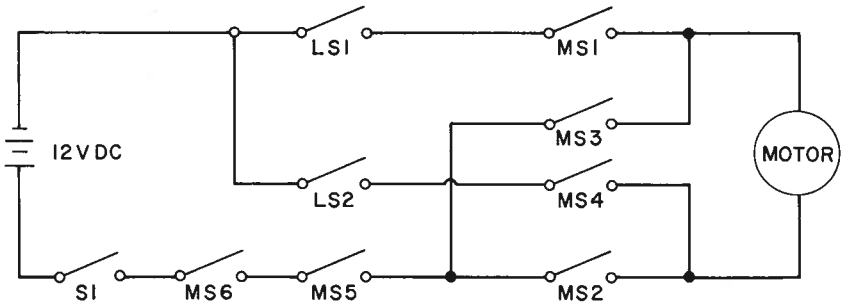

FIGURE 7.—Diagram of electrical system. S1 is master on-off switch, MS denotes mercury switch, and LS denotes limiting microswitch.

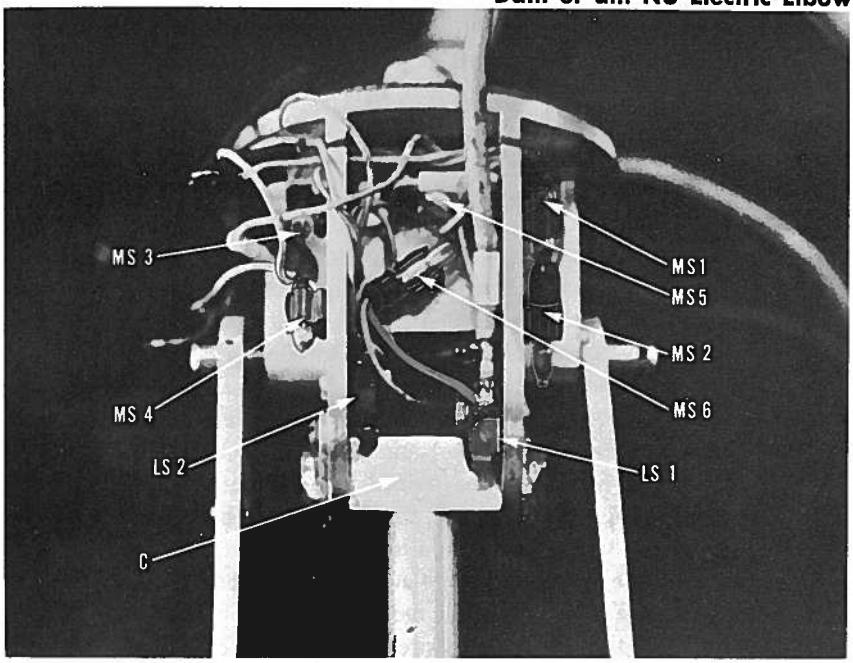

FIGURE 8.—Rear view of elbow unit. Photograph shows six mercury switches (MS1–MS6), limiting microswitches (LS1, LS2), and cam (C).

1. *Master On-Off Switch.* This switch (S1) is attached to the elbow lock billet, a standard part of the conventional above-elbow harness used to control elbow locking in conventional above-elbow prosthetic systems. The amputee can activate and shut off the electrical system with successive pulls on this switch. His control motion is the same as is used with the conventional arm (depression of the shoulder and slight extension of the humerus backwards from the vertical plane). This switch is in series with the battery and motor; thus, no operation is possible except when the switch is closed.

2. *Forearm Flexion Switches.* Flexion is controlled by the amputee in an on-off manner only. No regulation of speed of movement is possible. The amputee initiates flexion by flexing his stump sufficiently to close switches MS1 and MS2 (Fig. 1). These switches are mounted on the turntable structure in a plane parallel to that of flexion and extension. Flexion can be stopped by returning the humerus to the normal relaxed position (0 deg. flexion), thus opening MS1 and MS2. Flexion of the elbow unit is also stopped automatically when a limiting microswitch (LS1) is opened by contact with the cam (proximal end) of the shaft from the motor unit. The position of LS1 defines the upper limit of flexion.

3. *Forearm Extension Switches.* Extension of the forearm occurs when MS3 and MS4 are closed by extension of the humerus past 0 deg. flexion

(see Fig. 2). Extension is halted either when LS2 is contacted and opened by the cam or when the humerus is returned to the normal position (0 deg. flexion).

4. *Humeral Abduction and Adduction Switches.* Switches MS5 and MS6 function as a safety feature. They are mounted at the rear of the elbow in a plane parallel to the coronal body plane (when the humerus is at 0 deg. flexion). Normally closed, these switches are preset so that one or the other will open when the amputee abducts or adducts the humerus beyond a specified deviation from the vertical plane. Without these switches it might be possible to position the elbow manually so that MS1 through MS4 would all be closed simultaneously, thus shorting the circuit. It is not likely that the amputee would abduct to the position necessary to achieve closing of all 4 switches ($>$ 90 deg.), since the elbow unit is quite heavy. However, the elbow unit could be placed in a dangerous position if the amputee were to fall, bend sideways, lie down with the prosthesis on, or remove the prosthesis and place it on a table without turning off the master switch. Thus, the main value of these switches is in eliminating the possibility of shorting due to *abnormal* or *inadvertent* positioning of the elbow.

A second use of the abduction switch is provision of a means to halt elbow rotation without having to return the humerus to 0 deg. flexion or to turn off the master switch. Ideally, this would allow the amputee to maintain the prosthesis in the area of task performance in a passive state, but with the option of activating the elbow quickly again. This would also permit terminal device activity without fear of accidental activation of the elbow.

In summary, the operation of the elbow can be halted by one of four methods: 1. turning off the master switch, 2. abducting or adducting the humerus, 3. returning the humerus to the vertical plane, or 4. allowing elbow motion to continue until a limiting microswitch is contacted.

5. *Battery.* The elbow can be operated by a portable, rechargeable nickel-cadmium battery. The one used with the prosthesis was manufactured by Gould-National. It weighs 9.3 oz., is cylindrical in shape (3.9 in. height, 1.4 in. diameter), and contains 10 cells. The battery provides 12 v. with a capacity of 0.50 amp.-hr.

During the evaluation, the arm was not operated from the portable source. A 12 v. power supply was used for convenience and to insure a constant, regulated operating voltage. It was found during testing that the maximum life of a charge on the portable battery was 2 hours. Recharging this battery takes considerably longer than 2 hours; thus, a large number of spares plus the ability to charge many of these simultaneously was required. The developers of the arm plan to provide a more adequate power source with production models. For this reason power usage and battery life were not considered in the present evaluation.

C. Engineering Considerations

1. *Maximum Force.* The maximum force of the motor in initiating forearm flexion was measured at two starting locations, 0 deg. and 15 deg. forearm flexion. Weights were hung from the terminal device at a point 15 in. from the axis of elbow rotation. It was found that the maximum weight that could be lifted was 3.5 lb. at 0 deg. and 4.25 lb. at 15 deg. These weights represent an external loading condition in addition to the normal weight of terminal device, forearm shell, and components of the elbow unit. The experimental arm mounted on a test stand is shown in Figure 3.

2. *Operational Failures.* After about 10 hours of use the extension limit switch (LS1) was broken. Failure occurred because successive extension cycles gradually bent the leaf spring (contact lever) of the microswitch back toward the body of the switch until the cam finally moved directly onto the switch during extension. Ideally, the motion of the elbow was designed to halt on extension at the vertical plane position. The microswitch was positioned so that it would make contact with the cam and open, halting movement at this position. However, the momentum of the elbow unit extended the elbow for approximately one additional degree after the extension circuit was broken.

This difficulty was corrected by rotating the position of the switch so that contact between the switch body and cam was impossible. The shape of the leaf spring was changed to compensate for the greater distance from cam to switch and to provide additional resistance to the distorting force of the cam. A more sophisticated improvement was advanced by the designer. The improvement replaced the simple leaf spring with a more durable roller lever (Minneapolis-Honeywell, J X 25). This component was later provided by the designer and was found to serve satisfactorily.

Another problem encountered in the initial phases of testing was that the worm gear became tightly screwed into its end nut when the elbow was allowed to extend past the vertical plane. This condition resulted from failure of the limit switch to halt elbow motion at the proper point of extension. The reverse motion, flexion, was then impossible, since these parts were locked together as a nut and bolt would be. It was necessary to dismantle the unit and unscrew the nut manually to correct this condition. When the modification of the limit switch mentioned above was made, this problem was eliminated. As an additional safeguard, a spring was attached between the elbow turntable and the forearm body to prevent extension past the vertical plane.

Another trouble point was the mounting of the mercury switches. Each of these switches is held in a spring clip attached to the arm by a single rivet. The rivets are loose enough to permit manual adjustment of the orientation of the switches. However, this lack of rigid attachment makes

it possible to shift them out of position inadvertently during use. A practical solution adopted during testing was to tape the clips and switches to the elbow once they were properly aligned. Future versions of this prosthesis should include a more stable means of mounting these switches, while permitting adjustment of the switch position.

A problem discovered during reaction time tests was oscillation or "contact bounce" of the mercury switches. Recordings of current through the switches revealed that initiation of movement of the elbow was delayed

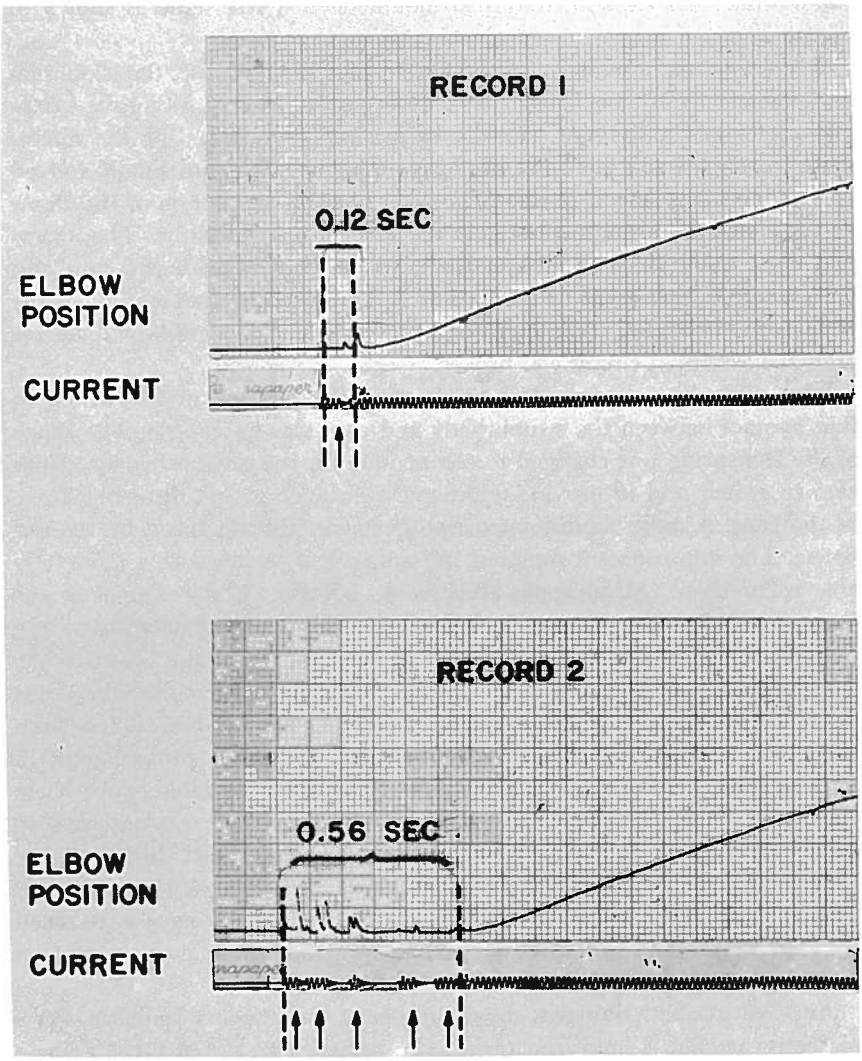

FIGURE 9.—Delays in initiating movement. Delays experienced ranged from 0.12 sec. with 1 bounce in switch (record 1) to 0.56 sec. with 5 bounces (record 2).

from 0.12 to 0.56 second because of oscillation of the fluid within the switches. The mean delay for 10 extension trials activated in the normal manner by the amputee was found to be 0.29 second. This delay occurs in addition to the normal lag associated with starting the motor and overcoming the inertia of the prosthesis. It can be reduced significantly if the amputee makes a slow, smooth control movement; however, no gain is achieved, since one lag (slower control movement) is merely substituted for another (switch delay). Sample recordings illustrating this delay are shown in Figure 9.

A final point to be mentioned is that considerable wear developed at the joints of the elbow as testing progressed. Since this arm was a prototype, this feature cannot be severely criticized. However, production models should employ ball bearings or some other means to decrease friction and wear, and the quality of metals used should be improved.

III. PERFORMANCE ASSESSMENT

A. Amputee Subjects

Originally, the evaluation was planned for a sample of three amputee subjects. However, one of these subjects became unavailable, due to other interests, soon after testing started. A second amputee was found to require extensive training with his conventional prosthesis as well as with the experimental arm. Since his difficulty was aggravated by poor health, evaluation of the prosthesis with this amputee was discontinued. Thus, the evaluation is based primarily on the performance of one amputee subject. All quantitative data reported here were collected using this amputee as the pilot-wearer. In a few cases, qualitative statements represent the combined opinions of more than one amputee.

The characteristics of the amputee pilot-wearer are presented in Table 2. This amputee is an experienced wearer of conventional above-elbow pros-

TABLE 2. — *Description of Subject*

Sex	Male
Age	45
Height	6′ 1″
Weight	175 lb.
Occupation	Real estate salesman
Health	Good
Attitude	Good
Amputation	Bilateral, standard AE
Cause of amputation	Industrial accident
Age at amputation	35
Present prosthesis	Conventional body-powered AE configuration with Hosmer internal elbow, Dorrance 5X hook, figure–8 harness
Use of present prosthesis	Constant use in daily life

thetic configurations and has participated in a number of training and demonstration sessions. In addition to being of good health, he is highly motivated.

B. Fitting

1. *Conventional Prosthesis.* The prosthesis used for comparison with the experimental elbow was the conventional body-powered above-elbow configuration worn by the amputee in his daily routine. It was fitted and serviced by a local Los Angeles prosthetic firm. The prosthesis consisted of a Hosmer internal elbow, a Dorrance 5X terminal device, and a Sierra Model B wrist flexion unit. The upper arm socket and lower arm shell were constructed of conventional plastic materials. A figure-8 harness provided the usual dual control arrangement for elbow flexion, elbow locking, and terminal device operation.

2. *Experimental Prosthesis.* The upper arm and forearm shells were fabricated by Adroit Prosthetics (Los Angeles) and Northwestern University Prosthetic Research Center, respectively. They were constructed from conventional plastic materials. The harnessing attachment was a figure-8 harness. Since the elbow is attitudinally controlled, the forearm lift loop usually found in above-elbow prostheses was unnecessary. The terminal device was body-powered and operated by the Bowden control cable arrangement typically used in below-elbow prosthetic systems.

A critical factor in fitting the electric elbow is adjustment of the mercury switches and the master switch. The amputee wishes to initiate powered motion with a minimum of control motion and physical exertion. In addition, he wants to use the prosthesis in passive activities without fear of accidentally initiating powered movement of the prosthesis. Inadvertent powered movement can be avoided by turning the master switch off, but this can prove difficult in the middle of an activity. Sometimes the amputee will forget to turn the switch off, and a margin of safety should be provided. Finally, even if he remembers to turn the switch off, in some cases it will be tedious and will slow his operation down considerably if he must constantly turn the unit on and off.

In fitting, a compromise was sought between ease of initiating movements and the chance of activating the elbow inadvertently. The mercury switches which activate the two control circuits are each mounted on the turntable structure with a spring clip. Each clip is attached to the structure by a single rivet which allows adjustment of the attitude of the switch. The particular angle of humeral flexion or extension required to close the switch can be selected by rotating the switch in one direction or the other.

It was found that setting MS1 and MS2 so that 15–20 deg. of humeral flexion were required for activation greatly reduced inadvertent activation of the elbow during terminal device operation or during other passive activ-

ity. Settings of 25 deg. or more reduced this possibility further but made the task excessively tiring for the amputee.

The optimum setting for MS3 and MS4 was found to be that which required about 10 deg. of humeral extension. This allowed adjustment of the elbow lock billet so that the master switch could be turned on and off with a motion of 20 deg. extension. With this adjustment these two control motions were easily distinguishable while a small amount ($<$ 10 deg.) of passive humeral extension was still possible.

MS5 and MS6 were adjusted to open at 40 deg. humeral abduction. This setting was sufficient to avoid shorting of the battery by simultaneous closure of flexion and extension circuits without restricting passive humeral abduction movements.

Flexion and extension control motions are illustrated in Figures 1 and 2.

C. Performance Tests

1. *Range of Motion.* The range of elbow motion was measured as the total angular rotation of the forearm for each prosthesis. The range of terminal device opening was measured at each of six positions of the prosthesis: 1. at the mouth, 2. at the breast pocket, 3. at 90 deg. flexion of the elbow, 4. with the entire arm straight out from the chest, 5. at the perineum, and 6. at the anus. These locations were taken, in part, from checkout specifications for upper-extremity prostheses (4).

Elbow rotation was measured with a plastic goniometer attached to the humeral socket with straps (Fig. 4). The goniometer was mounted so that its axis of rotation coincided with that of the elbow and so that its zero position coincided with the vertical plane. The goniometer was read to the nearest degree.

Terminal device opening was measured with a caliper. The readings were taken between the inner edges of the terminal device at the maximum possible opening in each condition. These readings were recorded to the nearest 0.05 in.

The total range of motion of the conventional elbow was 125 deg., from 9 deg. at full extension to 134 deg. at full flexion. With the experimental prosthesis the elbow range was 124 deg. from 0 deg. at full extension to 124 deg. at full flexion.

Table 3 presents the amount of terminal device opening achieved at each elbow location with the two prostheses. The results are presented as percentages of the maximum opening possible, which varies with the terminal device used.

The ability to operate the terminal device at the positions tested was found to be nearly identical for the two prostheses. In all cases the percent of opening achieved was equal to or greater than that recommended as a minimum (4).

TABLE 3. — *Terminal Device Opening at Six Elbow Positions*

Location	Prosthesis		Minimum recommended[a]
	Experimental	Conventional	
Mouth	100	100	50
Breast pocket	100	100	50
90° flexion	100	100	100
Arm perpendicular to chest	100	100	50
Perineum	100	100	50
Anus	48	100	50

Note — All values are in percents.
[a]Santschi, 1958.

The range of elbow motion was about the same for both prostheses. The conventional model was superior in that the upper limit of rotation permitted activity at the face while this could only be achieved with the experimental prosthesis by additional stump movement.

2. *Motion Completion.* The objective of this part of the evaluation was measurement of the maximum speed at which the amputee could flex and extend the forearm of his prosthesis.

The amputee was instructed to position his elbow at one or the other extremes of the range of elbow motion. Upon a verbal signal he made the appropriate control motion necessary to flex or extend the forearm to the opposite end of the range of motion. Ten trials were performed for each direction of movement with each prosthesis. The average speed of motion for each trial was calculated simply as the ratio of total movement in degrees to time elapsed from beginning to end of actual movement.

One channel of an Offner (Type RS) oscillograph was used to record movement. The channel was part of a circuit arranged to provide a voltage analog to the angle of elbow rotation. A power supply and a rotary precision potentiometer formed the remainder of the circuit. The potentiometer was attached to the elbow of the prosthesis by a special mounting bracket. The oscillograph was calibrated so that records could be read directly in degrees of elbow rotation. Elbow position was read to the nearest 1.5 deg. and time was read to 0.02 second.

The average speed for each prosthesis is presented in Table 4. The speeds of the two prostheses were tested for significant differences in flexion and extension by separate analyses of variance. A 5 percent significance level was adopted for these tests and for all others in this report.

Tests for flexion ($F = 195.94$, $df = 1, 18$) and extension ($F = 184.78$, $df = 1, 18$) both showed the conventional prostheses to be significantly faster than the experimental prosthesis.

TABLE 4. — *Average Speed of Flexion and Extension*

Function	Experimental prosthesis		Conventional prosthesis	
	Mean	SD	Mean	SD
Flexion	42.9	2.4	198.4	31.8
Extension	50.3	0.8	202.4	11.2

Note — All values are degrees/second.

Visual inspection of data records confirmed the impression that the movement of both prostheses was smooth and not markedly nonlinear. On this criterion both prostheses were satisfactory; however, in regard to speed of operation, the conventional prosthesis is to be preferred, since movement over the total range required less than one second as opposed to 2.5 to 3 sec. with the experimental prosthesis.

3. *Reaction Time.* The objective was measurement of the speed with which the amputee could initiate the movements of his prosthesis.

In these tests the amputee stood facing a display board which was positioned at eye level. The board was normally blank, but one of three command words (flex, extend, or open) could be illuminated by the experimenter using a concealed, muffled switch. The amputee's task was to initiate the proper motion as soon as possible after the display was illuminated. Reaction time was measured as the time elapsed from the illumination of the command display to the start of movement of the prosthesis.

Two types of reaction time scores were taken: simple and complex. In the simple case the amputee was told in advance the movement to be performed, while in the complex case he was unaware of the proper motion until the trial actually began.

A total of 51 trials was performed with each prosthesis in the simple reaction time situation and in the complex reaction time situation. The order of presentation of the three functions was semi-randomized. On complex reaction time trials, in addition to reaction time scores, the number of errors (incorrect movements) was also noted. From reaction time scores a derivative measure, decision time, was computed. This measure represented the difference between mean simple and mean complex reaction time scores for each function of each prosthesis. Decision time and number of errors represent two gross indices of the increased complexity of operation of the control system when the prosthetic function to be performed is not known beforehand.

In the analyses which follow, N was not always 51 since recording errors required discarding of a few trials.

Movements of the terminal device and elbow unit were monitored by circuits analogous to those used for motion completion measurements. Two

channels of an Offner oscillograph were used to record these movements. The event-marker channel of the oscillograph was wired so as to be activated simultaneously with the illumination of the display board, thus providing the reference point for the start of a trial. Reaction time scores were read to the nearest 0.02 second.

Results of the reaction time tests are presented in Tables 5, 6, and 7. Simple analyses of variance revealed that flexion ($F = 126.62$, $df = 1, 30$) and extension ($F = 37.02$, $df = 1, 30$) were initiated significantly faster with the conventional arm in the simple reaction time situation. The same differences were found in the complex reaction time situation (flexion, $F = 64.49$, $df = 1, 32$; extension, $F = 63.02$, $df = 1, 32$). Speed of terminal device opening did not differ significantly between the two prostheses.

TABLE 5. — *Simple Reaction Times*

Function	Experimental prosthesis		Conventional prosthesis		Mean difference
	Mean	SD	Mean	SD	
Flex	1.11	0.14	0.37	0.11	*0.74
Extend	0.80	0.30	0.34	0.05	*0.46
Open	0.30	0.06	0.27	0.04	0.03

Note — All values are in seconds.
*Difference significant at 0.05 level.

TABLE 6. — *Complex Reaction Times*

Function	Experimental prosthesis		Conventional prosthesis		Mean difference
	Mean	SD	Mean	SD	
Flex	1.07	0.07	0.72	0.32	*0.35
Extend	1.08	0.26	0.58	0.07	*0.50
Open	0.66	0.14	0.58	0.15	0.08

Note — All values are in seconds.
*Difference significant at 0.05 level.

Separate analyses of variance also revealed that complex reaction time scores were significantly larger than simple scores for each function of the conventional prosthesis (flexion, $F = 147.06$, $df = 1, 32$; extension, $F = 16.73$, $df = 1, 32$; opening, $F = 64.06$, $df = 1, 32$). For the experimental prosthesis extension ($F = 4.55$, $df = 1, 32$) and terminal device opening ($F = 87.20$, $df = 1, 32$) were significantly slower in the complex reaction time case than the simple case, but flexion reaction time was about the same in both cases. This latter result is due to the fact that the mean simple reaction time for flexion with the experimental prostheses was unusually high. Inspection of the data provided no basis for discarding this result.

Excluding the case of flexion with the experimental model, decision times were on the order of 0.3 second; that is, for both prostheses, complex reaction time scores were about 0.3 second longer than simple reaction time for each function. Few errors occurred in selection of the proper prosthetic function on complex reaction time trials. No errors were made with the conventional prosthesis while four errors out of 48 trials were made with the experimental prosthesis.

The results of the reaction time tests do not provide a clear basis for preferring one prosthesis over the other. The number of errors committed with the experimental model was low. Decision times, inferred from the complex reaction time situation, were fairly low for both prostheses. They were in agreement with results from a previous study (3) employing a conventional above-elbow prosthetic configuration for comparison purposes. Thus, on the basis of complexity of operation, both prostheses appear to be satisfactory.

TABLE 7. — *Decision Times*

Function	Prosthesis	
	Experimental	Conventional
Flex	—0.04	0.35
Extend	0.28	0.24
Open	0.36	0.31

Note — All values are in seconds. Decision time equals mean complex reaction time minus mean simple reaction time for each function.

No differences were found in reaction times for terminal device operation. However, the externally powered flexion and extension motions of the experimental prosthesis were on the average 0.5 second slower to be initiated than the corresponding body-powered motions of the conventional prosthesis in the simple and complex reaction time situations. Additional familiarity with the experimental prosthesis would allow the amputee to reduce his reaction time somewhat, but part of the increased latency is undoubtedly due to lags in the mechanical system, such as in the mercury switches. These are therefore irreducible by the amputee, but may be minimized by engineering design. It has been noted above that the average lag through the mercury switch system was about 0.3 second, which would account for much of the observed difference in reaction time between the prostheses for flexion and extension movements.

4. *Precision of Motion.* This evaluation was conducted to determine the accuracy with which the amputee could control the motions of the elbow unit.

The amputee's task was to match the position of the forearm with that of a visual indicator adjusted to some angle along the range of elbow rotation. The amputee started a trial from one of two locations — the extreme upper or lower position of the range of motion. Those trials starting at the top of the range were extension trials, and the others were flexion trials. With both prostheses, each trial started and ended with the prosthesis locked. Since the conventional prosthesis locks only at certain fixed points along its range, while the experimental prosthesis locks anywhere over its range, test positions were chosen to coincide with the fixed locking positions of the conventional prosthesis. These positions were 34, 71, 96, 108 and 120 deg. of flexion as measured from the vertical plane. All five angles were used in testing flexion, but certain angles were omitted on extension trials.

Seventeen trials were performed at each test position for each starting direction. The order of presentation of test angles and starting directions was randomized. After every trial the amputee was instructed to return the prosthesis to the starting position. Then the visual indicator was returned to zero before being set at the next test angle. All trials were completed with the conventional prosthesis before proceeding to the experimental prosthesis.

Both speed and errors were taken as indices of precision of control. The amputee was instructed to be "as accurate as possible while working at a normal pace." Three measures were taken on every trial. These were: 1. the absolute deviation, in degrees, of the first attempt to reach the test position, 2. the number of extra movements needed to reach the test position, and 3. the time, in seconds, from start of movement until the final position was reached.

Errors in recording required discarding of a few trials. Thus N sometimes was less than 17 in the analyses which follow.

The elbow goniometer used in measuring the elbow range of motion was used as a visual indicator for these tests (Fig. 4). It was strapped to the outer side of the upper arm shell, and aligned so that its axis of rotation was the same as that of the elbow unit, and calibrated so that its movable arm could be set precisely at any of the five locking positions of the conventional prosthesis. In precision of motion trials the amputee attempted to align a point on the forearm of the prosthesis with the movable arm of the goniometer.

Each trial was recorded using the instrumentation described earlier in Section III.C. The number of readjustments was counted from the data records. Trial duration was measured to the nearest 0.02 second, and elbow position was read to the nearest 1.5 deg.

Tables 8 and 9 summarize the results of these tests. Analyses of variance were performed between the means for the two prostheses at each test position for each of the three measures of precision.

TABLE 8. — *Precision of Motion, Flexion Trials*

Score	34°		71°		96°		108°		120°	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD
First error[a]:										
Experimental	10.6	8.2	12.6	6.1	5.5	3.4	5.0	6.8	17.9	33.9
Conventional	6.0	3.2	9.2	14.5	18.4	29.1	13.9	22.7	26.5	40.0
Readjustments[b]:										
Experimental	0.6	0.7	0.5	0.9	0.4	0.7	0.2	0.6	0.3	0.7
Conventional	2.8	1.1	0.9	0.6	1.3	1.3	2.4	2.0	2.6	1.0
Time[c]:										
Experimental	2.0	1.1	2.5	1.5	3.1	1.3	3.5	2.0	3.4	0.8
Conventional	2.4	0.7	2.2	0.5	2.8	1.4	4.1	2.1	4.0	1.5

[a]Values are in degrees.
[b]Values are in number of extra attempts to reach test angle.
[c]Values are in seconds.

TABLE 9. — *Precision of Motion, Extension Trials*

Score	71°		96°		108°		120°	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD
First error[a]:								
Experimental	4.1	2.7	6.1	4.5	7.0	3.0	—	—
Conventional	—	—	4.7	7.2	1.4	3.2	3.6	4.2
Readjustments[b]:								
Experimental	0.1	0.2	0.6	1.0	0.6	0.9	—	—
Conventional	—	—	0.4	0.8	0.3	0.6	0.9	0.8
Time[c]:								
Experimental	1.3	0.3	2.3	2.8	2.6	3.1	—	—
Conventional	—	—	1.8	0.7	1.3	0.8	1.6	1.8

[a]Values are in degrees.
[b]Values are in number of extra attempts to reach test angle.
[c]Values are in seconds.

With one exception (error on first attempt at 108 deg. test angle) no significant differences between prostheses were found on extension trials for any of the measures taken. For flexion trials neither error at first attempt nor time to complete trial yielded significant differences at the five test angles, while the number of readjustments was significantly lower with the experimental prosthesis at every angle tested except 71 deg.

In general, performance was satisfactory with both prostheses on all three measures of precision. A negative aspect of performance with the experimental prosthesis was revealed on extension trials at the 120 deg. test position. Starting at full flexion (124 deg.), the amputee was never able to reach this position and stop in one movement. He consistently overshot the test position and finally refused to attempt it at all. The same problem occurred to a lesser degree at 108 deg. This indicates that it is not possible to make small corrective movements with the experimental prosthesis. The mean number of adjustments taken on precision trials with the experimental prosthesis, however, does not reflect this difficulty. This contradiction is explained by the fact that the amputee's criterion for completing a trial differed between the two prostheses. In the case of the conventional prosthesis, there was no question whether the proper angle had been reached, since test angles coincided with fixed elbow locking positions. With the experimental prosthesis, which has no fixed locking position, the amputee decided when to stop the trial based on his own standard of an error. He usually settled for a "reasonably close" (± 6–8 deg.) final position that varied from trial to trial. His strategy was to do as well as possible on the first attempt and avoid readjustments unless absolutely necessary.

5. *Everyday Tasks.* Six tasks, representative of the routine daily activities of an amputee, were chosen for this phase of the evaluation. These tasks were: 1. answering a telephone, 2. opening and closing a door, 3. unzipping and zipping a trouser fly, 4. drinking soup, 5. transporting a briefcase, and 6. hanging up a coat hanger. These and other standardized tasks have previously been described in detail (3).

Each task was practiced and performed with the conventional prosthesis before proceeding to the experimental prosthesis. The amputee practiced the task until both he and the observer agreed that his performance was adequate. Then a series of timed trials was begun. On each trial the observer recorded both the length of the trial and a subjective appraisal of the smoothness of performance, noting such things as errors (e.g., dropped objects, etc.), posture, body movements, and effort. When the amputee was able to perform five successive trials within a range of variation of 0.5 second, testing was stopped for that task.

Props used in these tasks included a telephone, a paper pad and pencil, a model door with standard knob, a bowl of water and soup spoon, a

clothes hanger, and an empty briefcase. Trials were timed with a stopwatch read to 0.1 second.

Table 10 summarizes the results of these tests. The task times presented represent the average of the last ten trials before the 0.5 second performance criterion was reached. Observation of performance on these tasks revealed that, in general, the conventional prosthesis was equal or superior to the experimental prosthesis in smoothness of performance. On the telephone task with the experimental prosthesis, the amputee was able to perform this sequence only when standing. It required excess body motion and he complained of back soreness. No difficulty occurred with the conventional prosthesis.

TABLE 10. — *Task Completion Times*

Task	Experimental prosthesis		Conventional prosthesis		Difference
	Mean	SD	Mean	SD	
Telephone	24.2	3.3	30.9	1.6	[a]—6.7
Door	10.5	2.3	10.9	2.1	—0.4
Zipper	33.4	6.7	26.7	9.0	[b]6.7
Soup	—	—	6.2	0.5	—
Briefcase	8.3	0.7	8.6	1.2	0.3
Hanger	7.4	0.3	8.2	1.0	[a]—0.8

Note — All values are in seconds.

[a]Mean for experimental prosthesis significantly faster than conventional at 0.05 level.

[b]Mean for conventional prosthesis significantly faster than experimental at 0.05 level.

The soup task could not be performed with the experimental prosthesis since liquid in the spoon was spilled during flexion. This task was also difficult to perform with the conventional prosthesis, but the amputee was able to complete it satisfactorily by bracing the forearm against the table edge after filling his spoon and bending down to meet the spoon.

The briefcase task could be performed with the experimental prosthesis only with the elbow preflexed and locked. It was not possible to raise the briefcase to table height using the elbow motor since the unloaded briefcase (5 lb.) exceeded the stalling torque of the motor. Performance with the conventional prosthesis was smooth and satisfactory.

Neither prosthesis was satisfactory on the zipper task. With the conventional model the amputee frequently lost his grasp on the tab and had to regrasp. With the experimental model he frequently activated the master switch while unzipping, causing the prosthesis to extend due to the angle

of the elbow. (Normally he performed this task with master switch off using "body english" to lower and raise the zipper tab.)

On the door and hanger tasks performance was smooth with both prostheses. Elbow motion was not required, however, for the door task outside of prepositioning for grasp, since the main feature of the task, rotating the doorknob, was accomplished using "body english."

The performance on everyday tasks with the experimental prosthesis was unsatisfactory in some cases. In others, although the task could be performed adequately, the experimental prosthesis offered no advantage over the conventional prosthesis. Task performance times, where comparable, did not differ greatly between the two models.

IV. SUMMARY

A. Review of Subject Performance

Performance with the experimental prosthesis was adequate with regard to range of motion, speed of motion, reaction time, and precision of motion. However, in no case was it superior to the conventional prosthesis. Furthermore, certain everyday tasks could not be performed at all with the experimental prosthesis. In general, the amputee preferred the conventional model because of its speed, flexibility, and naturalness of motion.

Probably the most serious limitation imposed on performance by the experimental prosthesis is the inability to make coordinated humeral and elbow movements in many cases. Since powered elbow motion is controlled by the position of the humerus, certain restrictions on the location of the humerus exist when powered motion is desired, and others exist when powered motion is to be avoided. Specifically, to have powered motion the humerus must be flexed or extended past a preset range, the elbow must not be abducted past a preset angle, and the master switch must be on. To avoid the chance of elbow rotation, at least one of these conditions must be violated. These restrictions can lead to incompatibilities in task performance. For instance, the movement (humeral flexion or extension) necessary to initiate a change in elbow rotation will have the effect of moving the terminal device away from the task which the amputee is trying to accomplish. Thus, the amputee loses his reference point and must guess at the amount of change in elbow position needed, rather than being able to compare terminal device position directly to the desired end-point. A similar problem occurs when the amputee must move his humerus toward 0 deg. flexion to halt movement. Serious positioning restrictions or complications in operation and task performance exist with the experimental prosthesis as compared to the conventional system.

B. Functional Cosmesis and Wearer Acceptance

In general, all three amputee-wearers considered cosmetic aspects of the electric elbow to be satisfactory. Elbow movement was smooth, although slower and more "mechanical" than the conventional prosthesis. Motor noise was considered annoying at first, but the subjects were able to adapt to it. A minor complaint was voiced by one amputee with regard to "stiffness" of the elbow. He felt that the fact that the elbow is not free-swinging would call attention to the wearer in public (as well as hamper those activities performed with the elbow unlocked, for example, steering a car).

Two amputees reported that they had to work considerably harder with the experimental prosthesis to achieve forearm flexion than with the conventional. After test sessions, they felt extremely tired on the side on which the electric elbow was fitted. The extra weight of the experimental prosthesis was partially responsible (3.8 lb. for the experimental prosthesis versus 3.0 lb. for the conventional prosthesis), although the need to make readjustments (extra movements) with it may have contributed. With regard to weight it should be noted that considerable adjustment was necessary to the harness systems of all amputees and, in one case, addition of another supporting strap (cross-back strap) was necessary.

C. Operational Safety

No problems were encountered with regard to operator safety.

D. Mechanical Reliability

With the exception of the minor problems noted earlier, the elbow performed adequately during testing. Since the evaluation was performed with a prototype model only, two unacceptable features of the present model have been overlooked. These are: (a) the high power requirement of the motor coupled with the lack of an adequate, long-term, lightweight power supply, and (b) poor quality of workmanship and materials in the arm. With regard to the latter, one clear need is the inclusion of ball bearing supports or other friction-reducing devices at the various axes of rotation in future models.

E. Conclusions

This evaluation is based primarily on the performance of one amputee pilot-wearer, with supporting observations from two other amputees who wore the device for a limited time. Based on our performance tests, the experimental prosthesis (electric elbow) in its present form seems to be an adequate device for above-elbow amputees. However, the electric elbow does not offer any advantage over the conventional body-powered configurations available. Some of the most serious drawbacks of the electric elbow

are its weight, the restrictions of humeral motion, and the difficulty of making fine (small) movements. Also, the arm can only be used when the amputee is in the normal vertical orientation (standing or sitting upright).

Externally powered systems will offer maximum benefit to the most severely disabled amputees, that is to shoulder-disarticulation and forequarter amputees. Unfortunately, the present prosthesis utilizes a control system which has no application for these cases. Of those cases (above-elbow) to which the arm properly applies, then, only short above-elbow amputees may realize some functional regain not possible to them with conventional systems. However, the weight of the elbow will be more of a problem for these amputees than for other above-elbow amputees. Thus, further development of this prosthesis, should include consideration of possible adaptations of the present control system for use by the more severely handicapped amputee population. This would involve radical change of the present control system, which is based on humeral motion.

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