SENSORY MOTOR CONTROL SYSTEM FOR AN EXTERNALLY POWERED ARTIFICIAL ARM [a]

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INTRODUCTION

The problem of developing a control loop for an externally powered prosthesis can be considered in two steps: 1. development of control signal pickup techniques, and 2. the design of an information loop between the control signal source and the output device.

In studies performed at UCLA the first step involved exploration of the control potential of mechanical deformations accompanying the contraction of existing muscle sites. The second step involved the development of an electronic, solid-state system which processed the control signals generated by upper body muscle sites for the control of a three-dimensional arm simulator.

Succeeding parts of this report will summarize in order: 1. Development of suitable muscle transducers and attachment methods, 2. efforts directed toward selection of muscle sites which would provide reliable and isolated outputs, 3. development of an electronic logic network for conversion of coded muscle transducer outputs into control signals to drive a prosthesis, and 4. the results of an extended training study designed to assess the adequacy of a selected set of muscle sites in control of a powered arm simulator.

TRANSDUCER DEVELOPMENT

Transducers are required to sense external displacement at a particular muscle site with respect to a stationary point on the surrounding skin. A number of different approaches were tried in our research at UCLA in the attempt to find a practical solution to the problem. Basically three types of transducers, which can be classified according to their electro-mechanical

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properties as 1. carbon, 2. photoelectric, or 3. strain gage transducers, were developed and evaluated. The development criteria were the following: low mass, low volume, reasonable sensitivity (gage factor), reasonable power gain, reproducible output, durability, and low cost.

The strain-gage-type transducer was found to meet the design criteria best. It was found to have a reproducible output, high durability, and reasonable cost of manufacture (1).

Three different types of strain gage transducers were developed for three separate body muscle sites: the abdominal wall, the chest, and the pectoralis muscle. Although a strain gage was the basic electrical element, the construction of each transducer varied considerably depending on the intended muscle site.

The chest transducer was activated by expansion of the chest cavity. The transducer consisted of an inverted U-shaped frame connected to two Dacron straps. An SR–4 strain gage was cemented to the upper side of the frame. The transducer was attached to the subject by straps which encircled the body and were held together by a buckle. The force required to operate the transducer was about 1.7 lb.

The abdominal wall transducer was activated by movements of the abdominal wall in a direction away from the body and perpendicular to the frontal body plane. Muscular contraction acted against a circular plate 2.0 in. in diameter. The plate was a part of a piston acting on a spring that pressed on another small plate supporting a strain gage. The spring allowed the subject to make a relatively large movement while the strain gage received only a proportionately smaller part of the movement. The same harnessing technique that was used for the chest site was used for the abdomen, that is, a Dacron webbing strap encircling the abdominal region. The force required to operate the abdominal transducer was 1.5 lb.

The pectoralis transducer consisted of a metal frame which supported a horseshoe-shaped piece of spring steel. One end of the spring was rigidly attached to the frame, and the other one rested upon the pectoralis muscle. A strain gage was cemented on the spring. By bulging the pectoralis muscle the operator strained the spring steel and thereby the strain gage. The force required to operate the transducer was 0.6 lb.

INITIAL SITE AND SIGNAL SELECTION

Abdominal muscle sites were chosen for investigation because abdominal muscles are usually intact in the upper-extremity amputee and the abdomen is well away from body areas involved in passive positioning and/or support of an attached prosthesis. Drawbacks to the use of these sites are: 1. The abdominal wall appears to act as a whole in expansion and contraction and true isolated contraction may not be possible for localized parts of it; 2. the abdominal wall is involved in breathing and in postural support and, depending on the activity of the individual, a high level of unwanted activity may occur; and 3. individuals vary greatly in the amount of overlying surface fat in the abdominal region. The last fact could result in the need for highly individualistic modes of transducer attachment and signal coding. Exploratory studies were undertaken to resolve some of these questions.

Three student subjects were recruited for the first phase of investigation. They represented poor, average, and good levels of muscular "definition" and absence of surface fat. Figure 1 shows the six abdominal sites chosen for study. Two carbon granule transducers were attached to the skin with adhesives at various combinations of pairs of these sites.

Preliminary results indicated the feasibility of using these muscles and it was decided to carry out a more intensive study with one subject. A right unilateral above-elbow amputee was fitted with two carbon-type transducers at locations corresponding roughly to sites 1 and 4 of Figure 1. Performance with this combination of sites and transducers was measured for four control tasks.

On the average the subject was able to generate signals and alternate between muscles sites at rates as fast as one per second. The number of errors and percent of time the task could not be completed decreased from the first to the last of ten 50-minute sessions. As would be intuitively predicted performance was best at the slower rates of alternation and signal generation.

The subject had no difficulty maintaining a fairly constant pressure at either site up to pressures of 2 p.s.i. for 15-second periods with visual feedback provided.

One-dimensional tracking ability was measured using one site at a time. The tracking function was a dot moving at frequencies from 0.05 to 1.00 Hz on the face of a cathode ray tube. Performance was about the same for each muscle site. Little improvement was found over five 3-hour training sessions. Error scores were much lower for the slower tracking frequencies of 0.05 and 0.10 Hz.

Performance was also measured using both muscle sites together in two-dimensional tracking of a dot moving at a frequency of 0.05, 0.10, or 0.20

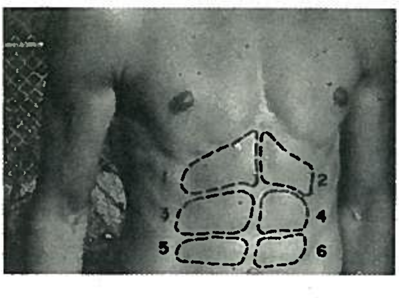

FIGURE 1.—Abdominal muscle sites.

Hz at a 45 deg. angle on the face of a cathode ray tube. Figure 2 shows that performance improved greatly over the first four of eight 90-minute training sessions. Little change in performance was noted over the last four sessions. Average (r.m.s.) error scores were about the same for all frequencies used.

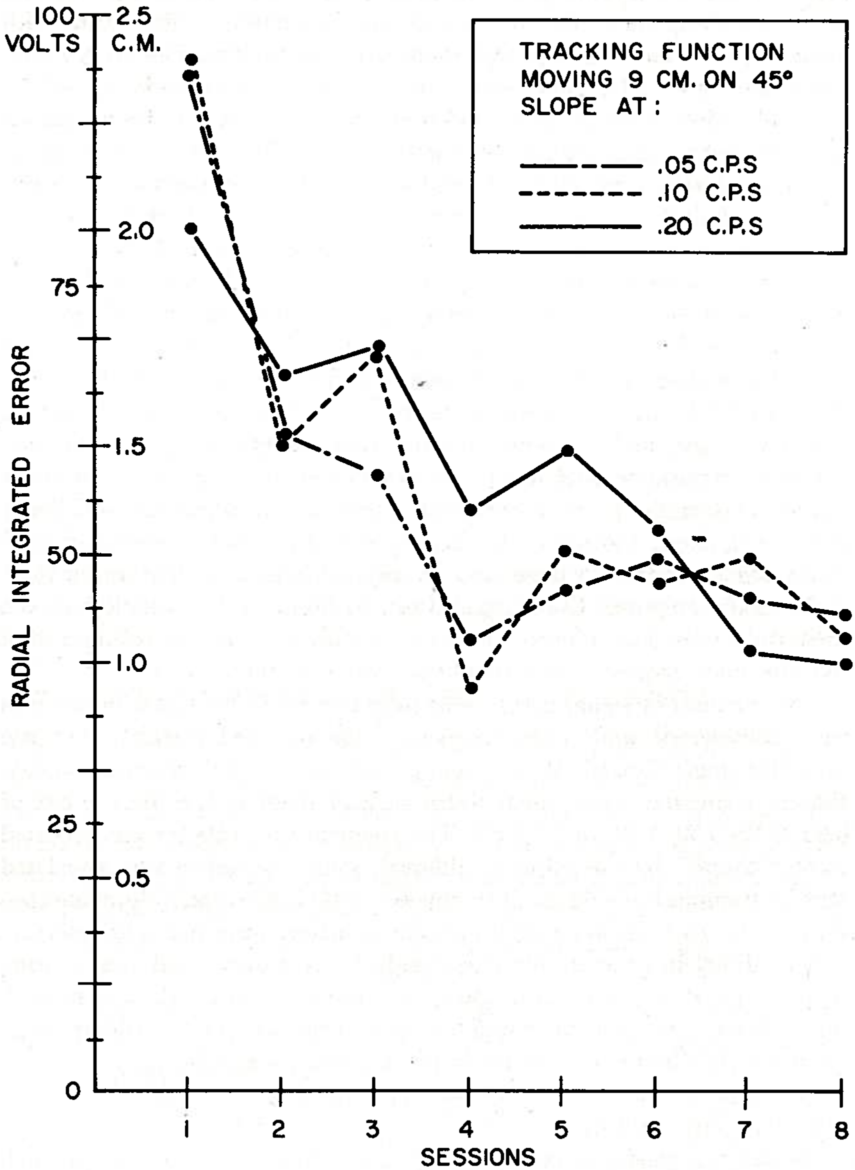

FIGURE 2.—Two-dimensional tracking with abdominal muscles.

With the abdominal sites certain functional limitations were indicated. Particularly for extended duration tasks the subject fatigued rapidly. On the tracking tasks it became apparent that breathing rate was affected by tracking frequency. Furthermore, it was not clear that a real functional isolation had been achieved by the subject. Rather, he appeared to have learned to utilize one of two gross patterns of extending his abdomen in order to activate one of the two transducers. In addition to interfering with breathing these patterns appeared to involve high physiological energy costs which may have contributed to rapid fatigue onset.

In light of these results additional sites were sought using three untrained subjects. A strain gage transducer was attached to the chest by straps. Strain gage transducers were also attached at locations overlying the pectoralis, trapezius, latissimus dorsi, and sacrospinalis muscles. The subjects were asked to generate discrete or continuous output patterns from combinations of these sites while standing, sitting, or bending to lift light weights. The subjects could generate high amplitude signals with the chest, pectoralis muscle, and to some degree with the trapezius muscle sites. Isolation of function was achieved between chest and trapezius, and between chest and pectoralis sites. Since the latter combination provided the best signals, the trapezius, latissimus dorsi, and sacrospinalis sites were dropped from further consideration. It was decided to include one of the two abdominal sites used earlier and study performance with three widely separated locations: abdominal, chest, and pectoralis sites. Special strain gage transducers were developed and fitted to these sites for two subjects, a student and a right above-elbow amputee. Good separation was found to be possible between these three sites. The subjects were able to generate signals at one or two of the sites upon request, while avoiding a signal at the other site.

The question of signal coding was next raised. Three signal dimensions were considered: amplitude, frequency (for repeated signals), and rise time (for single signals). With training, subjects were able to generate continuous triangular wave signals from each of these sites at frequencies of 0.05, 0.10, 0.50, 1.00, or 1.50 Hz. The frequency of 0.50 Hz was reported most "natural" by the subjects, although some discomfort was associated with continuous tracking at all frequencies tested. Attempts to provide continuous signals from more than one site at a time were not satisfactory.

The ability to generate discrete signals, i.e., contract, hold, and release, at each site at one-third, two-thirds, or maximum amplitude was studied. Subjects were not able to reproduce these levels accurately without visual feedback. Furthermore, they could not maintain a steady signal level. On the average, signal level declined the longer the signal was held; the subjects were not aware of the drop.

As another condition subjects were asked to generate a discrete high amplitude signal at their own choice. When initiating such signals subjects were asked to increase signal amplitude either slowly or fast, i.e., to vary the rise time of the discrete signal. They were able to generate slow or fast rise time signals fairly well without visual feedback. Fast signals took an average of 0.1 second to reach peak amplitude, while slow signals took 0.3 second. Signal amplitudes varied widely, but were generally 50 percent or more of maximum. Again it was found that signal amplitude decline with time.

While the results of the site and signal selection experiments did not yield detailed practical design specifications, it was decided that the data were encouraging enough to justify a new set of experiments. It was planned that the additional experiments would utilize a specific control logic and an "arm simulator" which could provide a more realistic set of visual geometric relationships than were obtainable with an oscilloscope.

The following system for generation of control signals was selected:

1. Three muscle sites would be used (abdominal, chest, and pectoralis sites).
2. Discrete signals would be generated at each site.
3. Signals from each site would be combined sequentially or simultaneously with signals from the other sites.
4. Each signal could have either a fast or slow rise time.
5. Signal amplitude would not be considered a critical factor as long as it could be maintained above a predetermined threshold level.

Implementation of these experiments required the development of the arm simulator and control signal processing network described in the following sections.

PROSTHESIS SIMULATOR AND CONTROL SYSTEM

The only multidimensional electrically actuated "arm simulator" available was a powered arm brace capable of movement in several dimensions. The arm brace had been constructed at Spacelabs, Inc., Van Nuys, California, for use in EMG control studies (2) and was obtained without cost. It was modified at UCLA to provide three movements somewhat analogous to movements of the human arm. These movements were termed forearm abduction-adduction, humeral abduction-adduction, and elbow rotation. The arm simulator is shown in Figure 3.

Each segment of the arm was driven by a d.c. servomotor through a Boston multijaw coupling and two Boston miter gears. The speed of motion of each segment was 30°/sec.

When any two of the three powered segments of the simulator were moved together, the tip of the simulator described one of the three surfaces of Figure 4. When all three segments were moved together the tip of the simulator could be positioned anywhere within the volume shown in Figure 5.

The simulator was used for training and for testing amputee performance on one-, two-, or three-dimensional end-point control tasks. The purpose of these tasks was to position the simulator so that its end touched a prepositioned 1-in. diameter steel ball. A steel wire brush ($\frac{5}{16}$ in. long, $\frac{3}{16}$ in. wide) was attached to the end of the simulator. Contact between brush tip

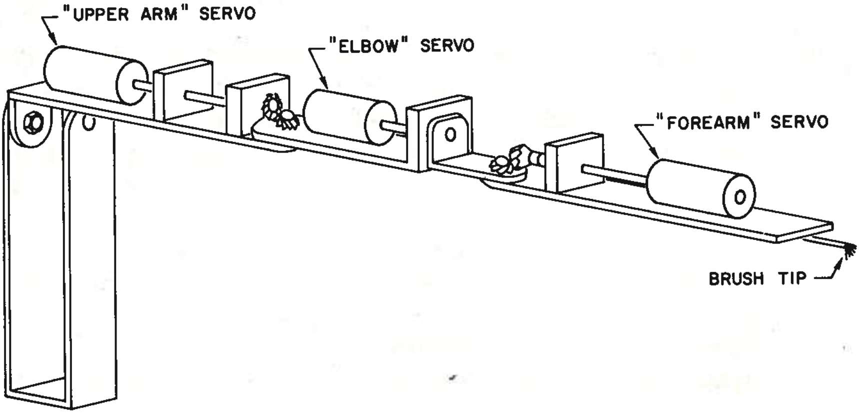

FIGURE 3.—Arm simulator.

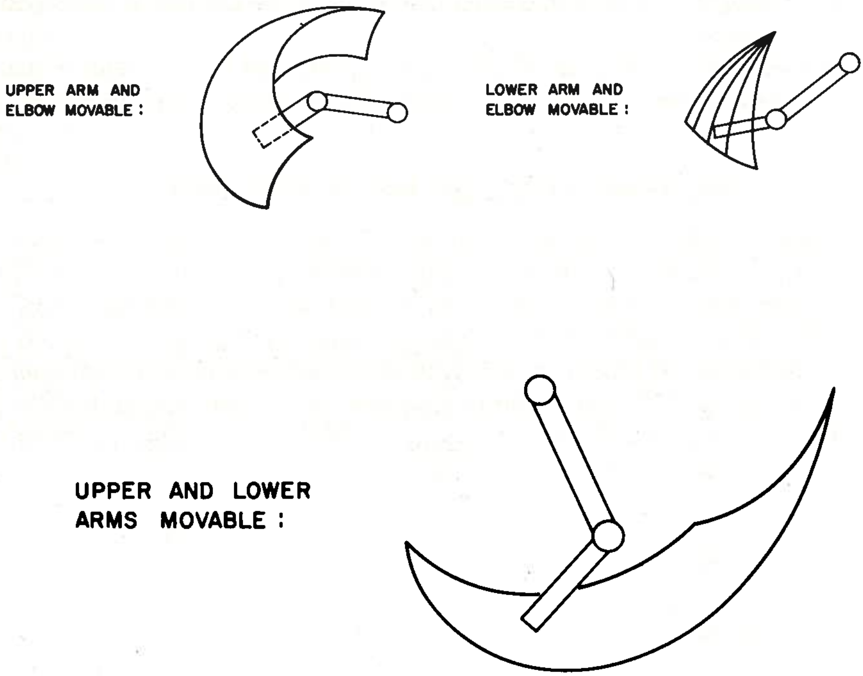

FIGURE 4.—Two-dimension movement spaces of the arm simulator.

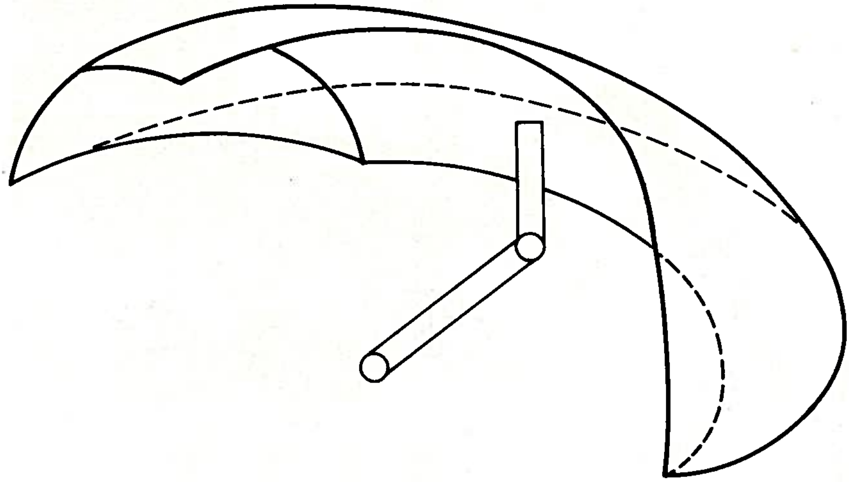

FIGURE 5.—Three-dimension movement space of the arm simulator.

and steel ball was electrically recorded. In addition a visual indicator was turned off as long as contact was maintained.

An electronic, solid-state logic system was designed to process signals generated by upper body muscle sites for the control of the three-dimensional arm simulator. Three strain gage transducers were located on the upper torso of the operator at abdominal, pectoralis, and chest sites. The operator with transducers in place and the arm simulator are shown in Figure 6.

Each transducer was permanently coupled to a separate motor of the simulator. In the selected configuration, movement of each joint of the simulator was controlled in both directions by a separate muscle site. Simultaneous parallel control of all three dimensions of movement of the simulator was possible through the three independent information channels provided by the three transducers and their associated control circuits.

Three phases of control may be distinguished in the operator's execution of a movement along one of the dimensions of movement of the simulator. In the first phase the operator selected both the dimension and the direction of the desired movement by contracting against the proper transducer. The direction of movement was determined according to whether or not the rate of his contraction exceeded a preset value.

Once the simulator started to move in the desired direction the operator could then vary velocity by generating a sawtooth wave in his muscle contraction as shown in Figure 7. Alternately, he could choose to allow the simulator to move according to a preprogramed velocity function, in which velocity gradually decreased as a function of time, merely by maintaining a contraction at least 15 percent above threshold. Figure 8 shows the latter mode. The first option, termed active speed control, was useful only for long

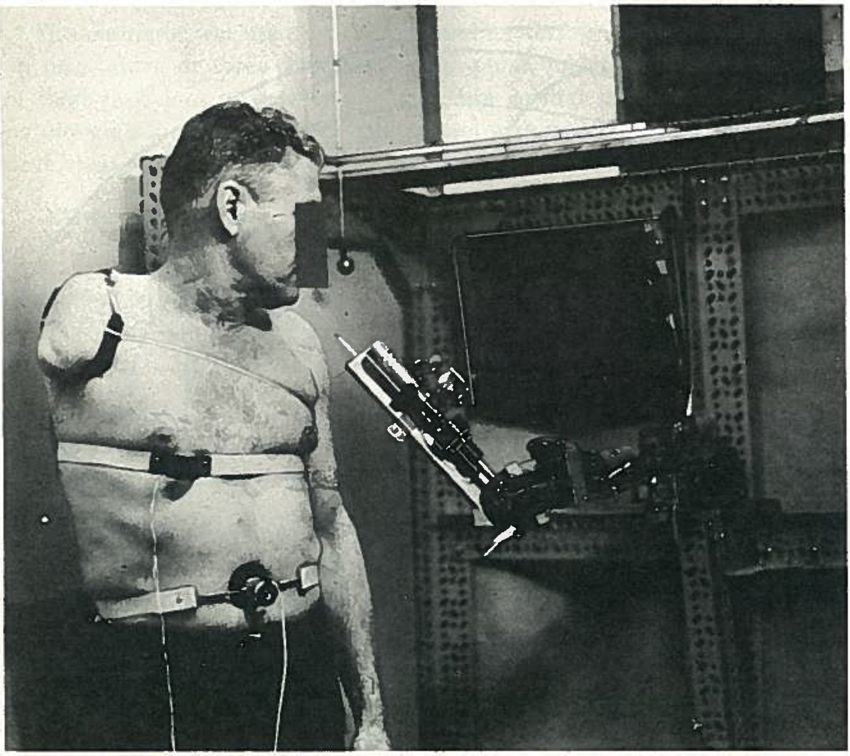

FIGURE 6.—Amputee subject instrumented for control of arm simulator.

movements in which it was desirable to maintain a high rate of speed for some time. The second option, termed preprogramed velocity control, was useful for the approach to a desired end-point of movement since velocity continued to drop with time of movement. This latter arrangement allowed finer control of approach to the end-point as the distance the simulator moved increased.

The third phase of control was termination of movement and was achieved by cessation of muscle bulge against the transducer.

A block diagram of one control channel is shown in Figure 9. Each channel consisted of four basic functional units: a muscle transducer, a directional switching decoder circuit, an analog velocity decoder, and a power amplifier. The operation of this arrangement will be briefly summarized here.

The operator's initial signal was a muscle bulge in the form of a ramp with rise time or steepness voluntarily determined. The transducer converted the muscle bulge into an electrical signal and fed it into the directional decoder which processed it and generated either a positive or negative output signal, according to the steepness of the ramp. The output signal

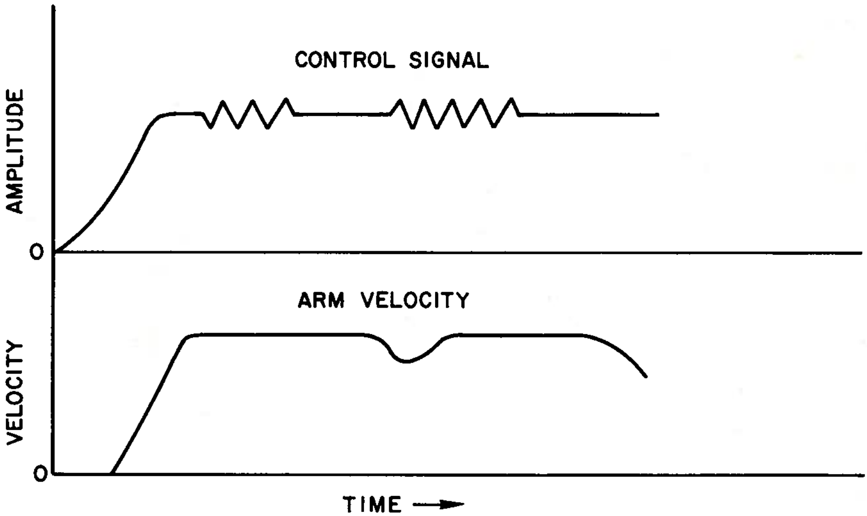

FIGURE 7.—Active speed control.

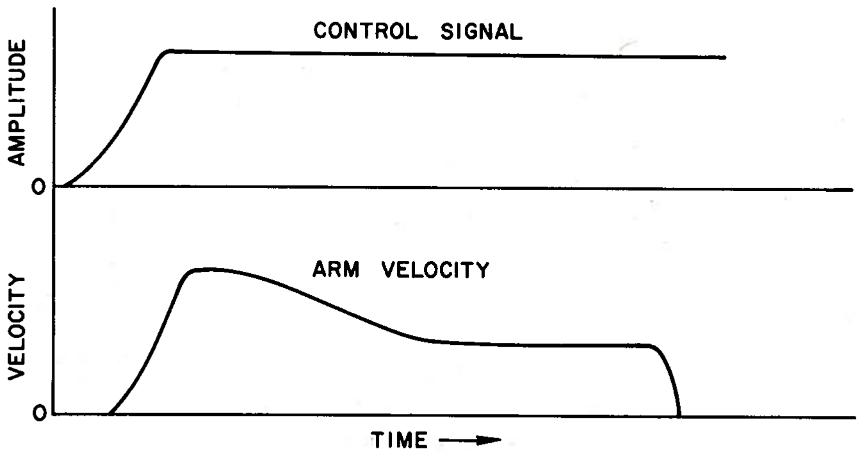

FIGURE 8.—Preprogramed velocity control.

(+ or −) activated one of two circuits controlling opposite directions of motion of the same joint. Each of these circuits had a velocity decoding unit that produced an output proportional to the instantaneous frequency of a sawtooth input provided by the operator for active speed control. This output fed to an AND gate which controlled the input to the power amplifier generating motor output. If no sawtooth was generated by the operator, but his input was maintained at least 15 percent above threshold, velocity was determined by the decaying output of the decoding circuit, which was preset by its time constant.

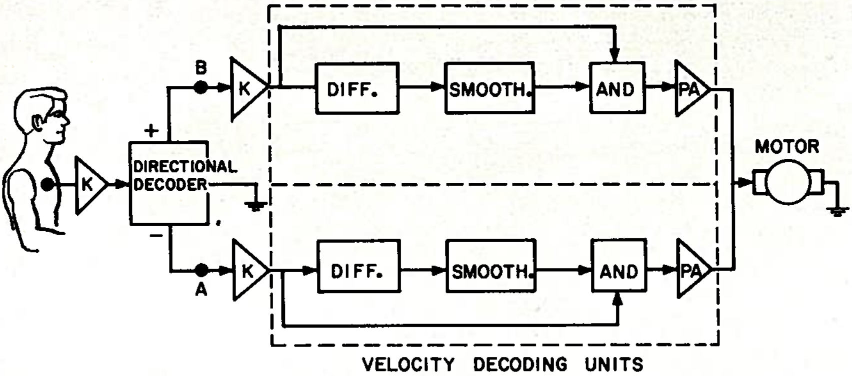

FIGURE 9.—Block diagram of one channel of the control system.

To keep the arm simulator moving in a particular direction, the operator had to generate a threshold input of about 15 percent of maximum. If his input was reduced below that value the input to the power amplifier was cut off by the AND gate through a feed-forward loop which stopped the motor instantly.

To reverse the direction of arm movement, the operator had to terminate his input signal and then generate a signal with a different rise time in order to switch the directional decoder to the opposite direction.

EXTENDED TRAINING AND ASSESSMENT

An extended training study was undertaken to assess the feasibility of the configuration of muscle sites and signal coding that had been developed in the control of a more realistic output device. Two subjects were used. The first (S1) was a 27-year-old university student. He was not an amputee. The second (S2) was a 45-year-old right-above-elbow amputee.

The output device used in this study was the high-mass, high-inertia arm simulator. The three strain gage transducers and the electronic signal processing logic described previously were used to drive the simulator.

The simulator was used as a positioning device only, with movements possible in two directions around three joints. No attempt was made to include a grasp function. Table 1 shows which subject control signals drove each movement of the simulator.

Five test situations and associated performance measures were employed:

1. Response time tasks—in situations similar to those for simple and complex reaction time (RT) testing.
2. Switching time tasks—the time taken to halt an ongoing motion of the arm and begin another.
3. One-dimensional (1-D) end-point control tasks—static positioning

of the simulator at a predetermined end-point, with only one segment of the arm movable.

4. Two-dimensional (2–D) end-point control tasks—with two segments movable.
5. Three-dimensional (3–D) end-point control tasks—with all three segments movable.

TABLE 1.—*Subject Control Logic*

Movement	Control signal	
	Transducer location	Signal rise time
Forearm abduction	Pectoralis site	Fast
Forearm adduction	Pectoralis site	Slow
Upper arm abduction	Chest site	Fast
Upper arm adduction	Chest site	Slow
Elbow rotation, in	Abdominal site	Fast
Elbow rotation, out	Abdominal site	Slow

The testing schedule and number of trials per session are given in Table 2. The tests took place over a period of about 3 months. During that time, S1 completed about twice as many test sessions on the end-point control tasks as did S2, who was employed on a full-time basis away from UCLA.

Simple and complex response times were measured with an experimental procedure similar to that employed for RT measurements with the AIPR arm (4). Two components of response time were measured, subject RT and the lumped electrical/mechanical lag (system delay) of the arm simulator and control logic system. These measures were made early in the testing schedule and again at the end.

TABLE 2.—*Schedule of Test Sessions*

Value	Subject	1–D End-point task	Re-sponse time	2–D End-point task	3–D End-point task	Re-sponse time	Switch-ing time
Number of sessions	Normal (S1)	5	1	13	30	1	1
	Amputee (S2)	3	1	7	11	1	1
Trials per session	Both	60	60	60	32	60	300

Switching time, basically the same measure as response time with the added requirement of having to stop an ongoing movement before beginning the required motion, was measured for each amputee at the end of the testing series.

The end-point control tasks, with the subject controlling one, two, or all three segments of the arm simulator, have been outlined in the section describing the arm simulator. In the 1–D tasks, a total of 60 trials was performed per session, 20 per arm segment, equally distributed over four combinations of arm starting location and target position. Sixty trials were also performed in each session of the 2–D task, 20 for each of the three possible combinations in which two of the three segments were made movable. Each set of 20 trials was equally distributed over four combinations of arm starting position and target location. In the 3–D tasks, 32 trials were performed in a session, equally distributed over 16 combinations of arm starting position and target location.

Two sets of performance measures were taken on the end-point tasks: 1. the time to complete a trial; 2. the number of movements made in each trial. These measures were further categorized as to whether only one arm segment was active (1–D time and moves), two segments were active simultaneously (2–D time and moves), or all three segments were in motion (3–D time and moves). The trial time when no segment was in motion (0–D time) was also recorded.

Analysis of variance and Duncan's range tests were used to test the data for statistically significant differences. Spearman rank correlation coefficients (3) were used to test for significant correlations between variables. The 0.01 significance level was used throughout.

System response and switching times are presented in Table 3 averaged over subjects. Only the response times measured at the end of the test series are presented. Average simple RT (Column A of Table 3) was 0.40 second and average complex RT (Column D) was 0.85 second, giving an average decision time of 0.45 second (Column G). No significant differences were found between mean simple RTs for the six arm movements. The same was true for complex RTs with the exception of four of the 15 comparisons.

Average total response time was 1.06 seconds in the simple case and 1.47 seconds in the complex case (Columns C and F). Mean response times for the six movements considered separately ranged from 0.60 second to 1.87 seconds in the simple case and from 1.10 seconds to 2.10 seconds in the complex case. Differences between the means were significant with the exception of one of 15 comparisons in the simple case, and two of 15 comparisons in the complex case. The factor responsible for these differences was system delay time, as inspection of Columns B and E suggests. Average overall system delay did not change from the simple to the complex testing

TABLE 3.—*Summary of Response and Switching Times*

Movement	Simple response time			Complex response time			Decision time G (D–A)	Switching time as a function of motion switched to H	Switching time as a function of motion switched from I
	A Subject RT	B System delay	C Total (A+B)	D Subject RT	E System delay	F Total (D+E)			
Forearm abduction	0. 42	0. 28	0. 70	0. 81	0. 29	1. 10	0. 39	1. 24	1. 71
Forearm adduction	0. 42	0. 91	1. 33	0. 81	0. 83	1. 64	0. 39	1. 77	1. 46
Upper arm abduction	0. 38	0. 41	0. 79	0. 71	0. 61	1. 32	0. 33	1. 40	1. 56
Upper arm adduction	0. 44	1. 43	1. 87	0. 99	1. 11	2. 10	0. 55	2. 22	1. 50
Elbow rotation, in	0. 38	0. 67	1. 05	0. 92	0. 60	1. 52	0. 54	1. 73	1. 52
Elbow rotation, out	0. 37	0. 23	0. 60	0. 85	0. 29	1. 14	0. 48	1. 17	1. 59
Average	0. 40	0. 66	1. 06	0. 85	0. 62	1. 47	0. 45	1. 59	1. 59

NOTE.—All values are in seconds.

case. The variation in system delay from movement to movement reflected the large differences in inertial load imposed on the motors of the arm. These differences in load were a function both of the direction of movement with respect to gravity and the number of other segments of the arm that the motor was supporting. The change in mean system delay from the simple to complex case was particularly large for the movements of the upper arm, reflecting mechanical unreliability of the unit.

Average switching time was 1.59 seconds. Mean switching times considered as a function of the motion switched from (Column I) varied little from movement to movement. The same data, when considered as a function of the motion switched to (Column H), showed differences between the means that were significant with the exception of one of 15 comparisons. The rank order correlation between mean switching times (Column H) and complex response time (Column F) was significant, suggesting further that the motion switched to was the determining factor in the variation in length of switching time. The added complexity in the task of having to halt one motion and start another required only a 0.12 second increase over complex response time, indicating that the subjects were able to overlap these activities easily.

Comparison of these results with values obtained with the AIPR arm (4) indicated that a reasonable level of skill in control activation had been reached by the end of the test series. Subject RT values were the same in the simple and complex testing situations with the AIPR arm as with the arm simulator. System delay was considerably higher with the arm simulator due to electromechanical factors, accounting for the 0.3 to 0.4 second increase in overall response time found with the simulator.

Performance on the three end-point control tasks is presented separately for each subject in Figures 10 and 11. It should be noted that the number of test sessions differed for each subject on each task as shown in Table 2 and that these figures are therefore only gross indicants of performance change.

The reductions in average trial completion time that occurred from the first to the last test session, as shown in Figure 10, were significant in all six cases. The reductions in number of movements per trial, as shown in Figure 11, were also significant except for the case of S1 on the 2–D task. The figures show that reductions in average trial time and average movements per trial were closely paralleled by reductions in 0–D time (inactive time) and in the number of 1–D movements.

Further evidence for this interpretation is given by Tables 4 and 5 which summarize rank correlations between selected performance measures for the 3–D task. High, significant correlations were found between daily averages for trial time, total movements, 0–D time, and 1–D movements for each subject considered separately.

Figure 12 further illustrates the close dependence of average trial time on 0–D time. Fifteen-second drops in trial time and 0–D time were shown by S1 on the 3–D task over sessions, while 1–D, 2–D, and 3–D time showed relatively minor changes.

Comments of the subjects, coupled with the objective results, indicated that two problems existed in mastering control of the simulator. The first

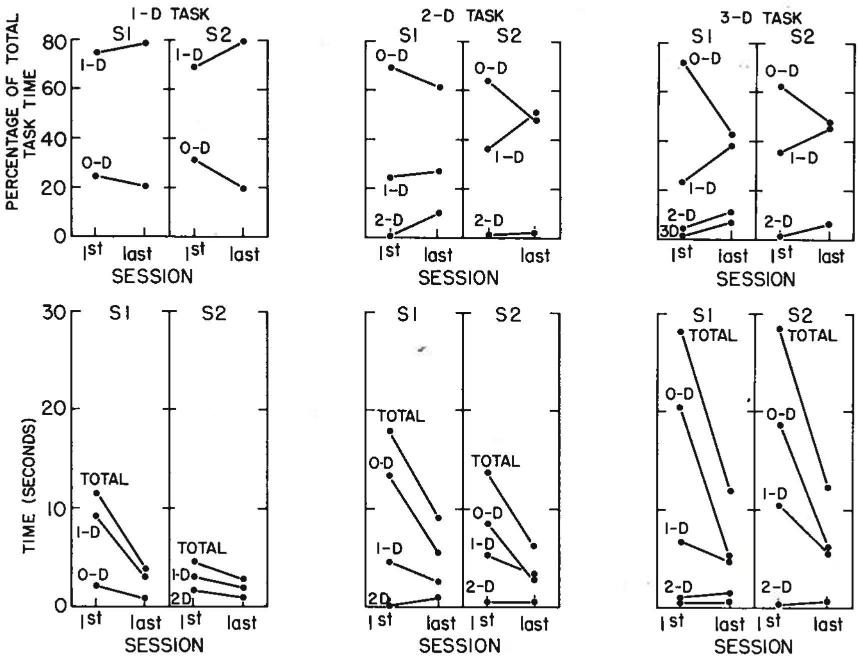

FIGURE 10.—Skill acquisition: Change in task completion times, relative and absolute basis. Data points summarize results of 32 trials for each subject on each task.

TABLE 4.—*Rank Order Correlation Coefficients for Performance Measures of S1 on 3–D End-Point Task*

Performance measure	Average 0–D time	Average number of movements	Average 1–D movements	Average 2–D movements
Average trial time	[a]0.980	[a]0.523	[a]0.807	−0.121
Average 0–D time		[a]0.646	[a]0.687	−0.022
Average number of movements			[a]0.894	0.357
Average 1–D movements				−0.378

[a] $p < .01$.

problem was the obvious one of learning the movement patterns of the arm simulator and the control signals necessary to generate desired movements in the proper amount. The second problem was learning to predict accurately where the arm simulator would stop at the end of a movement.

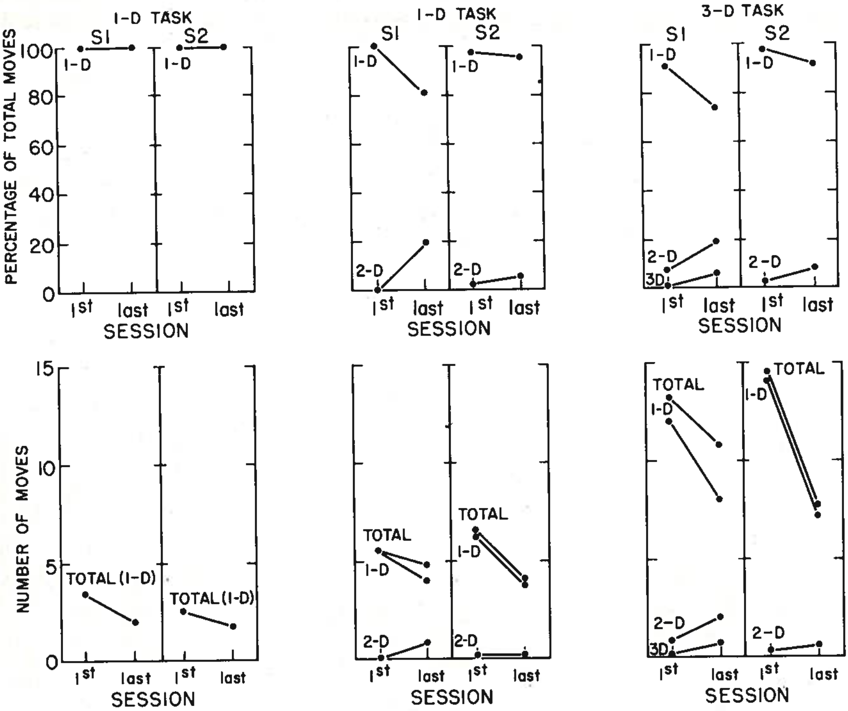

FIGURE 11.—Skill acquisition: Change in movements per trial, relative and absolute basis.

TABLE 5.—*Rank Order Correlation Coefficients for Performance Measures of S2 on 3-D End-Point Task*

Performance measure	Average 0–D time	Average number of movements	Average 1–D movements	Average 2–D movements
Average trial time	[a]0. 973	[a]0. 882	[a]0. 900	—0. 500
Average 0–D time		[a]0. 836	[a]0. 832	—0. 527
Average number of movements			[a]0. 991	—0. 155
Average 1–D movements				—0. 191

[a] $p < .01$.

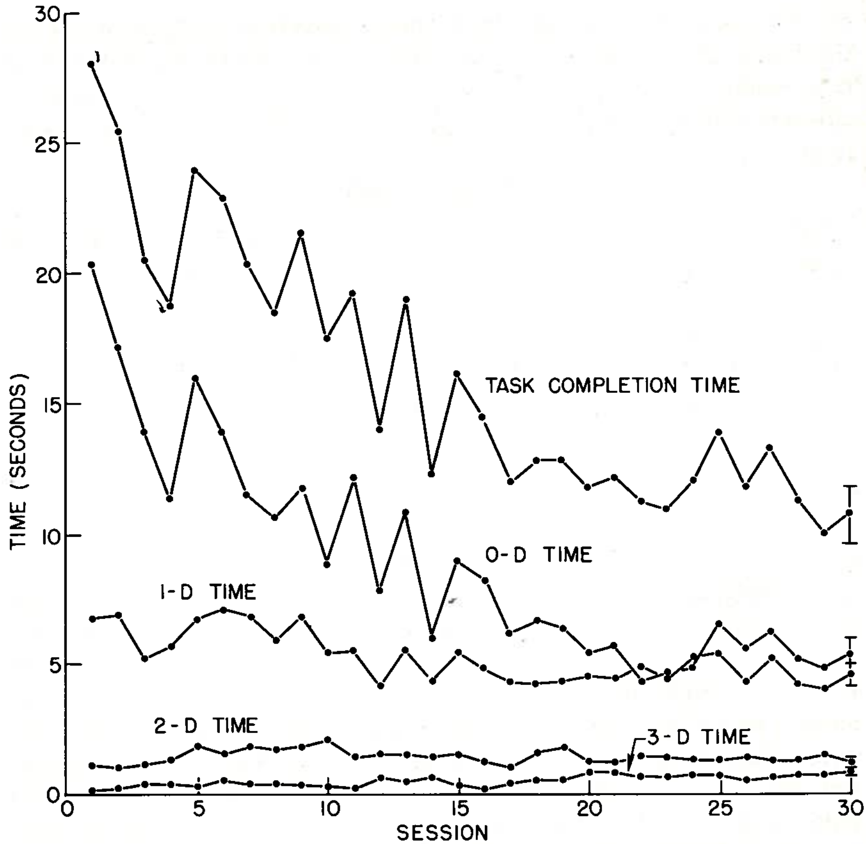

FIGURE 12.—Performance of S1 on three-dimensional end-point task.

The high inertia of the arm and the mechanical nonlinearities in the gears made it quite difficult to predict exactly where the arm would stop. These factors coupled with the fairly rigorous positioning criterion imposed made the cost of reaching end-point high in terms of time and movements. As a result, in the 3–D tasks the subjects apparently adopted a "move and wait" strategy (5) at first, preferring to let the arm come to a complete halt before beginning a corrective motion. In later sessions they were able to predict the result of a movement more accurately and could overlap corrective motions with the ongoing motions. Inspection of the timing data showing the initiation of control signals supported this observation.

Figure 12 indicates that little additional improvement could be expected for S1 on the 3–D task with further training. The data for S2 (not shown) indicated that learning was not complete and that his performance might have approached a much lower asymptote with more testing than did S1. However, both subjects were able to control the simulator well by the end of testing in spite of the difficulty of the task and the time lags inherent in the system. Only S1 showed a significant increase in the absolute measures of coordinated control over test sessions (2–D and 3–D time and moves), although both subjects showed relative increases in these measures in Figures 10 and 11.

CONCLUSIONS

These studies indicate that mechanical muscle bulges generated from upper body muscle sites can be successfully sensed and utilized to position an externally powered upper-extremity prosthetic system. Using three upper body control sites, two subjects developed proficiency in the control of an arm simulator that was capable of three-dimensional movement. A number of limitations inherent in the test situation, particularly in the arm simulator, did not allow realization of the full potential of the control technique. Nevertheless, the results obtained were sufficiently promising to recommend that additional research should be undertaken.

Specifically, studies should be conducted to determine which control motions may be optimally linked to available arm motions; whether faster rise time signals or signal dimensions other than rise time could be used for control; and whether a hybrid control system combining conventional harness control motions with the control motions developed in this study would be feasible. In implementing these suggestions, it is further recommended that a lightweight, fast-response arm simulator be constructed which would eliminate the long system response time lags found with the present arm simulator. The simulator should be capable of geometric movement patterns that coincide more closely with positioning movements of the intact arm. It might be possible to utilize existing electrically powered prostheses or remote manipulators for additional research with the control technique. Finally, a larger number of subjects, preferably representing various levels of above-elbow amputation, should be trained in the use of proposed simulator. Use of such a group would facilitate generalization of results to a broader range of amputation levels, would reduce the susceptibility of results to individual differences in the subjects tested, and would allow more sensitive experimental designs to be used in evaluating promising prosthetic control configurations.

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