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DEVELOPMENT AND CONTROL OF THE UTAH ARM

Jacobsen, S.C., Jerard, R.B., and Knutti, D.

The ultimate goal of the artificial arm project at the University of Utah is the development of a multiaxis prosthetic arm and its practical application as a rehabilitation tool. This paper reviews progress towards that goal. A prototype "Utah Arm" with three independently powered degrees of freedom has been recently completed. A control theory for multiaxis artificial arms is being developed and results of initial tests of that theory are presented.

Introduction

Our previous papers have discussed the following:

1. Theoretical basis for a controller for artificial arms (1)(4).
2. A preliminary report on the development of the Utah artificial arm hardware (2)(5).
3. The anticipated computation required for control of the "Utah Arm" (3).

This paper will present:

1. Recent hardware developments in the "Utah Arm" project.
2. Results of the first actual application of our control theory of the "Utah Arm."

Utah Arm Hardware

Implementation of our control theory requires a prosthetic arm with exceptional operating characteristics. Since no prosthetic arms currently available have adequate performance, a great deal of effort has been spent in making a high response torque servosystem with low friction and low inertia. The latest version of the "Utah Arm" is shown in Figures 1 and 2.

Figure 1 - Mr. Jack Wiseman wears the "Utah Arm" in recent tests of the control theory.

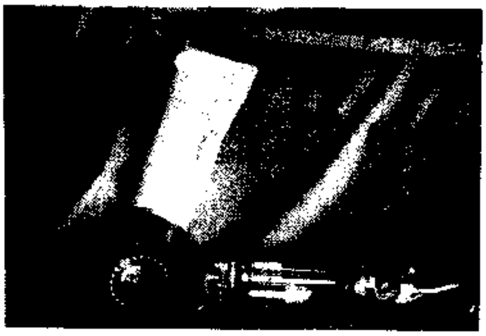


Figure 2 - The arm has three independently controlled, electrically powered degrees of freedom - elbow flexion, wrist rotation, and terminal device closure.

The arm's excellent characteristics are attributable to the use of "LADD" actuators for elbow flexion (5)(8). The arm includes electrically powered elbow flexion, wrist rotation, and terminal device operation. The device, complete with electronics and batteries, weighs approximately 2.5 pounds. It can actively lift four pounds and can passively support 50 pounds. A complete excursion of 135^{o} of elbow flexion occurs in less than one-half second. Another desirable characteristic of the arm is that the generated acoustic noise is less than 53db on the A scale.

The arm also includes a power conserving clutch which mechanically senses both torque and velocity information and then acts to lock the elbow when the amputee wishes to support a static load.

The arm utilizes a lightweight center-pull hook which was developed in our laboratory. The hook can be operated in a configuration which is normally closed, normally opened, or under double acting proportional control.

We have recently developed inexpensive, low power drain, EMG preamplifiers with a common mode rejection ratio of over 95db at 1 KHz. and a very high input impedance (EMGPA) (6). Control of up to four motors is accomplished with a four channel low power drain, pulse width modulated power supply (PWMPS) (7).

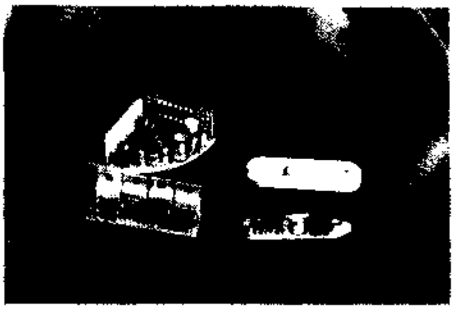


Figure 3 -

A. PWMPS controller.
B. PWMPS power transistor bridge.
C. EMGPA exposed circuitry.
D. EMGPA encapsulated in epoxy resin.

Arm Control Theory

The controller for an artificial arm is shown in block diagram form in Figure 4.

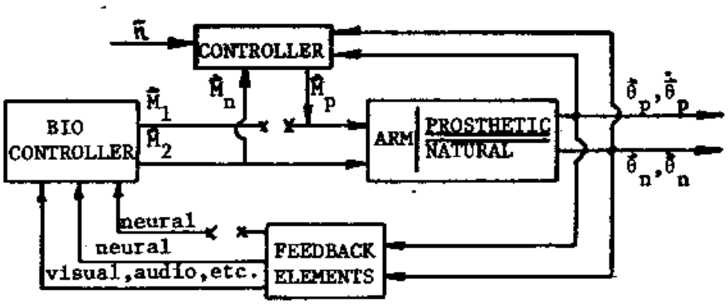


Figure 4 - Block diagram representing the interrelationship between natural arm, prosthetic arm, biocontroller, and the prosthetic controller.

An important observation is that the controller, the artificial arm and the remnants of the natural arm form an interrelated and indivisible system. A controller cannot be defined without understanding the entire system.

The blocks in the diagram represent:

1. Biocontroller. The biocontroller consists of the brain, neuro pathways, and muscles which apply torques to the skeletal frame. Output from the biocontroller is a vector $\bar{M}_2$ which represents the torques applied to the natural remnants of the amputee's arm. The output $\bar{M}_1$ represents the torques which would have been applied to the amputated parts of the arm now replaced by the prosthesis. Note that this pathway is interrupted by amputation. Inputs to the biocontroller are neural, visual, and audio information about the state of the arm-prosthesis combination.

2. Arm. The arm block represents both the natural portions of arm and the electromechanical replacement for the missing arm. Inputs to the arm block are the torques which are applied to the natural and prosthetic joints by the muscles and servomotors. Output of the arm block is the angular position and angular velocity of the prosthesis $(\bar{\theta}_p, \dot{\bar{\theta}}_p)$ and the angular position and angular velocity of the natural portion of the arm $(\bar{\theta}_n, \dot{\bar{\theta}}_n)$.

3. Controller. The controller is essentially responsible for re-establishing communication in the interrupted signal pathway $\bar{M}_1$ (torques which were applied to the amputated portion of the arm). The controller inputs are the kinematic state of the arm prosthesis combination, the vector of torques which are applied by the musculature on the remnants of the natural arm, and the parameter $\bar{\eta}$. The additional parameter $\bar{\eta}$ will be explained later. The output of the controller is the vector of torques $\bar{M}_p$ which should be the same as $\bar{M}_1$.

   Note that the equations describing the controller may be derived from a fundamental postulate which will be discussed shortly.

4. Feedback Elements. Feedback elements are responsible for monitoring the kinematic state of the limb and presenting that information to the biocontroller.

Fundamental Postulate

A controller can be formulated from the following postulate:

> For a given arm prosthesis state $(\vec{\theta},\dot{\vec{\theta}})$ and for an instantaneous set of torques applied to the natural joints by the musculature $(\vec{M}_n)$ the controller should apply torques to the prosthesis $(\vec{M}_p)$ which will cause the clavicle to move in an experimentally determined relationship with the humerus (1)(2)(3).

Controller Development

Implementation of the control theory requires completion of three tasks.

1. Development of arm hardware to mimic the performance of the natural arm (as previously described).
2. Equations for the controller must be derived from the linkage equation based upon the fundamental postulate.
3. Develop the ability to monitor the vectors $(\vec{\theta},\dot{\vec{\theta}})$ and $\vec{M}_n$ which are the input to the controller.

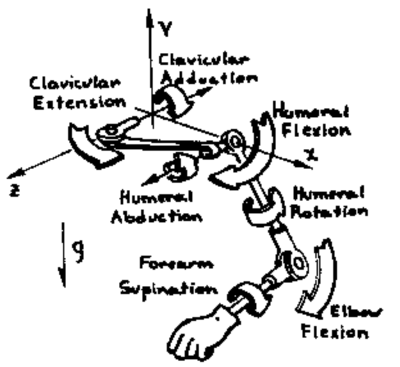
Controller Equations

Derivation of the controller equation begins with the motion equations for the arm linkage as shown in Figure 5. These equations can be written in matrix form as in the following equation.

$$\tilde{P}(\vec{\theta},\dot{\vec{\theta}})\ddot{\vec{\theta}} + \vec{Q}(\vec{\theta},\dot{\vec{\theta}}) + \vec{R}(\vec{\theta}) = \tilde{S}\vec{M}_{all} \qquad (1)$$

$\tilde{P}$, $\vec{Q}$, $\vec{R}$, and $\tilde{S}$ are matrices whose elements depend on the linkage angle vector, $\vec{\theta}$, and angular velocity vector, $\dot{\vec{\theta}}$. $\ddot{\vec{\theta}}$ is the angular acceleration vector. $\vec{M}_{all}$ is a vector which contains the torques applied to the joints of the linkage. Note that for the limb shown in Figure 5, some of the joints

Figure 5 - Linkage which represents the degrees of freedom of the arm.



will be the natural, musculo-skeletal joints of the amputee and other joints will be those of the electromechanical prosthesis.

Observations of the behavior of the normal limb reveal that clavicular extension and clavicular adduction is correlated with behavior of the remaining limb. That is, for most arm tasks when the arm reaches forward, the clavicle moves forward slightly. The fundamental postulate states that the torques applied to the prosthesis will cause the clavicle to move "in an experimentally determined relationship with the humerus." This relationship is included in the model by the inclusion of a "constraint" ($\tilde{\eta}$) in the equations. This constraint is very much present in natural limb motion and is supplied subconsciously by the biocontroller. The constraint allows the natural limb to be deployed in a manner that is optimum with respect to desired criteria such as speed, energy, force, accuracy, etc.

While establishing the stability and controllability of a postulate-based control system, the constraint $\tilde{\eta}$ will be a major area of investigation. Possibilities include finding a constant $\tilde{\eta}$ that is most compatible with the biocontroller, or the amputee might consciously select $\tilde{\eta}$ with the flick of a switch or perhaps most intriguing is to use artificial intelligence to select $\tilde{\eta}$ on the basis of kinematic and EMG parameters. $\tilde{\eta}$ is included in the controller equations by the following relationship:

$$\vec{\theta}_{constrained} = \tilde{\eta}\,\vec{\theta}_{free} \qquad (2)$$

For the general case, the vector $\vec{\theta}_{constrained}$ contains the axes of clavicular extension and clavicular adduction. Vector $\vec{\theta}_{free}$ contains the remaining degrees of freedom as shown in Figure 5. Maxtrix $\tilde{\eta}$ represents the experimentally determined relationship between the clavicle and the remaining arm.

If $\vec{\theta}$ and $\ddot{\vec{\theta}}$ vectors are partitioned into constrained and free subvectors and if $\vec{M}_{all}$ is partitioned into subvectors which contain the torque applied to the natural joints and the torque applied to the prosthetic joints, then the preceding equations can be combined to eliminate the constrained degrees of freedom as shown in the following equations (1)(3).

$$\vec{\theta} = \begin{Bmatrix} \vec{\theta}_{constrained} \\ \vec{\theta}_{free} \end{Bmatrix}, \qquad \dot{\vec{\theta}} = \begin{Bmatrix} \dot{\vec{\theta}}_{constrained} \\ \dot{\vec{\theta}}_{free} \end{Bmatrix}$$

$$\vec{M}_{all} = \begin{Bmatrix} \vec{M}_{prosthetic} \\ \vec{M}_{natural} \end{Bmatrix}, \qquad \ddot{\vec{\theta}} = \begin{Bmatrix} \ddot{\vec{\theta}}_{constrained} \\ \ddot{\vec{\theta}}_{free} \end{Bmatrix} \qquad (3)$$

$$\vec{M}_{prosthetic} = \vec{\beta}_1(\vec{\theta}_{free}, \ddot{\vec{\theta}}_{free}, \tilde{\eta})\,\vec{M}_{natural} + \vec{\beta}_2(\vec{\theta}_{free}, \dot{\vec{\theta}}_{free}, \tilde{\eta}) \qquad (4)$$

Controller Input

Information necessary to implement the controller includes $\vec{\theta}, \dot{\vec{\theta}}$ and $\vec{M}_n$. The vectors $\vec{\theta}$ and $\dot{\vec{\theta}}$ represent the kinematic state of the arm-prosthesis combination and may be obtained directly by goniometers and tachometers. $\vec{M}_n$ is the vector of torques applied by the natural musculature to the intact skeletal structure. Unfortunately there is no direct way to monitor this information. However, EMG (electromyographic) signals picked up by surface electrodes and suitably processed have been shown to be a fairly good indicator of isometric muscle tension. With a knowledge of how all the shoulder muscles pull on the clavicle-humerus linkage it is then possible to resolve these forces into torques about the sterno-clavicular joint and the shoulder joint. These relationships may be expressed in the following form (1)(3)(4).

$$\vec{M}_{natural} = \vec{G}(\vec{\theta}_{free}, \dot{\vec{\theta}}_{free}) \vec{E}_{\substack{selected \\ cutaneous}} + \vec{N}(\vec{\theta}_{free}, \dot{\vec{\theta}}_{free}) \qquad (5)$$

The matrices $\vec{G}$ and $\vec{N}$ are derivable from a knowledge of musculo-skeletal anatomy, the relationship between muscle tension and electromyographic signals and the effect of sensing EMG's cutaneously rather than at their source.

In practice it would be fairly difficult to analytically derive shoulder torques as a function of EMG's in this manner. An alternative approach is to monitor EMG's while experimentally measuring shoulder torque with suitable instrumentation. Experimental data is stored and then statistical techniques are used to relate the torques to the EMG's in such a way that a selected set of shoulder muscles can be used for accurately predicting torques. The statistical technique used is multivariable linear regression. Since the problem is typically illconditioned, a method called "ridge regression" is utilized to give accurate linear regression coefficients (9).

One Degree of Freedom Experiment

In order to verify the validity of the control theory, we are currently involved in completing a one degree of freedom experiment.

To reduce the general equation to a one degree of freedom controller, a coordinate system for the linkage must be defined as shown in Figure 6.

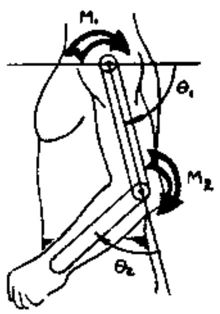


Figure 6 - Coordinate system for the one degree of freedom experiment.

θ_1 represents the degree of freedom shoulder flexion and θ_2 represents the degree of freedom elbow flexion. Torque about the shoulder, M_1, will be determined by monitoring EMG signals of selected shoulder muscles. The elbow torque to be applied to the prosthesis, M_2, will be determined by the controller.

Determination of the control equations begins with the definition of the $\bar{P}$, $\vec{Q}$, $\vec{R}$, and $\bar{S}$ matrices. For the linkage shown, the matrices are:

$$\widetilde{P} = \left\{ \begin{array}{c|c} 2g_1 + 2g_2 \cos\theta_2 & 2g_3 + g_2 \cos\theta_2 \\ \hline 2g_3 + g_2 \cos\theta_2 & 2g_3 \end{array} \right\}$$

$$\vec{Q} = \left\{ \begin{array}{c} -2g_2 \dot{\theta}_1 \dot{\theta}_2 \sin\theta_2 - \dot{\theta}_2^2 g_2 \sin\theta_2 \\ \hline \dot{\theta}_1^2 g_2 \sin\theta_2 \end{array} \right\} \tag{6}$$

$$\vec{R} = \left\{ \begin{array}{c} (w_1 L_{11} + w_2 \ell_1)\sin\theta_1 - w_2 L_{22} \sin(\theta_1 + \theta_2) \\ \hline w_2 L_{22} \sin(\theta_1 + \theta_2) \end{array} \right\}$$

$$\widetilde{S} = \left\{ \begin{array}{c|c} 1 & 0 \\ \hline 0 & 1 \end{array} \right\}$$

m_1 and m_2 are the linkage masses. I_1 and I_2 are the linkage moments of inertia about the links centers of gravity. ℓ_1 and ℓ_2 are the linkage lengths. L_{11} and L_{22} are the lengths from the pivots to the center of gravity of the links.

The motion constraint is expressed by the following equation.

$$\theta_1 = \eta\,\theta_2 \tag{7}$$

The final controller equation is shown below.

$$\begin{aligned} M_2 = K\Big\{ & M_1 + (2\eta+1)(g_2 \sin\theta_2)\dot{\theta}_2^2 - (w_1 L_{11} + w_2 \ell_2)\sin(\eta\theta_2) - \\ & - w_2 L_{22} \sin[(\eta+1)\theta_2]\Big\} + \Big\{(\eta^2 g_2 \sin\theta_2)\,\dot{\theta}_2^2 + \\ & + w_2 L_{22} \sin[(\eta+1)\theta_2]\Big\} \end{aligned} \tag{8}$$

$$K = \frac{2g_3(\eta+1) + 2g_2 \cos\theta_2}{2\eta g_1 + 2g_3 + (2\eta+1)g_2 \cos\theta_2}$$

$$\begin{aligned} g_1 &= \tfrac{1}{2}(m_1 L_{11}^2 + I_1) + \tfrac{1}{2}(m_2(\ell_1^2 + L_{22}^2) + I_2) \\ g_2 &= m_2 \ell_1 L_{22} \\ g_3 &= \tfrac{1}{2}(m_2 L_{22}^2 + I_2) \end{aligned} \tag{9}$$

GAINS* OF EMGS FOR NINE SHOULDER MUSCLES
NECESSARY TO ESTIMATE M_1 (SHOULDER FLEXION)

1. Lower Pectoralis Major	-12,800	6. Posterior Deltoid	-59,790
2. Sternal Pectoralis Major	1,310	7. Infraspinatus	-3,180
3. Clavicular Pectoralis Major	46,500	8. Teres Major	-5,550
4. Anterior Deltoid	70,200	9. Latissimus Dorsi	-104,400
5. Mid Deltoid	23,080	N=0	

Table 1

*Gain=(Filtered DC Output)/(Differential RMS Skin Potential)

This equation computes the elbow torque, M_2, which, for a given forearm state $(\theta_2, \dot{\theta}_2)$ and shoulder torque, M_1, will cause the humerus to move in a manner determined by η.

As previously mentioned the vector $\vec{M}_N$ is theoretically derivable from anatomical and physiological relationships. These relationships may be expressed in the following form for the one degree of freedom experiment.

$$M_1 = \tilde{G}\vec{E} + N = G_1E_1 + G_2E_2 + \ldots + G_9E_9 + N$$

The $\tilde{G}$ matrix and constant $\vec{N}$ use the vector of EMG's, $\vec{E}$, to determine the torque M_1.

There are three steps in the experimental procedure for testing the validity of the control theory.

1. Gather experimental data which compose the M_1 and $\vec{E}$ vectors.
2. Perform a multivariable linear "ridge" regression to determine $\tilde{G}$ and $\vec{N}$.
3. Implement the controller equations for closed loop amputee control of elbow flexion.

The experimental set-up for step 1 is shown in Figure 7. The subject is suitably instrumented to obtain the EMG output from the muscles shown in Table 1. Shoulder torque, M_1, is also obtained by monitoring a load cell connected to the amputee's hook (Figure 7). Data points for each channel are collected at the rate of 100 per second for a total duration of ten seconds.

The experimentally derived regression coefficients (Gains) obtained in step 2 are also shown in Table 1. Note the comparatively small effect of the Sternal Pectoralis Major, Infraspinatius and Teres Major. This indicates that the number of muscles could easily be reduced to six. There's also good reason to believe that this number could be reduced to four because of the high correlation among the remaining muscles.

Figure 7 - EMG's from nine shoulder muscles (E) and shoulder torque (M_1) are simultaneously recorded at the rate of 100 samples per second.

The controller for the third step of the experiment is a DEC PDP 11/40 minicomputer which contains the appropriate controller equations and which receives the nine channels of filtered and rectified EMG's as well as θ_2 and $\dot{\theta}_2$. The computer calculates M_1 using the regression coefficients given in Table 1 and then finds M_2 which is output to the analog torque servo-system. The value of M_2 is updated at the rate of 30 times per second with no perceivable discontinuity in the arm control.

The complex nature of the controller and associated time delays due to necessary filtering raised some questions about the stability of the system. However, initial experiments have proven that stability is not a problem as long as the forward loop gain of the torque control system is kept below a certain maximum value. The elbow was controllable, smooth, and did not exhibit oscillatory behavior for a variety of values of η between 0 and 1, although best performance was obtained when η approaches 0. It was not necessary to add any damping to the system although we intend to investigate this possibility to see its effects on stability as η and the gain of the torque servo vary.

Summary

A theory for multiaxis artificial arm control based on a fundamental postulate has been established. Hardware necessary for verification has been built and a test of the theory as used to control elbow flexion has been made. The test showed that EMG's can adequately estimate $\vec{M}_n$ for the case with the shoulder fixed. Also shown is that stability is not a problem and that control is smooth and natural for at least elbow flexion. Further experiments will expand the implementation to humeral rotation and wrist rotation.

Acknowledgments

The efforts described in this paper have been accomplished at the University of Utah in collaboration with the Liberty Mutual Research Center at Hopkinton, Massachusetts. The project has been supported in part by the National Science Foundation and Mr. Maurice Warshaw of Salt Lake City, Utah. Grateful thanks is also extended to Dr. Willem Kolff who is the Director of the Division of Artificial Organs at the University of Utah.

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