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COMPUTATIONAL REQUIREMENTS FOR CONTROL OF THE UTAH ARM

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Artificial Arms, Prosthetic Control Systems, Computer Controlled Limbs

Figure 1 illustrates an above-elbow amputee fitted with a three-degree-of-freedom prosthetic arm. The development of a successful prosthetic arm requires the completion of two tasks. First, a controller must be devised which can determine an amputee's desired limb motions by monitoring a set of amputee generated signals. For example, the controller monitors a set of EMG signals from selected muscles and the kinematic state of the limb. Secondly, an electromechanical arm must be developed which can replace the amputee's missing musculo-skeletal components.

The principal objective of this paper is to discuss the computational requirements for implementation of a control theory currently being developed at the University of Utah (Refs. 1, 4). Brief mention will also be made of other aspects of our artificial arm development project. Therefore, the following text is subdivided into four sections.

I. The Electromechanical Hardware Currently Being Developed at the University of Utah (The Utah Arm)

II. The Requirements for Artificial Arm Control

III. A Review of the General Limb Control Theory

IV. Specific Application of the Control Theory to Command of a Limb with Powered Elbow Flexion

I. THE UTAH ARM

A photograph of the initial "Utah Arm" prototype is shown in Figure 2 Ref. (2). The device is actuated by a unique new mechanism (LADD) which is further detailed in Ref. (3). The actuator permits realization of a multi-axis artificial limb which is superior to prior art in many respects. The arm is acoustically quiet (50 db), lightweight, inexpensive, and highly responsive. The device shown incorporates the degrees of freedom of elbow flexion, forearm pronation, and terminal device closure. Later prototypes will be capable of powered humeral rotation.

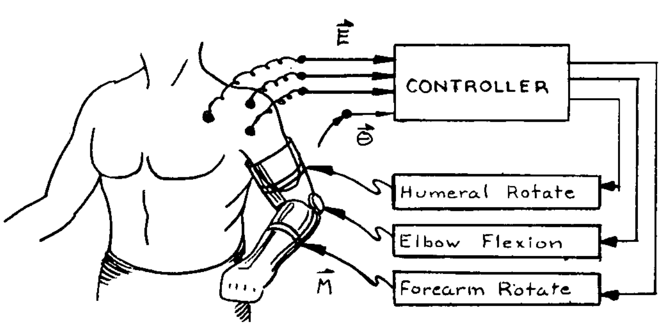

Fig. 1

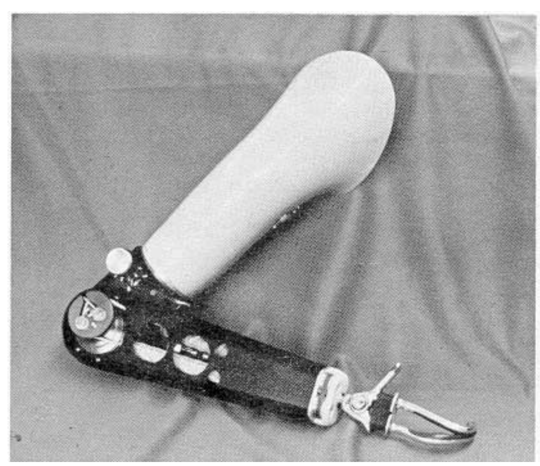

Fig. 2 The "Utah Arm" is an externally powered prosthesis for above-elbow amputees.

The limb is powered by a small battery pack located within the humeral shell and utilizes pulse width modulated power supplies for electric motor control.

II. ARTIFICIAL ARM CONTROL REQUIREMENTS

Consider a controller which utilizes as its inputs the EMG signals from a selected set of arm and shoulder muscles (vector $\vec{E}$ contains the EMG signal magnitudes) and the kinematic state of the limb. (Vector $\vec{\theta}$ contains the limb joint angular positions, vector $\vec{\omega}$ contains the limb angular velocities.) The output of the controller is the torques which are to be applied by the prosthetic servo motors to the artificial limb (vector $\vec{M}_{prosthetic}$). If the strategy which defines the controller equation is correct, the limb system will behave in a manner very similar to the natural limb and will require a minimum amount of retraining by the amputee, i.e., the amputee will not be required to train any muscular control site to perform in an unnatural manner. The controller determines the amputee's desired limb motions by interpreting a set of normally behaving muscle EMG signals.

The use of EMG signals for control of a one degree of freedom powered elbow is relatively straightforward. Since the Biceps-Triceps muscle pair were originally responsible for flexion of the amputee's elbow, it is logical to use those signals for control of the prosthetic elbow. In practice the difference between the average rectified value of the EMG signal (ARV) of the Biceps and Triceps muscles is used as an angular velocity command for the mechanical elbow. Force feedback can be used to render the servo mechanism load sensitive Ref. (7)(8).

Control of a multiple degree of freedom arm such as shown in Figure 1 is a considerably more complicated task. Control of more degrees of freedom requires the monitoring of additional muscle signals. The only additional muscles available for monitoring are located in the shoulder of the amputee and the relationships between their activities and the desired limb motion of the amputee are not obvious. Control of three degrees-of-freedom requires the determination of which EMG signals to monitor, which limb kinematic state information to monitor, and a definition of the functional relationship between the monitored signals and the amputee's desired limb motion.

The most successful attempt to control a multi-axis arm has been achieved by the "Temple Arm" (Ref. 6). The control approach used is pattern recognition and the procedure for its implementation is listed below.

1. Assume the mathematical structure for the controller.
2. Determine quantitative values of parameters within the controller by experimentation with a normal subject (non-amputee).
3. Apply the controller to an amputee in closed loop control of a prosthetic limb.

At the University of Utah we are currently investigating a new technique for prosthetic limb control (Refs. 1, 4). Initial, open loop feasibility trials of the control theory have been very promising. Closed loop experiments utilizing an amputee fitted with the "Utah Arm" are scheduled for fall of 1974. The first experiments will be primarily concerned with the determination of system controllability, system stability, and system repeatability.

III. GENERAL CONTROL THEORY

The control equation computes the torque levels which hetic servos based on measurements of selected shoulder muscle EMG mb. Derivation of the general controller equations is briefly re of the theory is contained in Refs. (1), (4), and (5).

The controller is actually a mathematical expression of n below.

Fundamental Postul

For a given arm prosthesis state ($\vec{\theta},\vec{\omega}$) and f torques applied to the natural joints by the controller should apply torques to the prost cause the clavicle to move in an experimenta with the humerus.

Derivation of the controller equation begins with the motion equations for the arm linkage as shown in Figure 3.

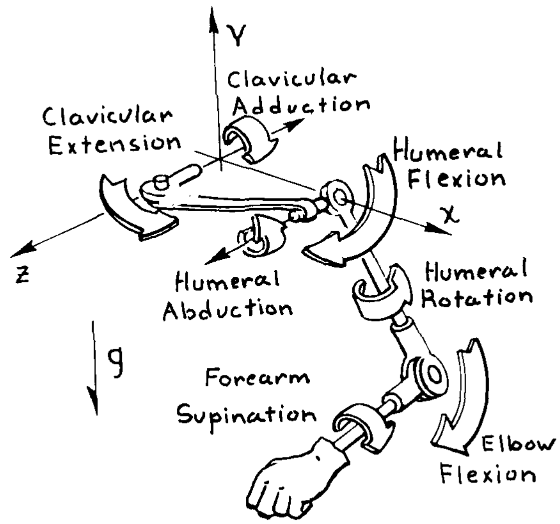

Fig. 3 Degrees of freedom of the natural arm

The equations can be written in matrix form as in Equation 1.

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

$\tilde{P}$, $\vec{Q}$, $\vec{R}$, and $\tilde{S}$ are matrices whose elements depend on the linkage angle vector, $\vec{\theta}$, and angular velocity vector, $\vec{\omega}$. $\vec{\alpha}$ is the angular acceleration vector. $\vec{M}_{all}$ is a vector which contains the torque applied to the joints of the linkage. Note that for the limb shown in Figure 3, some of the joints 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. When the arm reaches up, the clavicle follows slightly. Such an observation may be represented by Equation 2.

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

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 3. Matrix $\tilde{\eta}$ represents the experimentally determined relationship between the clavicle and the remaining arm (humerus and forearm).

If $\vec{\theta}$ and $\vec{\omega}$ and $\vec{\alpha}$ vectors are partitioned into constrainec tioned into subvectors which contain the torque applied to the the prosthetic joints, then Equations 1, 2, and 3 may be combi of freedom as shown in Equation 4.

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

$$\vec{M}_{all} = \begin{Bmatrix} \vec{M}_{prosthetic} \\ \vec{M}_{natural} \end{Bmatrix}, \qquad \vec{\alpha} = \begin{Bmatrix} \vec{\alpha}_{const} \\ \vec{\alpha}_{free} \end{Bmatrix}$$

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

The $\tilde{\beta}_1$ and $\tilde{\beta}_2$ matrices contain the partitioned $\tilde{P}$, $\tilde{Q}$, $\tilde{R}$, and $\tilde{S}$ matrices. The elements of the $\tilde{\beta}_1$ and $\tilde{\beta}_2$ matrices depend on $\vec{\theta}_{free}$, $\vec{\omega}_{free}$, and the parameter matrix $\tilde{\eta}$. Output from the controller equation is the torque to be applied to the prostheses, $\vec{M}_{prosthetic}$. Inputs are the kinematic state of the free portion of the linkage, $\vec{\theta}_{free}$, $\vec{\omega}_{free}$, and the torques exerted on the natural joints by the muscles, $\vec{M}_{natural}$. Section IV will detail the form of the $\tilde{\beta}_1$ and $\tilde{\beta}_2$ matrices for the case of elbow flexion control.

The vector of natural torques, $\vec{M}_{natural}$, must be determined using the EMG signals from selected shoulder muscles ($\vec{E}$). The desired relationship is shown in Equation 5.

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

The submatrices which combine to form the $\tilde{G}$ and $\vec{N}$ matrices are derivable via knowledge of musculoskeletal anatomy, the relationship between muscle tension and electromyographic signals, muscle recruitment patterns, and EMG transmission characteristics.

The torques exerted on the natural joints, $\vec{M}_{natural}$, are related to the tension generated by all of the muscles which act on the joints as shown in Equation 6.

$$\vec{M}_{natural} = \tilde{A}(\vec{\theta})\,\vec{f}_{all} \tag{6}$$

The anatomy matrix, $\tilde{A}(\vec{\theta})$, represents the moment arms of each muscle in vector $\vec{f}_{all}$ on each torque in vector $\vec{M}_{natural}$. Note that the elements of the $\tilde{A}(\vec{\theta})$ matrix can depend on the linkage position vector, $\vec{\theta}$.

The tension developed by the muscles may be related to their EMG signals by the approximate relationship shown in Equation 7.

$$\vec{f}_{all} = \tilde{H}_1(\vec{\theta},\vec{\omega})\,\vec{E}_{all} + \vec{H}_2(\vec{\theta},\vec{\omega}) \tag{7}$$

Vector $\vec{E}_{all}$ represents the average rectified value of the EMG signals from each of the muscles contained in vector $\vec{f}_{all}$. The matrices $\tilde{H}_1$ and $\tilde{H}_2$ are the muscle modeling matrices which depend on the limb kinematic state $(\vec{\theta},\vec{\omega})$.

Since all of the muscles represented in vector $\vec{E}_{all}$ are not independent, it is possible to reduce the number of EMG signals which must be monitored. Specifically, an experimental study yields Equation 8.

$$\vec{E}_{\substack{other\\subcutaneous}} = \tilde{\alpha}_1\,\vec{E}_{\substack{selected\\subcutaneous}} + \vec{\alpha}_2 \tag{8}$$

The matrices $\tilde{\alpha}_1$ and $\vec{\alpha}_2$ represent the relationship between the EMG signals of the muscle group $\vec{E}_{\substack{selected\\subcutaneous}}$ and the group $\vec{E}_{\substack{other\\subcutaneous}}$. The subscript "subcutaneous" indicates that the EMG signals are measured subcutaneously at the muscle site. Therefore, variations in electrode conduction to the skin surface (cutaneous) and cross talk effects may be included by Equation 9.

$$\vec{E}_{\substack{selected\\subcutaneous}} = \tilde{T}\,\vec{E} \qquad)$$

The matrix $\tilde{T}$ is the EMG transmission matrix. Note tha ... ıown below.

$$\vec{E}_{all} = \begin{Bmatrix} \vec{E}_{\substack{selected\\subcutaneous}} \\ \vec{E}_{\substack{other\\subcutaneous}} \end{Bmatrix} \qquad \text{and} \quad \text{if}$$

Combining Equations 6, 7, 8, and 9 results in the vect...

$$\begin{aligned}\vec{M}_{natural} = {} & [\tilde{A}\tilde{H}_{11} + \tilde{A}\tilde{H}_{12}\tilde{\alpha}_1]\,\tilde{T}\,\vec{E}_{\substack{selecte\\cutaneo}} \\ & + [\tilde{A}\vec{H}_2 + \tilde{A}\tilde{H}_{12}\vec{\alpha}_2 \qquad ,.0)\end{aligned}$$

Note that Equation 10 is the desired form of Equation 5. In practice, Equation 5 has been determined by performing a multi-variable linear regression on experimentally obtained data rather than by analysis (Refs. 4 and 5).

Combining Equations (4) and (5), yields Equation (11).

$$\vec{M}_{prosthetic} = (\tilde{\beta}_1 \tilde{G})\, \vec{E}_{\substack{selected\\ cutaneous}} + (\tilde{\beta}_1 \vec{N} + \vec{\beta}_2)$$

or

$$\vec{M}_{prosthetic} = \tilde{\Phi}_1 (\vec{\theta}, \vec{\omega}, \tilde{\tau})\, \vec{E}_{\substack{selected\\ cutaneous}} + \vec{\Phi}_2 (\vec{\theta}, \vec{\omega}, \tilde{\tau}) \tag{11}$$

Equation 11 is the controller equation which contains the effects of factors such as musculo-skeletal anatomy, muscle modeling, muscle recruitment, limb dynamics, and observed normal limb motion.

The next section will utilize the above theory to develop a controller for a one degree of freedom powered elbow.

IV. APPLICATION OF THE THEORY TO CONTROL A POWERED ELBOW

For initial evaluation of the theory, we intend to apply the controller equation to a powered prosthetic elbow. Later studies will include more degrees of freedom.

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 4.

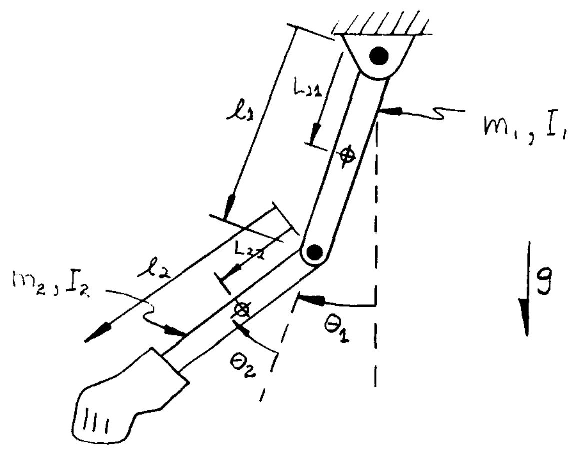

Fig. 4 Coordinate system for derivation of the powered elbow controller

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 monitori
The elbow torque to be applied to the prosthesis, M_2, will be

Determination of the control equation begins with the de e
Equation 1). For the linkage shown in Figure 3, the matrices

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

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

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

$$\tilde{S} = \left\{ \begin{array}{c|c} 1 & 0 \\ \hline 0 & 1 \end{array} \right\} \quad \text{and} \quad \dot{\theta}_1 = \omega_1 , \quad \dot{\theta}_2 = \omega_2$$

$$g_1 = \frac{1}{2}(m_1 L_{11}^2 + I_1) + \frac{1}{2}(m_2(\ell_1^2 + L_{22}^2) + I_2)$$

$$g_2 = m_2 \ell_1 L_{22}$$

$$g_3 = \frac{1}{2}(m_2 L_{22}^2 + I_2)$$

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 Equation (12).

$$\theta_1 = \eta\, \theta_2 \tag{12}$$

Partitioning Equation (1) with the $\tilde{P}$, $\tilde{Q}$, $\tilde{R}$, and $\tilde{S}$ matrices inserted and combining with Equation (12) yields the controller Equation (13).

$$\begin{aligned} M_2 = K\{M_1 &+ (2\eta+1)(g_2 \sin\theta_2)\dot{\theta}_2^2 - (w_1 L_{11} + w_2 \ell_2)\sin(\eta\theta_2) - \\ &- w_2 L_{22} \sin[(\eta+1)\theta_2]\} + \{(\eta^2 g_2 \sin\theta_2)\dot{\theta}_2^2 + \\ &+ w_2 L_{22} \sin[(\eta+1)\theta_2]\} \end{aligned} \tag{13}$$

where

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

For the case where the humerus is to be relatively immobile ($\tilde{\eta}$ is approximately equal to zero), Equation (13) reduces to Equation (14).

$$\begin{aligned} M_2 = \frac{2g_3}{2g_3 + g_2\cos\theta_2} &\{M_1 + (g_2 \sin\theta_2)\dot{\theta}_2^2 - w_2 L_{22}\sin\theta_2\} + \\ &+ \{w_2 L_{22} \sin\theta_2\} \end{aligned} \tag{14}$$

Equation (14) computes the elbow torque, M_2, which, for a given forearm state (θ_2,ω_2) and shoulder torque, M_1, will cause the humerus to remain immobile (θ_1 remains approximately equal to zero).

The shoulder torque, M_1, is determined via the VMG equation (Equation 5). Elements of the $\tilde{G}$ and $\overleftarrow{N}$ matrices are assumed to be constants. Numerical values for $\tilde{G}$ and $\overleftarrow{N}$ are determined by performing a multivariable linear regression on experimental data obtained from an amputee. Data is obtained by simultaneously recording muscle EMG's and shoulder torque outputs while the experimental subject performs specified tasks. References (1), (4), and (5) further detail the procedure for obtaining the $\tilde{G}$ and $\tilde{N}$ matrices.

Computation

With Equations (5) and (14) determined, the controller may be implemented by a digital computer. The procedure will be as follows:

1. The EMG signals from up to 9 shoulder muscles of the amputee will be monitored. Using analog circuitry the raw EMG signals will be rectified and filtered to produce a voltage which corresponds approximately to the tension generated within each muscle (the voltage is the average rectified value of the EMG, ARV). Preliminary experiments indicate that the number of signals monitored might be reduced to as low as 4 or 5.

2. The processed EMG signal, vector $\overleftarrow{E}$, will then be sampled by the digital computer at at a rate of from 20 to 70 samples per second on each of the channels. Additional numerical filtering of the signals may be accomplished digitally if desired. The elbow position and velocity, θ_2, ω_2 will also be monitored and sampled by the digital computer at the same rate as the EMG channels.

3. At a rate from 20 to 50 times per second, the input data vector, $\overleftarrow{E}$, and θ_2, ω_2, will be used to compute M_1 via Equation (5) and then to compute M_2 via Equation (14). M_2 will then be converted to an analog signal and applied to the torque servo located in the elbow of the prosthesis. Depending on the filter requirements, the M_2 command will be refreshed at a rate of from 20 to 50 times per second.

Once the arm, the amputee, and the digital computer are operational as a system, extensive studies will be conducted to evaluate performance as a function of various system modifications. Of particular interest will be:

1. System stability
2. System controllability
3. Minimal permissible monitoring sample rates and minimal permissible servo refresh rates
4. The effect of variations in the motion constraint parameter $\tilde{\eta}$.
5. The effect of assuming that certain of the variable elements in the $\tilde{\beta}_1$ and $\overleftarrow{\beta}_2$ matrices are constant.

The next step will be to apply the control procedure to limbs with additional degrees of freedom. The ultimate goal of the project is the development of an artificial limb which contains a micro-processor for control of the degrees of freedom, humeral rotation, elbow flexion, forearm rotation, and terminal device closure.

SUPPORT AND COLLABORATION

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