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On the dynamic control of myoelectric powered prostheses for the disabled

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Abstract

Dynamic properties of the man–machine system must be considered while designing human–prosthesis systems. In this paper, position control during arm manipulation is discussed. A bilinear mechanism of the musculoskeletal system is proposed to assist the amputee in controlling the position of a myoelectric powered arm. It is shown that if a bilinear structure, such as the one proposed, is added to the human–prosthesis interface the arm position control can be substantially improved.

Introduction

It is generally agreed that an artificial limb should simultaneously satisfy the following conditions (Jabobson, 1982):

(a) easy and natural control,
(b) cosmetically acceptable,
(c) light, quiet, and durable,
(d) efficient enough for compact energy storage,
(e) inexpensive.

Except for the first factor, the above list is primarily hardware oriented. However, "controllability" depends largely on how to design the interface between the amputee and the prosthesis. A simple block diagram, Figure 1, illustrates an example of the interface of multiple degree-of-freedom myoelectric prosthesis. The cutaneously measured electromyographic (EMG) signals from residual muscles are used to identify the motions that the amputee intends to begin, and to select a set of prosthesis functions, e.g., wrist flexion/extension, forearm pronation/supination and the like. In parallel, the muscle force is estimated from the same EMG signals driving the prosthesis. On the other hand, the position, velocity, and force informations from the pros-

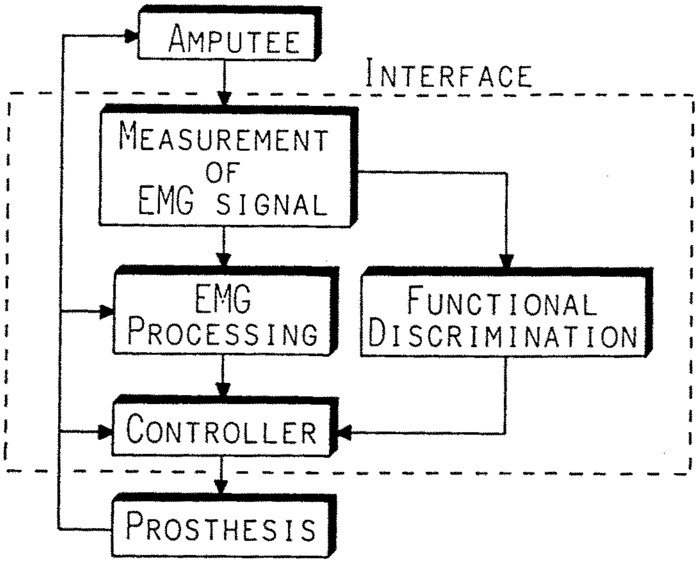

Figure 1. Human–prosthesis interface.

thesis are returned to the amputee through visual or tactile sensation, or electrical stimulation.

The human-limb system includes a highly integrated network of multi-level controllers, actuators and information receptors, connected by efferent and afferent pathways, which allows the musculoskeletal system to modulate a wide variety of dynamic behaviours. Unfortunately, the amputee may have lost not only the abilities of the musculoskeletal system as the effector organs, but also the physiological mechanisms which provide him/her with locomotion, postural control and manipulative skills. Since the information exchange in human–prosthesis system has to be done through residual functions, it is difficult for the amputee to control the prosthesis unless some kind of compensatory mechanism is introduced into the interface design.

In general, the myoelectric prostheses available today adopt on-off control or proportional control. The former uses the myoelectric signal only to turn on or off the actuator of the prosthesis. The latter uses the processed myoelectric signal directly as command signals of the actuator. The lock-unlock control is accomplished by high and low levels of contraction of the antagonist muscles or by a third signal acquired from a motion switch (Jacobson, 1982). The ultimate goal of the prostheses research is to develop artificial limbs which are controlled naturally by the amputee's motor intents and are functionally responsive (like the natural human limbs). Progress in prostheses requires a more intimate cybernetic interface between the amputee and the artificial limb, and a clearer understanding of the neuro-muscular-skeletal system which controls the natural limbs.

The purpose of this paper is to report on the problems of improving the control aspects in human–prostheses systems. Figure 1 suggests that the amputee and the limb prosthesis should be regarded as a man–machine control system where the controlled element is the powered prosthesis. Therefore, when designing the human–prosthesis system, it is necessary to pay attention to the dynamic properties of the man–machine system as a whole. Force, position, velocity and impedance can be used in such a system as the controlled variables. Here, only the position control, which is one of the most basic problems in arm manipulation, is discussed. It is proposed that a bilinear mechanism of the musculoskeletal system be built into the human–prosthesis interface so that the amputee can change the dynamic characteristics of the powered prosthesis continuously by his intents. Consequently, the myoelectric powered arm can be simulated on the computer, and the position control through the surface EMG signal can be evaluated from the viewpoint of the dynamic properties. Finally the tracking tasks can be analysed experimentally in order to predict to what degree the amputee's controllability can be improved in the position control.

Bilinear interface

The skeletal muscle of the human body is activated by impulses of the motorneurones. The relationship between the nerve impulses and the contractile force of a muscle constitutes an ingenious and a very complex mechanism. It is well known, however, that the macroscopic mechanical properties of the muscle can be represented using two fundamental functions of length–tension curves and force–velocity curves (Dowben, 1980).

Assuming that the muscle force is in proportion to the level of activation ($0 \leqslant \alpha \leqslant 1$; normalized by the maximum), the muscle force F can be given by:

$$F = \alpha \cdot g(L, V) \tag{1}$$

where $g(L, V)$ is a nonlinear function under the maximum level of activation. Approximating $g(L, V)$ by the Taylor expansion around rest length $L = l_0$ and the velocity of contraction $V = 0$, and neglecting the second and higher terms, the following linear relation can be obtained:

$$g(L, V) = f_0 - k_1 x - b_1 \dot{x} \tag{2}$$

where f_0 is the maximum tension at isometric contraction ($V = 0$), x is the relative length of muscle ($x = 0$ at rest and $x > 0$ is shortening), $\dot{x}$ is velocity of shortening, and k_1 and b_1 are positive constants. Substituting (2) into (1) yields:

$$F = u - k' u x - b' u \dot{x} \tag{3}$$

where $u = \alpha \cdot f_0$, $k' = k_1/f_0$, and $b' = b_1/f_0$. This is the visco-elastic model of a muscle, where the viscous and elastic coefficients are not constant but are in proportion to the contractile force u.

By assuming that the forearm and hand can be regarded as a rigid link rotating about a fixed axis, the flexor and extensor have the same properties, and the moment arm d is angle-independent, the muscle torques T_f and T_e about the joint are given by:

$$T_f = d\,(u_f - k u_f \theta - b u_f \dot{\theta}) \tag{4}$$

$$T_e = d\,(u_e + ku_e\theta + bu_e\dot{\theta}) \tag{5}$$

where $k = k'\,d$, $b = b'\,d$ and θ is defined as 0 in the right angle for the upper arm and is positive toward flexion. u_f and u_e are the contractile forces of the flexor and extensor.

Based on the above, the dynamic model equation for the horizontal movements of the forearm can be obtained as follows:

$$I/d \cdot \ddot{\theta} = u_f - u_e - (u_f + u_e)k\theta - (u_f + u_e)b\dot{\theta} \tag{6}$$

where, I is the moment of inertia. The visco-elastic linear model proposed hitherto is as follows:

$$I/d \cdot \ddot{\theta} = u_f - u_e - k\theta - b\dot{\theta} \tag{7}$$

Equation (7) is based on the assumption that the stiffness and viscosity of the muscle are invariant and independent of the contractile forces. On the other hand, Equation (6) represents the fact that the driving torque about the joint and the stiffness and viscosity about the joint can be controlled independently through the sum and difference of the controlled forces, respectively. Since the difference $u_f - u_e$ controls the input, while sum $u_f + u_e$ controls the system parameters, Equation (6) is nothing but a bilinear system. Thus, one of the important phenomena of the natural limb is that the parameter adaptation can be performed by co-activation of antagonistic groups of muscles without proprioceptive feedback. The control advantages of the bilinear structure are given in Hogan (1980) and Ito and Tsuji (1985).

If one adds the bilinear structure to the human–prosthesis interface, the amputee might be able to adjust not only the driving torque, but also the mechanical impedance about the joint of the prosthetic arm. For example, when starting to move the forearm toward the desired position, it will be more desirable to make the mechanical impedance as small as possible, because it becomes easy to move the arm. But conversely, when bringing the arm to rest, it is desirable to make the impedance large to brake the arm. Thus, the prosthetic arm with the bilinear interface performing like the natural limb would give an amputee an essential component of the natural adaptive capability despite the severe sensory loss due to amputation. The effects of bilinear interface on the amputee's control manners are discussed in the next section.

Position control

Experimental procedures

The experimental arrangement for position control is shown in Figure 2. The subject's forearm was fixed on the horizontal table by keeping the elbow at a right angle. The EMG signals were taken from biceps and triceps under isometric contraction. The pair of differential electrodes on each muscle were 15 mm diameter discs placed 20 mm apart. After full wave rectification the EMG signals were processed by a couple of low-pass analogue filters ($f_c = 1$ Hz, the first-order). The filtered signal, called integrated EMG, is proportional to the muscle force (Hogan and Mann, 1980). The filtered outputs are the command signals u_f and u_e.

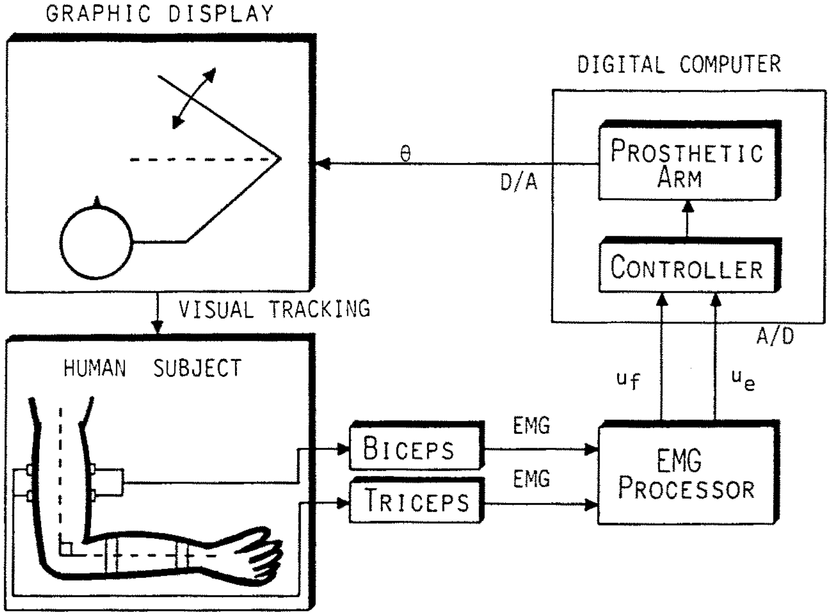

Figure 2. Experimental arrangement for position control.

The forearm was drawn on the graphic screen. The solid line represented the current position and the broken line, the desired position. The computation time for the controller and mathematical model of the prosthesis was insignificant. The forearm shown on the computer display was rotated corresponding to the output (joint angle) θ from the computer. Therefore, the human subject was able to rotate the forearm with the aid of the EMG signals generated by biceps and triceps.

The dynamic equation, which simulated the horizontal movement about the elbow joint of a prosthetic arm, is given as follows:

$$I\ddot{\theta} + B_J\dot{\theta} = T \tag{8}$$

where

I : the moment of inertia.
B_J : the coefficient of viscosity about the joint.
T : the input to the actuator.
θ : elbow joint angle, ($\dot{\theta}$ = angular velocity; $\ddot{\theta}$ = angular acceleration).

Two interface controllers were tested. The first one was bilinear as shown in Figure 3; the second one was linear, shown within the broken line. In the bilinear controller, the driving force and system parameters can be controlled independently through the difference and sum of the flexor and extensor.

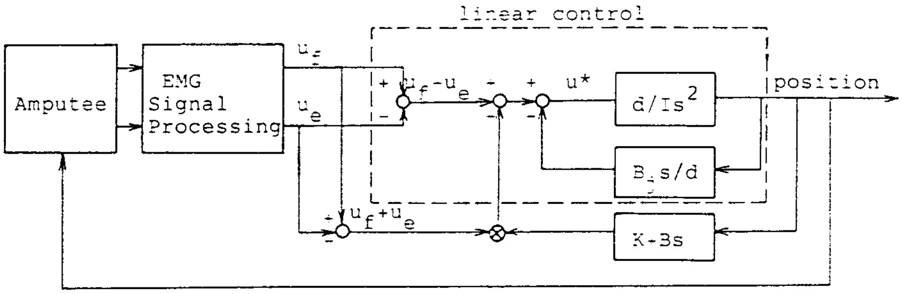

Figure 3. Interface controllers for position control.

Arm positioning control

Some computer-controlled tracking tests were performed. The human subject sat in front of the graphic display and was instructed to move the forearm on the display from the current position to the desired position as fast as possible. The step responses were superimposed (Figure 4). The solid lines denote the results for the bilinear system; broken lines denote results for the linear system. The parameters were set at $d/J = 2.0$, $B_J/d = 0.4$, $K = 0.1$, $B = 0.2$. Figure 4(a) shows the time responses of the joint angle (which was the controlled variable). It can be seen that the rising time of the bilinear system is shorter than that of the linear system. u_f (flexor) and u_e (extensor) are shown in Figure 4(c). The control of the linear system is a bang-bang form. First, the flexor accelerates the forearm and then the extensor is activated to decelerate it, synchronizing with the relaxation of the flexor. On the other hand, in the bilinear system, co-activation of the flexor and extensor is observed in the latter half. Since co-activation gives an increase of $u_f + u_e$ and consequently makes the viscous coefficient larger, the prosthesis control system can be switched to a system with larger damping. This shows that the human subject skilfully utilizes the variable structure of the bilinear system.

The input u^* to the inertia element d/Is^2 makes the difference of both position controls clearer. As shown in Figure 4(b), both inputs are bang-bang forms. However, it is seen that the linear system yields loose switching and small amplitude, while the bilinear system yields sharp switching and large amplitude, i.e., about three times as large as the linear one. This leads to the differences in speed of response between both systems.

Linear systems never allow the control to change the system parameters. Therefore, when the large input is added to get a short rising time, it becomes more difficult to predict the switching point, which will lead to the overshoot. In contrast, the multiplicative control of bilinear systems can be used to change the system to a large damping system at any time, which may result in large input in the first half.

In the next phase of the experiment, the desired positions were shifted by angles (within 80 degrees) and time spans (4–8 seconds) supplied by the random number generator of the computer. Duration time was 50 seconds and three normal subjects

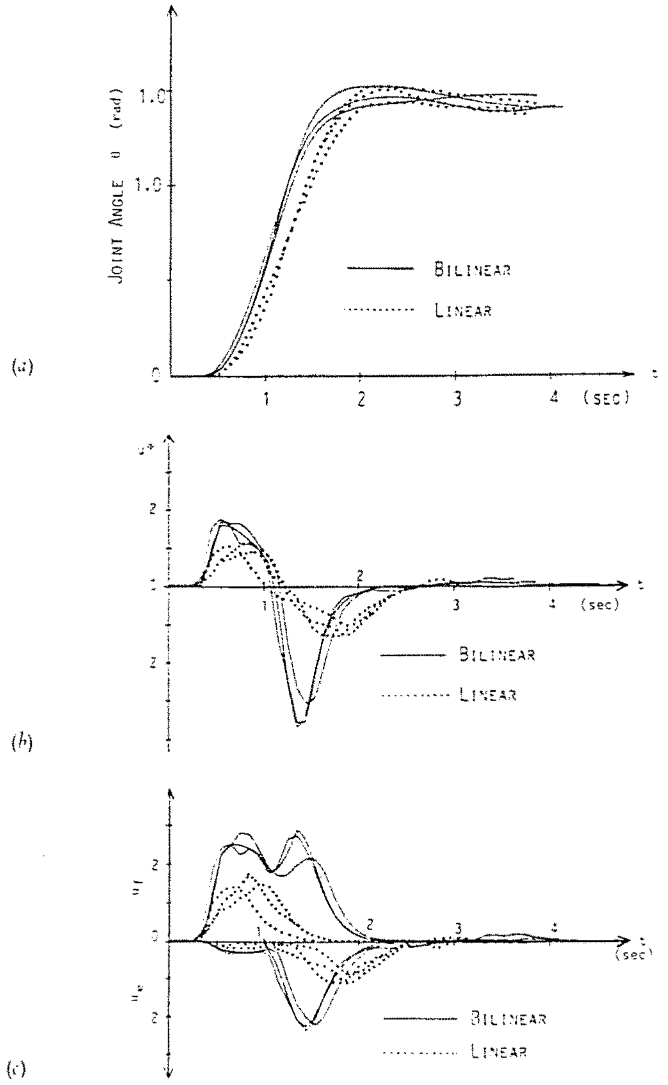

Figure 4. Step responses:
(a) Time response of the joint angle;
(b) Input (u).*
(c) Response of flexor (uf) and (ue);

were trained to a stable level of performance scores. In general, at least twenty trial runs were carried out before the runs were recorded.

Figure 5 shows time scores of: (a) linear and (b) bilinear system, respectively. From the top, r denotes reference input, θ the joint angle, EMG_f the myoelectric surface signal of the flexor, u_f the filtered EMG signal, EMG_e the myoelectric surface signal of

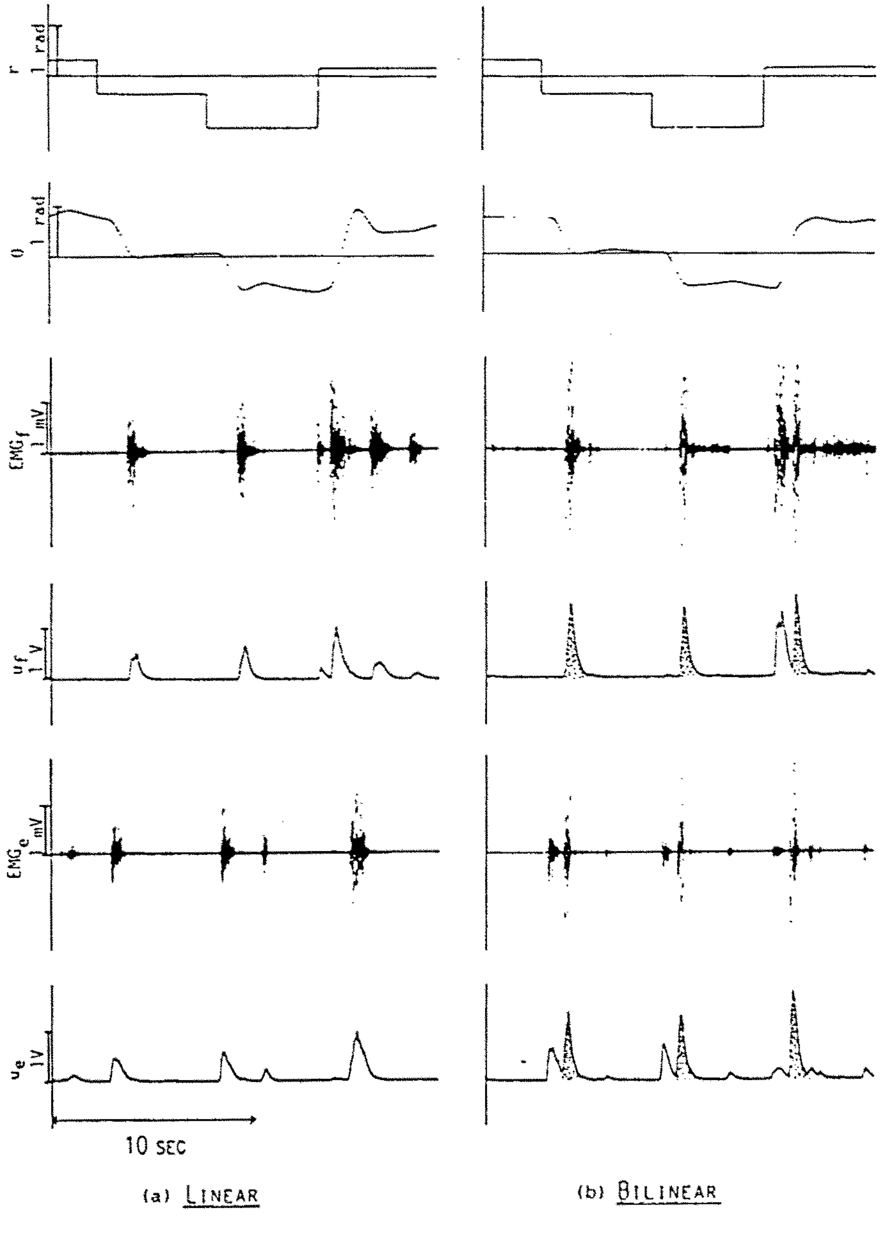

Figure 5. Tracking time scores of (a) linear and (b) bilinear systems for random step inputs.

the extensor, and u_e the filtered EMG signal. It can be seen that the control patterns between both systems exhibit striking contrasts. In the linear system, the flexor and extensor are activated reciprocally. In contrast, the bilinear system allows the subject to co-activate both muscles.

The following performance scores were also computed;

1) Integrated squared error

$$ISE = \sum_{i=1}^{N} [\int_0^{T_i} e_i^2(t)dt / \int_0^{T_i} r_i^2(t)dt] \tag{9}$$

2) Integrated absolute error

$$IAE = \sum_{i=1}^{N} [\int_0^{T_i} |e_i1(t)|dt / \int_0^{T_i} |r_i(t)|dt] \tag{10}$$

3) Integrated time absolute error

$$ITAE = \sum_{i=1}^{N} [\int_0^{T_i} t|e_i(t)|dt / \int_0^{T_i} t|r_i(t)|dt] \tag{11}$$

4) Integrated control cost

$$ICC = \sum_{i=1}^{N} [\int_0^{T_i} (u_{fi}(t) + u_{ei}(t))^2 dt / \int_0^{T_i} |r_i(t)|dt] \tag{12}$$

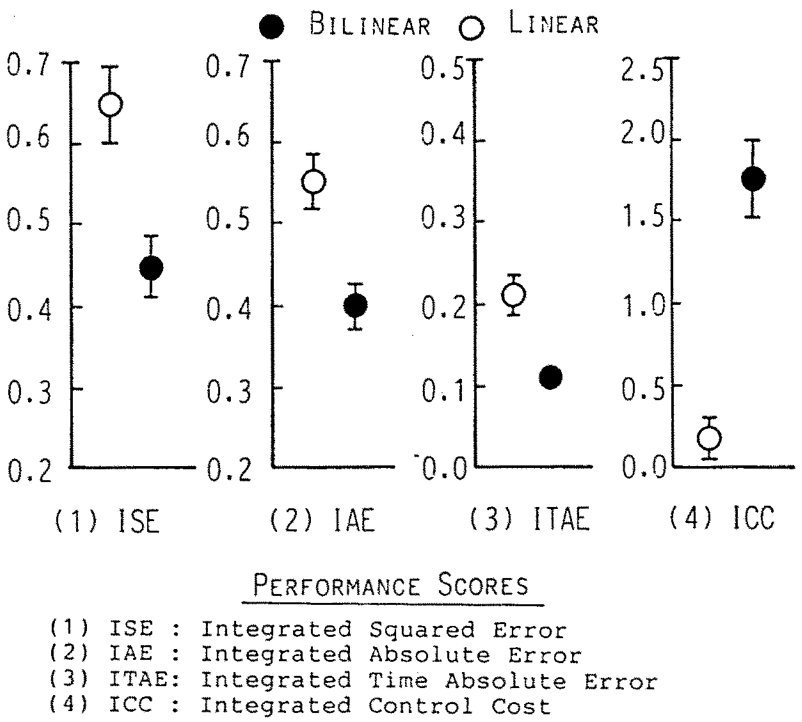

Figure 6. Performance scores.

The averaged scores and standard deviations of ten trials after training are shown in Figure 6. Three scores (ISE, IAE, ITAE) of the bilinear controller, except ICC, were lower by 30 per cent–50 per cent than the linear controller. This suggests that the bilinear controller can improve the amputee's position control. Although the control cost (ICC) increased substantially, the subject's EMG levels did not exceed 40 per cent of the maximum voluntary contraction.

Continuous arm movements

It may be concluded that the bilinear controller is not a good one for continuous and smooth arm movements. To examine this assumption, another tracking test was performed using the continuous and smooth inputs. The experimental conditions were the same except for the input and display.

The reference input was a random-appearing signal composed of ten sine waves of arbitrary phases and different frequencies in the range of 0·309 to 9·865 rad/sec ($w_c = 2\cdot265$ rad/sec). The visual display was an oscilloscope. The controlled element, which simulated the horizontal movement about the elbow joint of a prosthetic arm, is given in Equation (8). The parameters were given by $d/J = 2\cdot0$, $B_J/d = 0\cdot4$, $K = 0\cdot1$, $B = 0\cdot2$. The subjects, three healthy persons (non-amputees), were instructed to keep the error as small as possible. Each subject was trained to reach a stable level again.

The tracking time scores for the bilinear system are shown in Figure 7. Few co-activations of the flexor and extensor were observed. Such results are much different from that depicted in Figure 5. It was not necessary to stop the prosthetic arm rapidly since the input was continuous and smooth.

The frequency characteristics for the above experiment were computed using the following relations:

$$G(jw) = P_{rc}(jw)/P_{re}(jw) \tag{13}$$

where G(jw) is the open-loop frequency responses of the overall system, and $P_{xy}(jw)$ is the cross-spectral density function between x and y and estimated by FFT method. The open-loop frequency responses from the error e to the controlled variable θ for both systems are shown in Figure 8. These are averages and standard deviations of ten trials. There is a little difference between both systems in terms of the amplitudes and phase characteristics.

The transfer characteristics in normal manual control with visual feedback was summarized as the crossover model and represented as follows:

$$G_h(s)G_c(s) = w_c e^{-\tau_e s} \tag{14}$$

where $G_h(s)$ denotes the human operator, $G_c(s)$ denotes the controlled element, and the crossover frequency w_c is equivalent to the loop gain and lies between 3–5 rad/sec. The effective time delay τ_e is an accumulation of additional lags due to the transport delays and the high frequency neuromuscular dynamics, and lies between 0·2–0·4 seconds. The crossover model suggests that the subject adapts himself to the variation of the controlled elements by adjusting his own transfer characteristics.

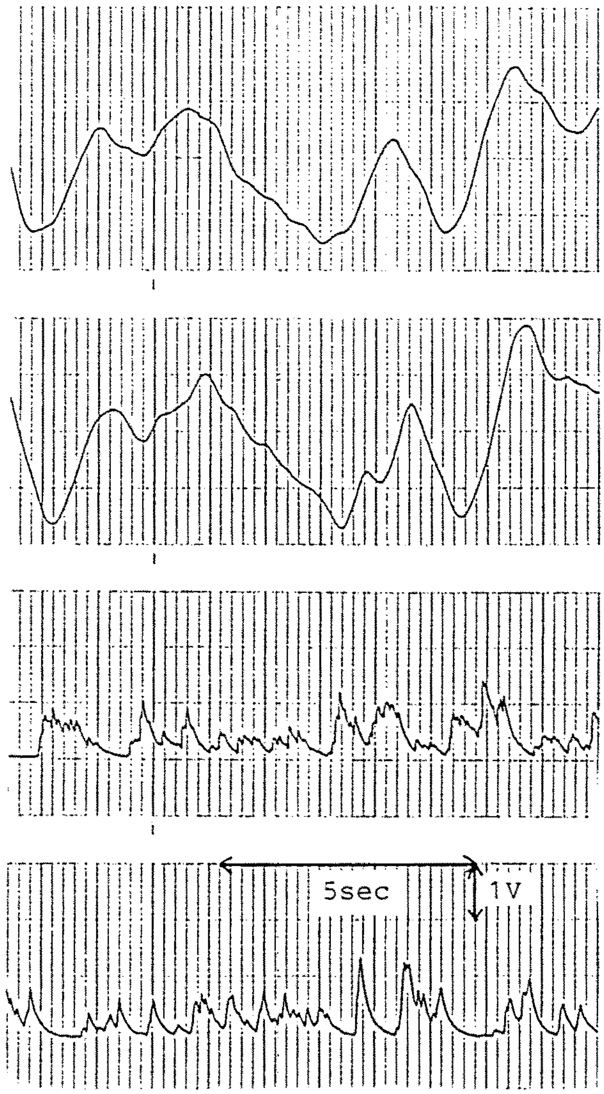

Figure 7. Tracking time scores of bilinear systems for continuous random-appearing inputs.

The solid line in Figure 8 indicates the crossover model. The frequency responses in myoelectric control can be fitted by the crossover model within the range near the crossover frequency w_c (5 rad/sec). From this, it can be seen that tracking characteristics of the myoelectric command signal with a bilinear controller can be maintained at almost the same level as in the manual control.

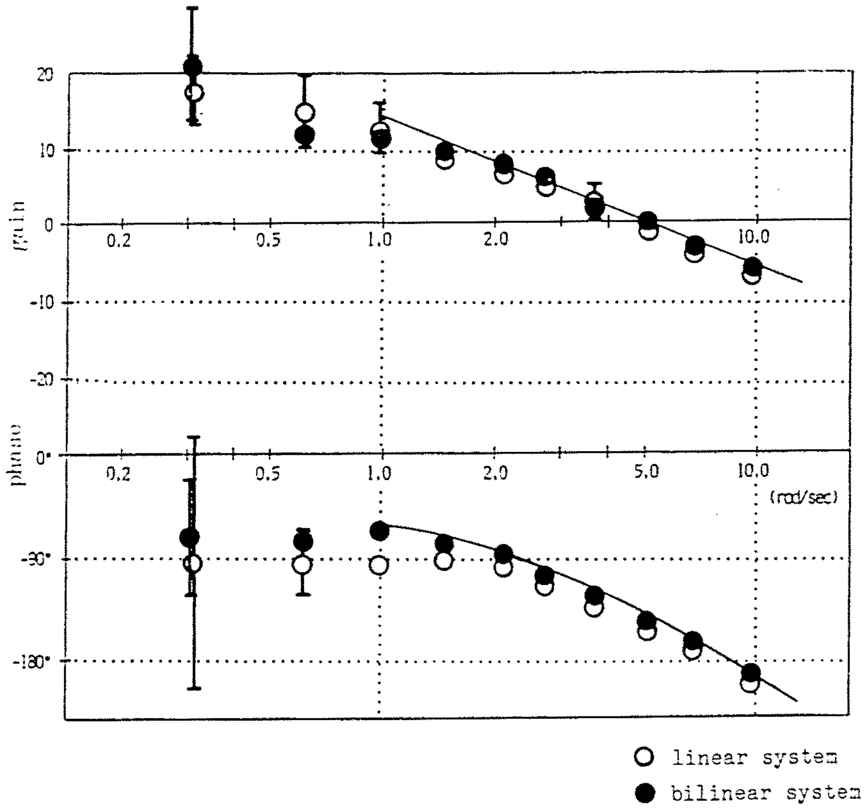

Figure 8. Open-loop frequency responses for linear and bi-linear systems.

Conclusion

It is an important property of the bilinear system that the control input is able to vary the system structure, where no feedback compensatory loop to the amputee is added. Modification of viscoelastic property about the joint by co-activation of the antagonist muscles increases the flexibility of the system and plays a significant role in facilitating the open loop control of movement (Houk, 1979; Hogan, 1984). In this study, it was shown that the position control could be improved substantially by adding the bi-linear structure to the present interfaces in human-prosthesis system.

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