DESIGN OF FEEDBACK FNS STANDING SYSTEM

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ABSTRACT

The new feedback control system is proposed for the stabilization of paraplegic standing with functional neuromuscular stimulation (FNS). We used the data, from the kinematic analysis of the inverted pendulum model of human skeletal system, for the controller parameters determination. In this research, we considered only, the motion in sagittal plane. The parameters for PID controller determined from the model analysis were compared with parameters estimated experimentally. The parameters obtained, showed good coincidence. This complex controller design and comparison of parameters were done in four normal subjects.

KEY WORDS: FNS, Standing, Feedback control, Human upright posture, Spinal cord injury

INTRODUCTION

Regaining the standing and ambulation are important elements for spinal cord injury (SCI) patients. Although, the motor restoration in SCI, stroke and upper motor neuron handicapped humans may be achieved, there are many questions to be answered. The nonlinear and time varying muscle properties are one of complex problems to be solved. The feedback control scheme has been applied to obtain repeatable and accurate muscle contractile force for different joint positions [1 - 8]. Now, the application of feedback system for standing or ambulation is getting more effective for the repeatable and reliable lower extremity functions.

In thoracic or lumbar complete lesions, there is a complete absence of motor and sensory functions in legs. The stimulation of quadriceps, gluteus maximus and other lower extremity muscles provides standing of a paraplegic. However, external disturbances (upper extremity movements, fop example) are not taken into account with existing control systems, resulting with unsafe standing. The use of FNS system combined with external orthotics and feedback using sensory data from external joints providing long time stable standing and walking if paraplegics is presented [9 -13]. The performance of the feedback control system depends on the parameters. Most of presented studies includes the description of the feedback system, but omits parameter selection, decision protocol and optimization procedure. This study considers the parameter determination for standing control based on the analysis of the inverted pendulum model.

POSTURAL CONTROL MODEL ANALYSIS

Human musculo-skeletal system in sagittal plane, as approximated as an inverted pendulum, is unstable system and requires the stabilization with an feedback control system. Several sensory organs are used for stabilization of upright posture in healthy humans. These information are integrated with the central nervous system (CNS). Figure 1 presents the adopted model of human musculoskeletal system for standing. The knee and hip joints are fixed in fully erected position and ignored in the analysis and experiments in normal subjects.

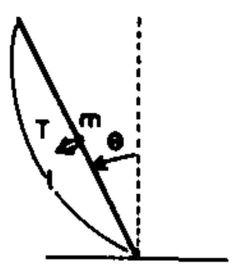

Figure 1. Inverted pendulum model of human standing posture

Assuming that the sway of the inverted pendulum is small, the motion can be described with (1)

$$\frac{1}{3}ml^2\,\Theta'' - \frac{1}{2}mgl\Theta = T \tag{1}$$

where

m : is mass,

Θ : is the ankle joint angle,

l : the height and

T : ankle joint torque.

The transfer function $G(S)$ can be directly obtained from (1) in form

$$G(S)=\frac{\Theta(S)}{T(S)}=\frac{K_s}{S^2-a^2} \quad (2)$$

where

$$a=\sqrt{3g/2l}$$

$$K_S=3/m\,l^2$$

The impulse response of this system is given by inverse Laplace transformation

$$\Theta(t)=K_s\,(e^{at}-e^{-at})/a^2 \quad (3)$$

We assumed the position and velocity feedback system. The sway (ankle) angle and its derivative are these feedback signals (Figure 2). The feedback controller transfer function is

$$C(S)=K_p+K_d\,S \quad (4)$$

where

K_p : is proportional feedback gain and

K_d : velocity gain.

Adopting that K_m is the muscle gain, the complete transfer function is

$$W(S)=\frac{K_s\,K_m}{S^2+K_s\,K_m\,K_d\,S+(K_s\,K_m\,K_p-a^2)} \quad (5)$$

The equation (5) shows that if the optimal parameters are determined the standing posture is stable second order delay system. The derivative K_d, for velocity feedback, is indispensable for the stabilization. The absence of K_d results with the oscillatory behaviour of the system.

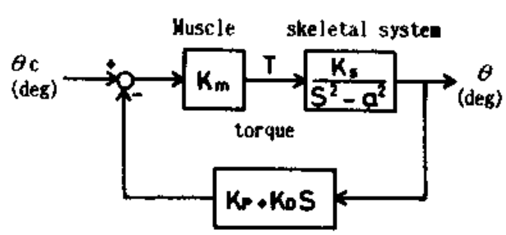

Figure 2. The block diagram of voluntary standing control system

DESIGN OF FEEDBACK FNS STANDING SYSTEM

The possibility of PD feedback posture control by FNS was explained. We will demonstrate the design procedure of the controller. We consider the action of ankle flexors and extensor only. Muscle model includes its viscosity and elasticity. We adopted the subject model as an second order delay system described with:

$$\frac{1}{3} ml^2 \Theta'' + B\Theta' + K\Theta - \frac{1}{2} mgl\Theta = T \tag{6}$$

where

B : is the moment due to ankle joint viscosity, and

K : is the moment due to elasticity.

This equation, using Laplace transformation, is

$$G(S) = \Theta\frac{(S)}{T(S)} = \frac{1}{JS^2 + BS + K - mgl/2} \tag{7}$$

where

$$J = m\,l^2/3$$

We assumed that the active joint torque is obtained by one linear actuator which is the combination of Tibialis anterior m. and Soleus m. The coactivation modulation is assumed to compensate the dead band ahead of the stimulation. We assumed the muscle force as a first order delay system described with:

$$M(S) = \frac{K_m\, e^{-DS}}{1+\tau S} \tag{8}$$

where

K_m : is the muscle gain,

τ : is the time constant of muscle actuator and

D : dead time.

The controlled subject (including actuator, skeletal system and muscle dynamics) is described with

$$G_W(S) = \frac{180/\pi K_r K_m\, e^{-DS}}{\left(1 + \tau S\right)\left(ml^2/3S^2 + BS + K - mgl/2\right)} \tag{9}$$

where

K_r : is the conversion from radians to degrees.

As mentioned before, a proportional and derivative feedback components are necessary for stabilizing the standing. We introduced PID controller as shown in Figure3. The optimum feedback gain of this system expected to offer the stability of the standing posture. The whole system transfer function gets the form

$$W(S) = \frac{G_W(S)\, I(S)}{1 + G_W(S)\, I(S)\, F(S)} \tag{10}$$

where

$$F(S) = K_p + K_d\, S \tag{11}$$

$$I(S) = \frac{K_i}{S} \tag{12}$$

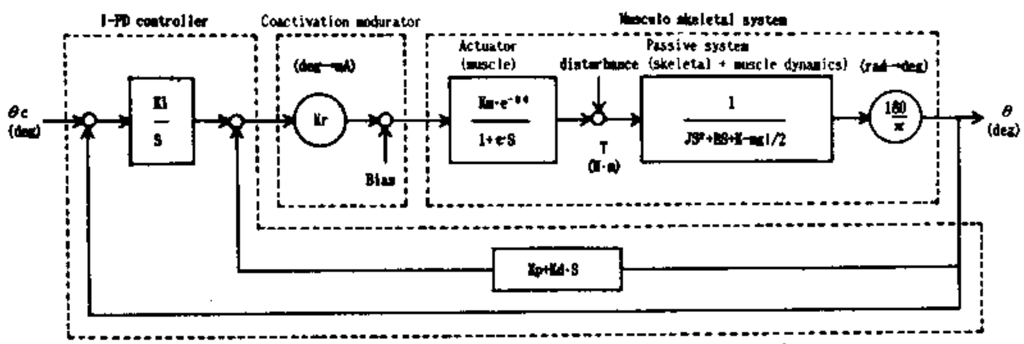

Figure 3. The block diagram of FNS standing control system

The feedback parameters can be determined by a mode matching method [14]. Expanding the dead time of muscles using Maclorin expansion method, the system transfer function (including PID controller) is

$$W(S) = \frac{1}{h_0 + h_1 S + h_2 S^2 + h_3 S^3 + \ldots} \tag{13}$$

Assuming that desirable characteristic is given as

$$R(S) = \frac{1}{r_0 + r_1 \sigma S + r_2 \sigma^2 S^2 + r_3 \sigma^3 S^3 + \ldots} \tag{14}$$

where

σ : is the time constant of the desirable model response.

The feedback parameters can be determined, one by one, from lower order parameter to higher one by adjusting *W(S)* to *R(S)*. Higher order components above three are considered neglectable.

HARDWARE AND SOFTWARE SYSTEM

We applied the eight channel stimulator, shown in Figure 4. Intel 80286 computer is used for the control with a numeric coprocessor and D/A converter (Figure 5). The amplitude was varied between 0 and 100 mA, the interpulse interval was 40 ms, and pulse width was modulated from 0 to 300 μs. The pulse modulation was realized using 0 to 3 V over D/A converter. We used carbon rubber surface electrodes from 20 to 40 cm^2. The electrode impedance is 100 Ω. The potentiometer was mounted at the ankle joint of the ankle-knee-hip brace (Figure 6).

The control program is written in assembler. The object module size is 4 Kb. The interrupt from programmable IC offers 20 ms sampling. The joint velocity is obtained through 1 or 3 Hz third order Butterworth low pass filter using ankle angle measurements.

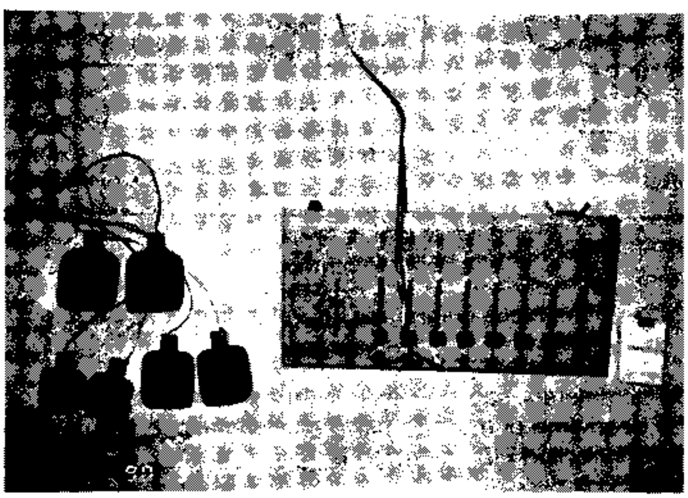

Figure 4. The eight channel electronic stimulator and conductive rubber surface electrodes

Stimulus pulse was modulated by a coactivation modulator. The modulator characteristics shown in Figure 7 is determined by threshold, maximal and balanced stimulating points measured previously. The balanced stimulating point means that zero torque is applied while standing, i.e. agonist and antagonist forces are balanced. The use of this modulator allows the linear approximation of muscle gain.

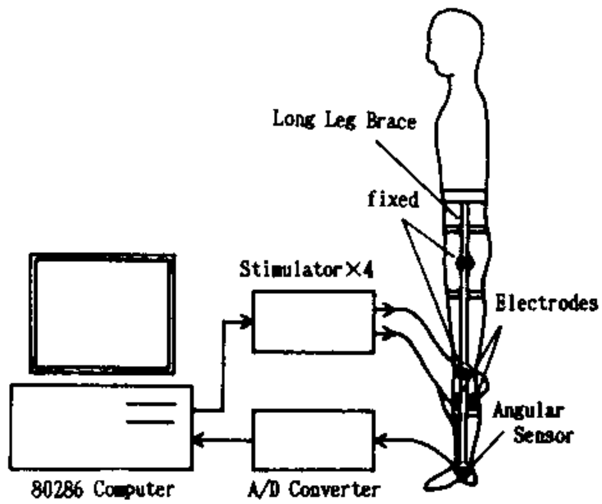

Figure 5. Experimental setup of an FES system for standing control. Hip and knee joints are fixed with an ankle-knee-hip brace. Stimulation parameters were varied from 0 to 100 mA (amplitude), inter pulse interval 40 ms and pulse width from 0 to 300 s.

Figure 6. The ankle-knee-hip brace during experiments in normal subjects. Potentiometers are fixed in all joints.

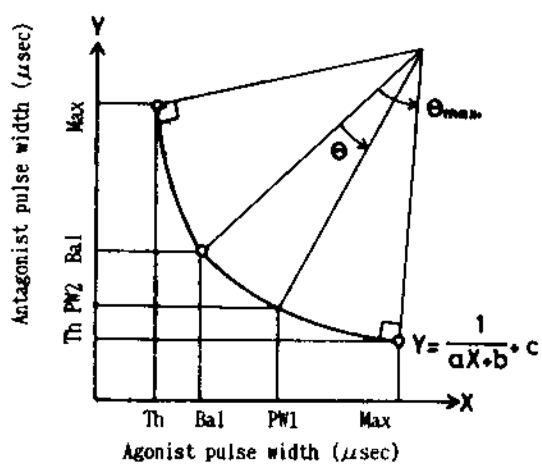

Figure 7. Coactivation modulator for the approximation of agonist and antagonist muscles to a single linear actuator. *Th* - threshold, Bal - balanced stimulating point, *Max* - maximal pulse width, Θ- input command from PID controller, *PW1* - agonist pulse width, *PW2* - antagonist pulse width

EXPERIMENTAL RESULTS AND DISCUSSION

The experimental determination of optimal feedback parameters was done by comparing these parameters with the analytical results. Experiments were carried out in four normal subjects, aged from 21 to 23 years.

Experimental estimation of control parameters

The threshold, maximal and balanced pulse width of Tibialis anterior m. and Soleus m. were measured to determine the characteristics of the coactivation modulators. An external disturbance was included by pushing the back of the subject while FNS postural control system was turned on with fixed knee and hip joints by the ankle-knee-hip brace. Parameters K_p and K_d were determined to compensate the external disturbance. The parameter K_i was determined to minimize the steady state error after optimal K_p and K_d were obtained.

Figure 8 demonstrates results with the proportional controller. The pointing arrow presents the external disturbance, i.e. the push. The body leaned forward after the initial push (a). The feedback gain is smaller compared to the calculated optimal value. The increase of the feedback gain resulted in oscillation (b and c). The optimal value was not found by variation of the parameter K_p. These results proves that proportional control is not enough for the stable posture.

The result of the application of PD controller is shown in Figure 9. (a) and (b) resulted in to small or to high feedback and failure of stabilization of the posture. When the gain was to low, the body leaned forward, while with a high gain backward leaning

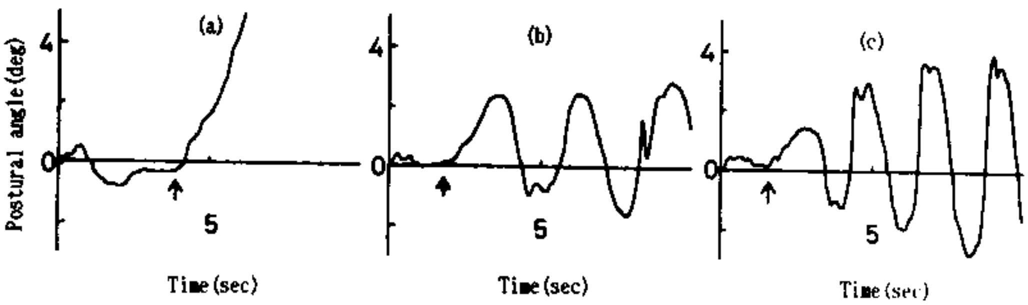

Figure 8. Experimental results with the use of P (proportional) controller in normal subject. Pointing arrow shows the moment of external disturbance. Values for control parameters are: a)K_p = 1; b) K_p = 5 and c) K_p = 10.

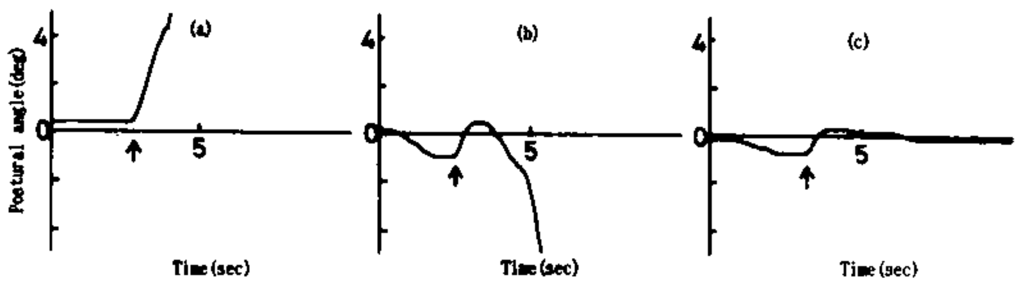

Figure 9. Experimental results with the use of PD controller in normal subject. Pointing arrow shows the moment of external disturbance. Values for control parameters are: a)K_p =0.1, K_d =0.1; b) K_p =2, K_d =10 and c) K_p =6, K_d =5.

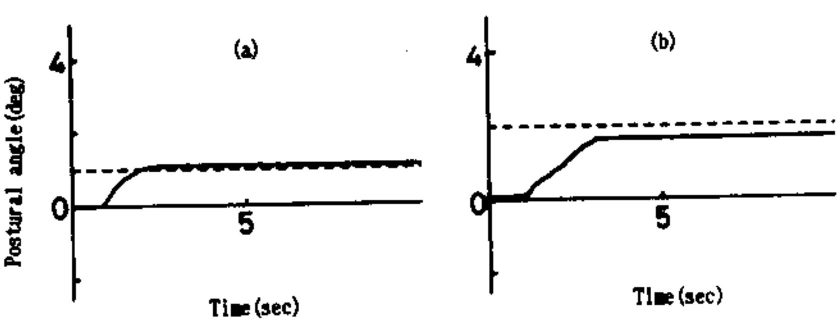

Figure 10. Experimental results with the use of PID controller in normal subject. Dotted line means command (desired) ankle joint angle. Values for control parameters are: a)K_i =15, K_p =15 and K_d =23. Figure (a) corresponds to vertical standing, and (b) 2 degrees leaning forward.

is recorded. The case presented in (c) is better because it compensates the disturbance. This result shows that the use of PD controller may resolve the problem.

The use of PD controller provides stable standing with the zero angle towards the gravity vector. The desirable angle for standing is few degrees in front of the vertical line (leaning slightly forward) because of the position of the center of the pressure while standing. The application of PID controller (Figure 10) demonstrates that with the small feedback gain desired angle does not coincide with the measured angle (a). The high gain leads to serious oscillation, while the medium case (b) is the best.

The time delay from the initiation of the process should be observed in Figure 10. This delay is an indirect result of the dorsal flexion torque.

Comparison of two parameter groups

The parameter determination process is time consuming (more than 30 minutes) in general case. When the experimentally determined parameters coincide with analytical results less time is needed for efficient use of the controller. The comparison of mentioned parameters is given in Table 1.

	(a)	(b)	(c)
K_i	10 ~ 25	14.0	19.2
K_p	5 ~ 15	33.9	16.4
K_d	10 ~ 20	11.1	9.0

Table 1. Controller parameters. (a) experimental results, (B) and (c) determined from the inverted pendulum model. l=1.7 m, m = 60 kg, T = 100 ms, K_mK_s=0.2(b) K= 50 Nm/rad, B = 0.5 Nm.srad (rest state), (c) K = 300 Nm/rad, B = 1.5 Nm/srad (flexed by 20 Nm)

The values of B and K are values in rest and flexed state with 20 Nm [15,16]. The parameter K_p in rest state is considerably different from K_p obtained experimentally. Other parameters K_i and K_d coincide with experiments. Parameters determined with flexed state values of K and B indicate very good coincidence with experimental values. The complete study indicates the benefit of the inverted pendulum model for the determination of control parameters.

Controlled range with FNS standing system

The controlled range with the FNS system and voluntary control of subject was examined and shown in Table 2. Almost 50% of the voluntary control is obtained through the FNS system. The evaluation of the reliable system was continued by comparing characteristics of voluntary and FNS controlled standing.

subject	FNS		voluntary	
	forward	backward	forward	backward
(a)	1.4	0.1	4.6	1.8
(b)	2.0	0.0	4.6	1.1
(c)	1.5	0.1	3.5	0.8
(d)	4.4	0.2	5.5	1.0

Table 2. Controllable range of FNS system and voluntary control in normal subjects

Standing control in paraplegic patients

Standing control of a paraplegic patient was carried out with the presented system. The difference was the use of ankle-knee brace, Fig. 11. Hip joint could not be fixed because of very weak gluteal muscles. There are a number of unsolved problems in paraplegic standing: 1) it is complicate to create sufficient hip extension through FNS with surface electrodes; 2) feedback obtained form the ankle joint is not sufficient, ground reaction force or some other dynamic measures should be introduced in the feedback loop; 3) system used is not portable which is desirable.

Figure 11. Standing control of paraplegic patient using ankle-knee brace. Joints of brace were not fixed and knee extension as well hip control was achieved with electrical stimulation of quadriceps and gluteal muscles.

CONCLUSION

Our comparative study on control parameters based on inverted pendulum model and experiments proved that it is possible to control the posture if the single joint control is needed. In multiple joint systems, as mentioned in control of paraplegic posture, the feedback system has to integrate the dynamic feedback. Comparative clinical studies on the closed-loop control system and detailed analysis of biomechanical models are important step in development of technique for the design of optimal FNS controller. The efficacy of the control system is one of most important elements in development of safe, reliable and practical systems for motor restoration in paralyzed humans.

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