E. 結論

機能的電気刺激（FES）システムを患者自身が制御するために，脳波から on/off のコマンドを検出する BCI（brain-computer interface）システムの開発のための検討を行った．ユーザの頭頂部から 1 チャネルの脳波を双極誘導し，それからユーザの足の運動イメージの有無を検出するシステムの開発を行った．6 名中 2 名の被験者で 60～90%程度の正答率でコマンド検出が可能であった．

本システムでは単一チャネルで on/off のコマンドを検出することが可能であり，単機能ながら先行研究に比べてユーザの負担の少なく，正答率の高いシステムの可能性を示すことができた．コマンド検出アルゴリズムの改良による正答率の向上，患者への適用可能性の検討などが今後の課題である．

なお，H16 年度に本事業で実施した，眼電図から眼球運動を検出することでメニュー選択を行うインターフェイスシステムに関する研究をまとめ，学会で発表を行った（平成 17 年度 1 件，平成 18 年度 1 件）．このシステムは眼部の上下，左右に電極を設置し，眼球運動に伴って生じる眼電図（EOG）を簡易な方法で計測することで，ユーザの眼球運動の方向（8 通り）と距離（2 通り）の 16 通りの組み合わせをおおよそ 95%の正答率で検出する技術を開発した．このシステムは，簡易な眼電図計測によって高い正答率が得られるインターフェイスシステムであることが示された．また，CMOS カメラによって撮像した眼部のイメージから視線方向をリアルタイムで検出することで障害者用インターフェイスを実現できる可能性もあわせて報告した（平成 18 年度 1 件）．

G. 研究発表（平成 18 年度）

1. 論文発表

なし

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H. 知的所有権の出願・登録状況（予定を含む）

なし

III. 研究成果の刊行に関する一覧表

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IV. 研究成果の刊行物・別刷

PAPER

Design of Fuzzy Controller of the Cycle-to-Cycle Control for Swing Phase of Hemiplegic Gait Induced by FES

Achmad ARIFIN[†a)], Takashi WATANABE[††], *Nonmembers*, *and* Nozomu HOSHIMIYA[†††], *Fellow*

SUMMARY The goal of this study was to design a practical fuzzy controller of the cycle-to-cycle control for multi-joint movements of swing phase of functional electrical stimulation (FES) induced gait. First, we designed three fuzzy controllers (a fixed fuzzy controller, a fuzzy controller with parameter adjustment based on the gradient descent method, and a fuzzy controller with parameter adjustment based on a fuzzy model) and two PID controllers (a fixed PID and an adaptive PID controllers) for controlling two-joint (knee and ankle) movements. Control capabilities of the designed controllers were tested in automatic generation of stimulation burst duration and in compensation of muscle fatigue through computer simulations using a musculo-skeletal model. The fuzzy controllers showed better responses than the PID controllers in the both control capabilities. The parameter adjustment based on the fuzzy model was shown to be effective when oscillating response was caused due to the inter-subject variability. Based on these results, we designed the fuzzy controller with the parameter adjustment realized using the fuzzy model for controlling three-joint (hip, knee, and ankle) movements. The controlled gait pattern obtained by computer simulation was not significantly different from the normal gait pattern and it could be qualitatively accepted in clinical FES gait control. The fuzzy controller designed for the cycle-to-cycle control for multi-joint movements during the swing phase of the FES gait was expected to be examined clinically.

key words: *cycle-to-cycle control, FES-induced gait, multi-joint control, fuzzy controller, adaptive fuzzy controller*

1. Introduction

Functional electrical stimulation (FES) has been utilized to restore gait in patient with impairment of the central nervous system caused by the spinal cord injury or the stroke [1], [2]. The human gait is a complex task, therefore FES-induced gait requires appropriate control methods. The control methods used in FES involve open-loop and closed-loop controls. The previous experimental researches [3], [4] showed limitation of the open-loop control to compensate muscle fatigue. A trajectory-based closed-loop control for the knee joint angle of paraplegic gait had poor tracking and oscillating responses, and could not reach full knee extension angle in some trials [5]. Capability of tracking a target joint angle is still not clear.

The cycle-to-cycle control was expected to be an alternative to the trajectory-based closed-loop FES gait control. Its capability to realize the target joint angle had been shown in experimental test of controlling maximum knee extension angle [6] or hip joint angle range [7] using a proportional-integral-derivative (PID) controller. However, these experimental tests were performed in single-joint control. The cycle-to-cycle control should be implemented for controlling multi-joint movements in clinical use. Because there is a lack of method of determination of parameter values of the PID controller of the cycle-to-cycle control, implementation of the cycle-to-cycle control for multi-joint movements using the PID controller is difficult.

We proposed fuzzy controller to realize the cycle-to-cycle control. The fuzzy controller was shown to be superior to the PID controller in controlling the knee joint movements of swing phase of FES gait [8]. The fuzzy control scheme had been shown to be effective in developing nonlinear controllers [9]. Using the fuzzy controller, system identification to determine the controller parameter values could be eliminated and design procedure was simplified. In order to develop a practical cycle-to-cycle control, the fuzzy controller has to be realized for controlling the multi-joint movements.

In this paper, we designed the fuzzy controller of the cycle-to-cycle control for multi-joint movements of the swing phase of FES gait. First, we designed three different fuzzy controllers (a fixed fuzzy controller, a fuzzy controller with parameter adjustment based on gradient descent method, and a fuzzy controller with parameter adjustment based on a fuzzy model) and two PID controllers (a fixed PID controller and an adaptive PID controller) for controlling two-joint (knee and ankle) movements. They were examined through computer simulations with the aim of exploring the control capability of the designed controllers in the basic structure of the lower limbs. Based on these results, we realized the cycle-to-cycle control for three-joint (hip, knee, and ankle) movements. The swing phase with the three-joint movements controlled by the designed controllers was evaluated in the computer simulations comparing to the measured joint angle trajectories of the normal gait.

2. Framework of the Cycle-to-Cycle Control

Gait is one of the cyclic movements. Each gait cycle is divided into the stance phase and the swing phase. In a cer-

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tain sub-phase of the gait, the joint movements reach certain joint angles (e.g., maximum knee flexion angle of swing phase, maximum ankle dorsifelxion angle of swing phase, ankle joint angle at initial contact). In the cycle-to-cycle control, stimulation burst duration, TB, is regulated, while pulse amplitude, pulse width, and frequency of stimulation pulse are fixed. Each muscle is stimulated by single burst of electrical pulses to produce relevant joint movement reaching a target joint angle. The controlled joint angle of the previous cycle is fedback to the controller. Regulation of the stimulation burst duration of the current cycle of gait is based on difference between the target joint angle and the controlled joint angle of the previous cycle.

A set of certain target joint angles and a stimulation schedule are required to realize the cycle-to-cycle control. We performed an experiment to measure the hip, knee and ankle joint angles during level gait. Five healthy subjects (males, 24±2.9 years old, 170.2±4.5 cm, 60.6±5.4 kg) participated in the experiment. Purpose of the experiment was explained to each subject and subject's consent was obtained. Three goniometers (M180, Penny & Giles, UK) were used to measure the hip, knee, and ankle joint angles of right leg of the subject. Two force sensitive resistors (FSRs) were attached to the heel and the toe of the foot to detect the stance and the swing phases. The subjects were instructed to perform five trials of level gait with comfortable speed. A trial of gait consisted of seven gait cycles. The output signals of the goniometers and the FSRs were recorded into the data cartridge using RTD-145 (TEAC, Japan) and digitized by an A/D converter (AT-MIO-16E-10, National Instruments) with 1 kHz of sampling frequency. Data processing was performed using MATLAB version 6.5 Release 13. The data of FSRs were low-pass filtered using fourth order, zero phase lag, Butterworth digital filter with cut-off frequency of 1 Hz. Data of the joint angles were smoothed using the fourth order, the zero phase lag, the Butterworth digital filter with 20 Hz of cut-off frequency. Gait analysis of the data collected from the experiment was performed to determine the target joint angles. Parameter $\Delta\theta$ was also determined in order to evaluate whether the target joint angle was reached or not. The controlled joint angle reached the target when the absolute error was less than or equal to $\Delta\theta$ as illustrated in Fig. 1. Values of $\Delta\theta$ of each controlled joint angle was set by average value of intra-subject standard deviation of each target joint angle from the gait analysis. Values of the target joint angles and the $\Delta\theta$ are shown in Table 1.

We developed the stimulation schedule to develope the joint movements of the swing phase as shown in Fig. 2, which was designed based on the knowledge about the joint movements during swing phase and muscle functions to produce the relevant joint movements [10]. Stimulations of the iliopsoas, the biceps femoris short head (BFSH) and the biceps femoris long head (BFLH), the vastus muscles and the rectus femoris, the gastrocnemius medialis, and the tibialis anterior were controlled to induce the joint movements reaching the following target joint angles: maximum hip flexion angle, maximum knee flexion angle, maximum knee extension angle, maximum ankle plantar flexion angle, and maximum ankle dorsiflexion angle, respectively. After the hip joint reached the target maximum hip flexion angle, the iliopsoas was stimulated again to keep hip flexion and to reach the target of hip joint angle at initial contact. The tibialis anterior and the soleus were stimulated simultaneously to reach the target of ankle joint angle at initial contact. Beginnings of the muscle stimulation were at the maximum hip extension, maximum knee extension and maximum an-

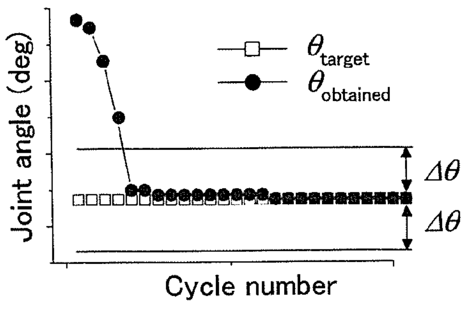

Fig. 1 Illustration of $\Delta\theta$ in control evaluation. $\theta_{obtained}$ is the obtained joint angle in each cycle of gait and θ_{target} is the target joint angle.

Table 1 Target joint angle and $\Delta\theta$.

Joint	Angle	Target	$\Delta\theta$
Hip	max. flexion	32.4°	2.0°
	initial contact	29.3°	2.8°
Knee	max. flexion	69.0°	1.9°
	max. extension	3.6°	2.7°
Ankle	max. plantar flexion	−16.4°	3.4°
	max. dorsiflexion	4.9°	1.3°
	initial contact	−0.3°	1.3°

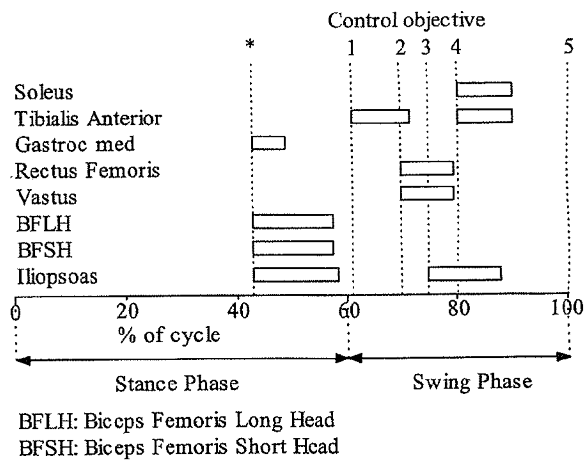

Fig. 2 The stimulation schedule. *: the beginning of the stimulation (the maximum hip extension angle, the maximum knee extension angle of the end of the stance phase, and the maximum ankle dorsiflexion angle of the end of the stance phase). The control objective: 1: the maximum ankle plantar flexion angle, 2: the maximum knee flexion angle, 3: the maximum hip flexion angle, 4: the maximum ankle dorsiflexion angle, and 5: the maximum knee extension angle and the hip and the ankle angles at initial contact.

kle dorsiflexion angles at the end of stance phase. In our result of the gait analysis, timing of those maximum joint angles in a cycle of gait varied among subjects. In order to facilitate the computer simulation, we assumed these maximum joint angles were occurred simultaneously.

3. Musculo-Skeletal Model for FES Gait

We developed the electrically stimulated musculo-skeletal model for FES gait based on Eom et al. [11]. Values of the musculo-skeletal model parameters were obtained from Ogihara et al. [12]. The joint movements were considered as the movements of a multiple pendulum. The motion equation was derived from the model using the Lagrange function as shown in Eq. (1),

$$\mathbf{M}\ddot{\boldsymbol{\theta}} + \mathbf{C}\dot{\boldsymbol{\theta}} + \mathbf{G} = \boldsymbol{\tau} \tag{1}$$

where $\boldsymbol{\theta}$, $\dot{\boldsymbol{\theta}}$, $\ddot{\boldsymbol{\theta}}$ are vectors of the joint angle, the joint angular velocity, and the joint angular acceleration, $\mathbf{M}$ is the inertial matrix, $\mathbf{C}$ is the coriolis vector, $\mathbf{G}$ is the gravitational vector, and $\boldsymbol{\tau}$ is vector of the joint torque. The model for the three-joint movements consisted of the paralyzed and the normal legs. In this study, the joint angles of the normal leg were simulated using the values of joint angles measured from the normal subject. The joint movements developed by the electrical stimulation of the muscles were calculated by integrating the motion equation in Eq. (1) by the fourth order Runge-Kutta method with $10\,\mu$s integration time step.

In order to test the capabilities of the designed controllers in controlling different subjects, we created twenty different subject models. The twenty different subject models were expressed by changing values of maximum muscle forces (50%–150%), mass of the thigh, the shank, and the foot (50%–150%), and/or length of the thigh, the shank, and the foot (75%–125%) of a reference subject model.

4. Computer Simulation Test in Two-Joint Control

The stimulated muscles in the two-joint control were the BFSH, the vastus muscles, the gastorocnemius medialis, the tibialis anterior, and the soleus. The sequence of the stimulation of the muscles was based on the stimulation schedule shown in Fig. 2. We designed the three different fuzzy controllers, controller A was a fixed parameter fuzzy controller, controller B was a fuzzy controller with the gradient descent based parameter adjustment and controller C was a fuzzy controller with fuzzy model parameter adjustment and oscillation detection. A PID controller (controller D) and an adaptive PID controller (controller E) were also designed. Concrete descriptions of the design of the fuzzy and the PID controllers are given in the following subsections.

4.1 Design of Fuzzy Controller

Algorithm of regulation of the stimulation burst duration, TB, by the fuzzy controller is shown in Eq. (2),

$$TB[n] = TB[n-1] + \Delta TB[n] \tag{2}$$

where $TB[n]$ is the stimulation burst duration for the current cycle, $TB[n-1]$ is the stimulation burst duration of the previous cycle, and $\Delta TB[n]$ is the output of the fuzzy controller. The fuzzy controllers were designed through five design steps: (i) selection of controller inputs, (ii) assignment of the fuzzy membership functions, (iii) design of the fuzzy rules, (iv) assignment of fuzzy inference and deffuzification, and (v) design of parameter adjustment.

4.1.1 Selection of Controller Inputs

The inputs for the BFSH, the gastrocnemius medialis, and the iliopsoas controllers were the *errors*. These controllers were designed as single-input single-output (SISO) controllers. The *error* was defined as the difference between the target and measured joint angles. The controllers for the vastus muscels, the soleus, and the tibialis anterior were designed as multi-input single-output (MISO) controllers. The inputs of these controllers were *error* and *desired range* of joint angle. The desired range of the knee extension angle was defined as the difference between the obtained maximum knee flexion angle of the current cycle and the target maximum knee extension angle. The desired range of the ankle dorsiflexion angle was defined as the difference between the obtained maximum ankle plantar flexion angle of the current cycle and the target maximum ankle dorsiflexion angle. The desired range of the ankle angle at the initial contact was defined as the difference between the obtained maximum ankle dorsiflexion of the current cycle and the target ankle joint angle at the initial contact.

4.1.2 Assignment of Fuzzy Membership Functions

Input membership functions were expressed as triangular and trapezoidal fuzzy sets. Output membership function was expressed in the fuzzy singletons. The fuzzy linguistic terms of the input and output variables are shown in Table 2. The values of parameters of the fuzzy membership functions were assigned and refined through preliminary computer simulation.

Table 2 Fuzzy linguistic term.

Input		Output	
desired range	*error*	ΔTB^*	
S	NL	NL2	S: small
M	NM	NL1	M: medium
L	NS	NL	L: large
	Z	NM	NL2: negative large 2
	PS	NS	NL1: negative large 1
	PM	Z	NL: negative large
	PL	PS	NM: negative medium
		PM	NS: negative small
		PL	Z: zero
		PL1	PS: positive small
		PL2	PM: positive medium
			PL: positive large
			PL1: positive large 1
			PL2: positive large 2

4.1.3 Design of Fuzzy Rules

The fuzzy rules directed control action to compensate the error by increasing TB when error was negative, otherwise decreasing TB when error was positive. In linguistic expression, the increasing TB was expressed by taking a positive value of ΔTB^*, and decreasing TB by taking a negative value of ΔTB^*. The fuzzy rule sets for the SISO controllers and the MISO controllers are shown in Tables 3 and 4, respectively. The examples of the fuzzy rule of the SISO and the MISO controllers are shown below.

SISO controller:

IF *error* is negative medium (NM) THEN ΔTB^* is positive medium (PM)

MISO controller:

IF *error* is negative small (NS) AND *desired range* is small (S) THEN ΔTB^* is positive small (PS)

4.1.4 Assignment of Fuzzy Inference and Defuzzification

In this study, the fuzzy inference was accomplished by using the Mamdani method. Defuzzification process converted the fuzzy inference outputs, ΔTB_i^*, which were resulted by the ith rule, into a crisp value ΔTB. Center of gravity (COG) shown in Eq. (3) was used in the defuzzification,

$$\Delta TB = \frac{\Sigma\mu(P_k[n])P_k[n]}{\Sigma\mu(P_k[n])} \tag{3}$$

where $k = 1, 2, ..K$, K is the number of the fuzzy linguistic term of ΔTB^*, $P_k[n]$ is the value of the fuzzy singleton of the kth linguistic term of ΔTB^*, $\mu(P_k[n])$ is membership value of $P_k[n]$.

4.1.5 Design of Parameter Adjustment

In the controllers B and C, the parameter adjustment was applied to the value of fuzzy singleton (P) of each fuzzy linguistic term of the output variable ΔTB^*. The parameter adjustment of the controller B was based on the gradient descent method shown in Eq. (4),

$$P_k[n] = P_k[n-1] - \eta\frac{\partial V}{\partial P_k} \tag{4}$$

where $k = 1, 2, ..K$, K is the number of the fuzzy linguistic term of ΔTB^*, η is the adaptation constant, and V is the objective function as shown in Eq. (5),

$$V = \frac{1}{2}(\theta_{tar} - \theta_{obt}[n-1])^2 \tag{5}$$

where θ_{tar} is the target joint angle and $\theta_{obt}[n-1]$ is the obtained joint angle of the previous cycle. Value of the adaptation constant was determined by trial and error method in preliminary computer simulation of automatic generation of stimulation burst duration with the reference subject model. The parameter adjustment of the controller B was applied continuously during swing phase control.

The parameter adjustment of the controller C was realized based on the fuzzy model. This parameter adjustment was applied when inappropriate responses of controlled joint angles (oscillations) were detected. We introduced a sensitivity of the stimulated musculo-skeletal system to adjust the parameter of the output membership function. The sensitivity was defined as ratio of change of joint angle range to change of stimulation burst duration. It was determined by averaging its value of five cycles after the absolute error was less than or equal to $\Delta\theta$ or after the first change of the sign of the error. The parameter adjustment algorithm was realized in the form of fuzzy model with three fuzzy rules mapping the sensitivity onto a weight, w^*, as shown below.

IF sensitivity is small (S) THEN w^* is large (L)
IF sensitivity is medium (M) THEN w^* is medium (M)
IF sensitivity is large (L) THEN w^* is small (S)

The real value of the weight, w, was calculated by using the COG method. The value of the fuzzy singleton (P) was adjusted by multiplying its value of the previous cycle with the weight, w, as shown by Eq. (6).

$$P_k[n] = wP_k[n-1] \tag{6}$$

Structure of the fuzzy controller for the cycle-to-cycle control with parameter adjustment based on the fuzzy model is shown in Fig. 3.

Table 3 Fuzzy rule set of SISO controller.

eror	NL	NM	NS	Z	PS	PM	PL
ΔTB^*	PL	PM	PS	Z	NS	NM	NL

Table 4 Fuzzy rule set of MISO controller.

		desired range		
		S	M	L
error	NL	PL	PL1	PL2
	NM	PM	PL	PL1
	NS	PS	PM	PL
	Z	Z	Z	Z
	PS	NS	NM	NL
	PM	NM	NL	NL1
	PL	NL	NL1	NL2

4.2 Design of PID Controller

The PID controllers (controllers D and E) were realized in the single-input single-output (SISO) controllers because of difficulty of implementation of the multi-input PID controller for the cycle-to-cycle control and the lack of determination method of the controller parameter values. The input was the error of the previous cycles. Control algorithm of the PID controller for the cycle-to-cycle control is shown in Eq. (7),

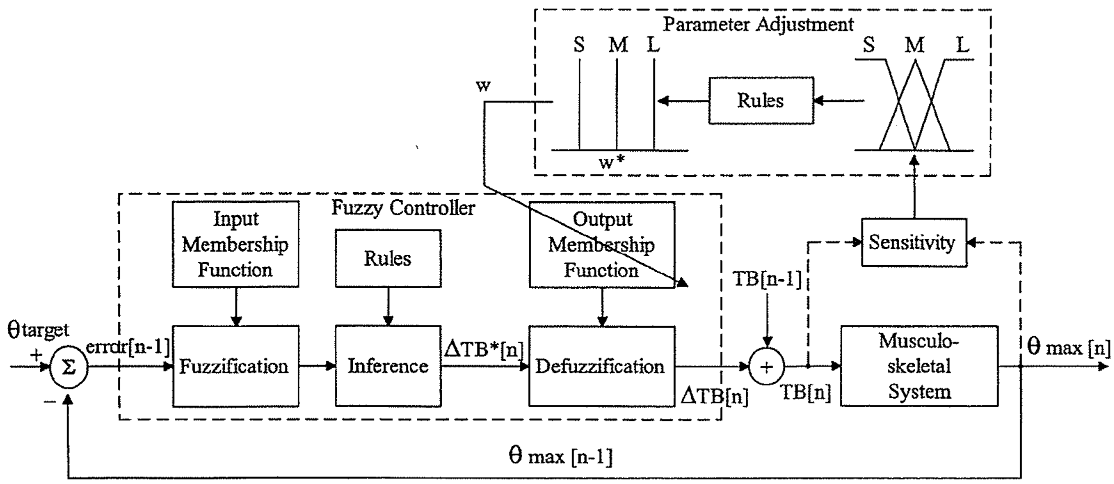

Fig. 3 Structure of the fuzzy controller with the parameter adjustment based on the fuzzy model. In this figure, the maximum joint angle is an example of the controlled joint angle.

$$\begin{aligned} TB[n] = TB[n-1] + G(&error[n-1] \\ &- (z_1 + z_2)error[n-2] \\ &+ z_1 z_2 error[n-3]) \end{aligned} \tag{7}$$

where G is the controller gain, *error* is the difference between the obtained joint angle and the target, z_1 and z_2 are zeros of the controller. The controller parameter z_1 was set as identified parameter value of lower leg model [16], [17]. The parameter G and z_2 were determined by the Jury's stability test and step response, respectively. Optimal values of the parameters z_1 and z_2 were determined by a preliminary computer simulation. The parameter adjustment of the controller E was applied to all controller parameters (G, z_1 and z_2) as shown in Eq. (8),

$$\begin{aligned} G[n] &= G[n-1] - \eta\frac{\partial V}{\partial G} \\ z_1[n] &= z_1[n-1] - \eta\frac{\partial V}{\partial z_1} \\ z_2[n] &= z_2[n-1] - \eta\frac{\partial V}{\partial z_2} \end{aligned} \tag{8}$$

where η is the adaptation constant and V is the objective function as shown in Eq. (5). Value of the adaptation constant of the adaptive PID controller was determined by trial and error method in preliminary computer simulation of automatic generation of stimulation burst duration with the reference subject model.

4.3 Method of Computer Simulation

Computer simulations to test capabilities of the designed controllers in controlling the knee and ankle joint movements were performed using a reference and twenty different subject models. In the two-joint control, the hip joint of the model was fixed at 15°. The computer simulation test was divided into two parts: in automatic generation of the stimulation burst durations and in compensating muscle fatigue. In the automatic generation of stimulation burst duration, the computer simulation was initiated with zero burst durations. The appropriate burst duration (TB_{app}) of each muscle was obtained by averaging its burst durations of five cycles after all the controlled joint angles reached the targets with absolute errors were less than or equal to $\Delta\theta$. In order to test the use of the TB_{app} of one subject as initial burst duration to the other subjects, the TB_{app} of the reference subject were applied to different subject models as the initial burst durations in the separate computer simulation. The computer simulation tests were performed in the stimulation course of 200 cycles of the swing gait.

We modeled the muscle fatigue as an exponential decrease of the maximum muscle force, F_{max}, to 50% of its original value as shown in Eq. (9).

$$F_{max} = F_{max0} - \frac{F_{max0}}{2}\left(1 - e^{-\frac{n-n_f}{\alpha}}\right); n > n_f \tag{9}$$

where F_{max0} is the original maximum muscle force, n is the cycle number, n_f is the cycle number when a muscle begins to fatigue, and α is a decay constant. The values of the decay constant were 5, 50 and 200 to represent the sudden, the moderate and the gradual fatigue, respectively. Each muscle was assumed to be fatigue after the 75th cycle. The test of the fatigue compensation was performed in the independent computer simulation of the fatigue of each muscle. In the muscle fatigue compensation test, electrical stimulation of each subject was started with its own TB_{app} obtained in the test of the automatic generation of the stimulation burst duration.

4.4 Evaluation of Control Performances

In automatic generation of the stimulation burst duration, the controlled joint angles could not reach the targets at the beginnings of the stimulations, because the muscle stimulations were initiated with zero burst durations. The controlled joint angles reached the targets when the appropriate stimulation burst durations were found. We defined the settling index as the number of cycles that were required to reach the target joint angle with absolute error that was less than or equal to $\Delta\theta$. Average value of the settling index of each controller is shown in Table 5. All the fuzzy controllers could regulate the stimulation burst durations properly, therefore the controlled joint angles reached the targets in a few cycles. However, the controller A resulted in oscillating responses of maximum knee extension angle in a subject with very large muscle force (very strong subject) as shown in Fig. 4. The parameter adjustment based on the gradient descent method of the controller B could not suppress the oscillating maximum knee extension angle. The parameter adjustment based on the fuzzy model in the controller C could suppress oscillation of the maximum knee extension angle. The fuzzy controllers had smaller settling indexes than the PID and the adaptive PID controllers (proved by paired t-test with 0.005 significance level). Parameter adjustment of the controller E could not significantly improve the capability of the fixed PID controller. Using the appropriate burst durations of the reference subject as initial burst durations for the different subjects could reduce settling indexes. However, this method resulted in occasionally inappropriate joint angles at the beginning of stimulation.

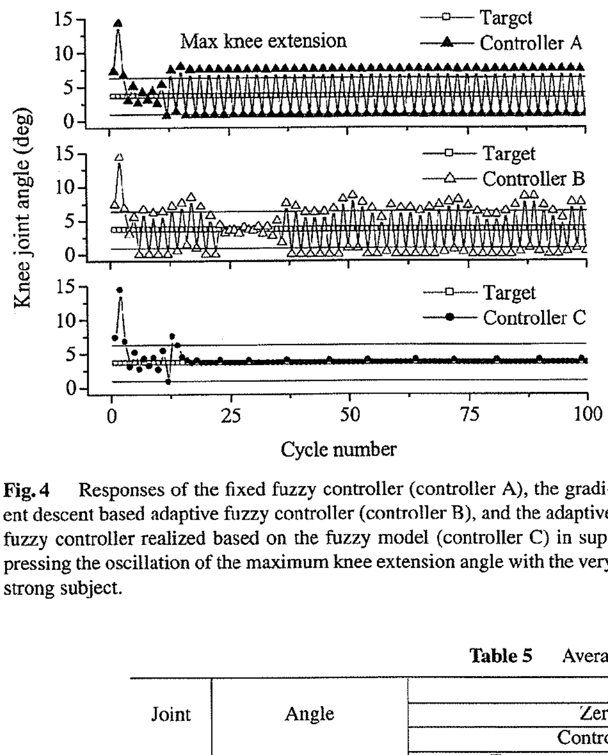

Fig. 4 Responses of the fixed fuzzy controller (controller A), the gradient descent based adaptive fuzzy controller (controller B), and the adaptive fuzzy controller realized based on the fuzzy model (controller C) in suppressing the oscillation of the maximum knee extension angle with the very strong subject.

Muscle fatigue caused decreasing of the controlled joint angles in achieving the targets. Criteria for the control performance in compensating the muscle fatigue were the recovery index and the maximum error. The recovery index was defined as the number of cycles that were required to compensate muscle fatigue. Here, the fatigue compensation was achieved when the absolute error decreased to be less than or equal to $\Delta\theta$. The maximum error was defined as the largest difference between the obtained joint angle and the target during muscle fatigue. Average values of the recovery index and the maximum error are shown in Table 6. The controller A resulted in the oscillating maximum knee extension angle in the very strong subject. The recovery indexes of the maximum ankle dorsiflexion angle and the ankle angle at initial contact of the PID controllers (controller D and E) were larger than those of the fuzzy controllers. The recovery indexes and maximum errors among the three fuzzy controllers were not significantly different. However, the controller C had smallest recovery index and smallest maximum error.

Table 5 Average settling index (cycles).

Joint	Angle	Initial TB									
		Zero					TB_{app} of ref. subject				
		Controller					Controller				
		Fuzzy			PID		Fuzzy			PID	
		A	B	C	D	E	A	B	C	D	E
Knee	max. flexion	8±2	6±1	8±2	16±2	18±9	3±2	3±2	3±2	5±4	4±4
	max. extension	4±1*	5±2*	4±1	13±4	12±4	2±1*	2±1*	2±1	4±3	4±3
Ankle	max. plantar flexion	5±1	5±1	5±1	10±2	10±2	1±1	1±1	1±1	2±2	2±2
	max. dorsiflexion	5±2	4±1	5±2	36±8	39±8	2±2	2±2	2±2	5±11	5±11
	initial contact	4±1	4±3	4±1	26±13	26±13	1±1	1±1	1±1	2±2	2±2

*: oscillating response

Table 6 Average recovery index (cycles) and maximum error (deg).

Joint	Angle	Recovery index					Maximum error				
		Controller					Controller				
		Fuzzy			PID		Fuzzy			PID	
		A	B	C	D	E	A	B	C	D	E
Knee	max. flexion	8±11	10±16	8±11	10±16	11±18	3.4±3.3	4.3±6.3	3.4±3.3	3.3±3.2	3.4±3.3
	max. extension	1±4*	12±22	0±0	3±6	3±7	0.8±0.9	2.3±3.4	0.6±0.5	1.3±1.3	1.3±1.3
Ankle	max. plantar flexion	0±0	3±14	0±0	0±0	0±0	1.1±0.8	1.5±2.3	1.1±0.8	1.1±0.9	1.1±0.9
	max. dorsiflexion	5±7	3±9	5±7	17±26	18±25	0.7±0.7	0.7±0.8	0.7±0.7	1.3±1.0	1.3±1.0
	initial contact	3±15	16±33	3±15	19±42	19±42	0.7±0.6	1.2±0.9	0.7±0.6	1.1±1.1	1.1±1.2

*: oscillating response

5. Fuzzy Controller for Three-Joint Movements

5.1 Design of Fuzzy Controller

The effectiveness of using fuzzy controller with the parameter adjustment based on the fuzzy model (controller C) was shown in controlling the two-joint movements. We tried to realize the cycle-to-cycle control for controlling the three-joint (hip, knee, and ankle) movements using this controller. The fuzzy controllers for the three-joint movements were developed by the addition of the controller for the hip joint muscle (the iliopsoas) to the controllers for the two-joint movements. The iliopsoas muscle controller was designed as a SISO controller. Although eight muscles were controlled based on the stimulation schedule shown in Fig. 2, the BFLH and the rectus femoris were controlled by the controller of the BFSH and the vastus muscles, respectively.

Table 7 Settling index of the three-joint control (cycles).

Joint	Angle	Settling index
Hip	max. flexion	9±3
	initial contact	7±3
Knee	max. flexion	4±1
	max. extension	6±2
Ankle	max. plantar flexion	3±1
	max. dorsiflexion	5±1
	initial contact	7±1

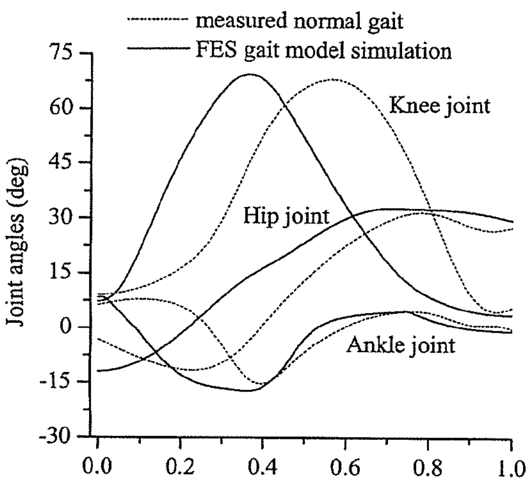

Fig. 5 Trajectories of the controlled joint angles of the very strong subject compared to the trajectories of the joint angles measured from the normal subject.

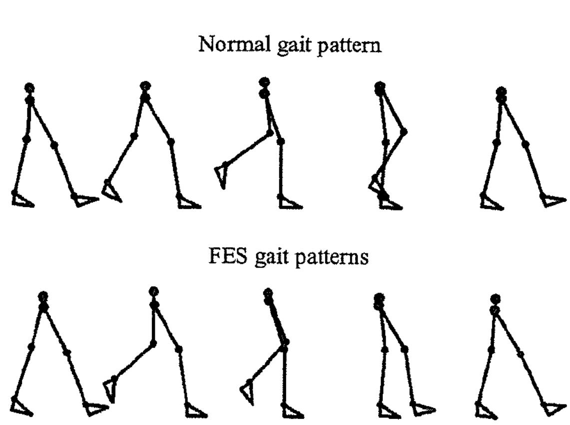

Fig. 6 Stick picture of the FES-induced gait pattern and the normal gait pattern. The black leg is the controlled swing leg and the gray leg is the stance leg.

5.2 Evaluation of Control Performances

The computer simulation test was examined in automatic generation of the stimulation burst duration with twenty different subject models. The fuzzy controller could regulate the stimulation burst durations of all the stimulated muscles, therefore all the controlled joint angles reached the targets in a few cycles as shown in Table 7. Although the hip joint was controlled simultaneously, the average settling indexes of the knee and ankle joint angles were not significantly different from the average value of the two-joint control.

Figure 5 shows the trajectories of the controlled joint angles obtained from the very strong subject as an example of the controlled joint angle trajectories. Where the other fuzzy controllers could not suppress the oscillation of the maximum knee joint angle of the very strong subject. The joint angles in Fig. 5 were obtained under the condition of all the controlled joint angles reached the target (well controlled gait). The measured joint angles of the normal gait were also shown in the figure. RMS error was defined as the root mean square value of the difference between the controlled joint angle and the measured joint angle of the normal gait. The RMS errors of the hip, the knee, and the ankle joint angles were 9.6°±1.0°, 21.6°±2.0°, and 8.1°±0.4°, respectively. Although the flexion of the controlled knee joint was earlier than the normal gait, it did not decrease the quality of the swing phase as seen in Fig. 6. Figure 6 shows the gait pattern of the very strong subject in the computer simulation compared to the measured normal gait pattern. The controlled gait pattern was not significantly different from the normal gait pattern. The minimum foot clearance was defined as the distance between the toe of the foot to the floor during swing phase. The minimum foot clearance that was obtained by averaging its values of five cycles under the well controlled gait ranged from 2.0 to 3.2 cm (2.8±0.4 cm). The average of the timing of the minimum foot clearance was at 60.8% of the normalized swing time. The stride length was defined as the distance of the horizontal displacement of the heel from the beginning of the swing gait to the end of that. The average value of the stride length was 121.5±13.3 cm.

6. Discussions

In this study, we designed the fuzzy controller of the cycle-to-cycle control for multi-joint movements of the swing of

FES gait. On the point of view of the controller design, use of the expert knowledge eliminated the system identification and the fuzzy controller for two-joint and three-joint movements could be designed with simplified procedure. Flexibilities in assignments of the input variables and the parameters of the membership functions provided the easy design. By using the fuzzy control scheme, the difficulty of implementation of the cycle-to-cycle control for multi-joint movements could be solved. On the point of view of the control results, the fuzzy controller for the three-joint movements could regulate the hip, knee and ankle joint angles to reach their targets during swing phase. The controlled swing gait pattern qualitatively resembled the normal gait pattern. The average values of the minimum foot clearance were similar to those of the normal gait (2.19 ± 0.66 cm) [14]. Although the timing of the minimum foot clearance was 10% earlier than in the normal gait [14] and flexion of the knee joint was earlier than those of the normal gait, the pattern of the entire swing gait (Fig. 6) can be acceptable in application of FES gait. From the above two points, the fuzzy controller would be effective to implement the cycle-to-cycle control for multi-joint movements.

The PID controller is considered to be less practical to implement the cycle-to-cycle control method. Standard method of parameter setting of the PID controller (e.g., the Ziegler & Nichols method [13]) can not be applied to the PID controller for the cycle-to-cycle control and there is a lack of determination method of the controller parameter values. The parameters values identified from the passive test of the lower leg model used in this study was an alternative method in order to cope the lack of the systematic method of the determination of the controller parameter values. However, several trials were required to find the optimal parameter values for the PID controllers of the ankle joint muscles. Additionally, implementation of the multi-input PID controller requires other controller parameters that must be identified from the system characteristics. Therefore, the PID controller for each muscle was designed in the SISO controller for the tests of controlling the two-joint movements, which were aimed to explore its capabilities in controlling the basic structure of the lower limb movements. Because of the lack of design of the PID controller and the advantage of the fuzzy controller as described in the previous paragraph, the fuzzy controller is considered to be more appropriate to implement the cycle-to-cycle control.

The parameter adjustment based on the fuzzy model would be effective to compensate the inter-subject variability. We considered that the oscillating maximum knee extension due to the inter-subject variability was essential to be suppressed. An extremely extended knee joint may reduce the patient comfort. On the other hand, an inadequate knee extension decreases stride length and influences quality and stability of the swing phase. Convergence of the controller with parameter adjustment based on the gradient descent method depends on the adaptation constant. Too small adaptation constant will have a slow convergence. On the other hand, too large adaptation constant will result in a divergent response such as an oscillating response. In the parameter adjustment based on the gradient descent method, although the adaptation constants were optimized in the preliminary computer simulation test using the reference subject model, it could not compensate the inter-subject variability properly. The gradient descent parameter adjustment method is not useful to ensure a convergent response at clinical site because the adaptation constant is required to be determined for each patient. We showed in this study that the parameter adjustment based on the fuzzy model could compensate the inter-subject variability. Therefore, it is expected to be more useful than the parameter adjustment based on the gradient descent method in the real application of the cycle-to-cycle control for FES gait.

The cycle-to-cycle control is expected to be effective and practical in controlling the multi-joint movements of the swing phase of FES-induced gait. Hatwell et al. [5] reported a trajectory-based closed-loop FES control for the knee joint of paraplegic gait had poor tracking and oscillating responses, and could not reach the full knee extension angle in some trials. Sliding mode control may be expected to improve tracking performance [15]. The sliding mode control requires less knowledge about parameter of the human neuro-musculo-skeletal system. Its inherent robust properties are expected to overcome the difficulties of FES control caused by parameter uncertainties or variations among subjects. However, experimental data showed the knee joint could not reach the maximum knee extension angle at the end of swing phase [15]. Additionally, this control method has not been examined in controlling the multi-joint movements. Although the cycle-to-cycle control can not control the joint angles to follow the target joint angle trajectories continuously, it can achieve the target joint angles at some important points in the swing phase, which will be more effective in generating a successful swing phase of gait than the trajectory-based control. Other advantage of the cycle-to-cycle control is the stimulation data of each patient can be generated automatically in a few cycles.

The designed fuzzy controller for the swing phase was expected to be tested clinically. A gait analysis of the both legs of the hemiplegic patient is required to capture the useful information for detection of the stimulation onset of the swing phase. In order to develop a control method for the entire gait, a proper control method for the stance phase that can be combined with the cycle-to-cycle control for the swing phase has to be studied.

7. Conclusions

The designed fuzzy controller was found to make it possible to apply the cycle-to-cycle control to control multi-joint movements during the swing phase of the FES-induced gait. The controlled swing phase of gait pattern was qualitatively similar to the normal gait pattern. The result implied that the system identification for controller parameter determination could be eliminated and that stimulation data for the open-loop FES control for each patient could be generated automatically after a few walking trial. The results that the fuzzy controller showed better performance than the PID controller in controlling two-joint movements supported the effectiveness of the fuzzy controller. The controller parameter adjustment based on the fuzzy model was also shown to become effective when oscillating response was caused due to inter-subject variability. The designed fuzzy controller for the swing phase was expected to be tested clinically.

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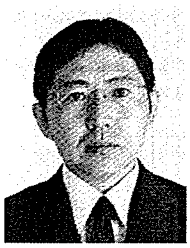

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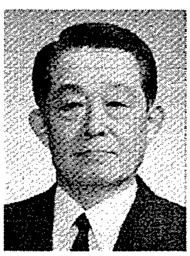

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情報技術の麻痺肢機能再建への応用

渡邉高志
東北大学

Application of Information Technology to Restoration of Functions of Paralyzed Extremities

Takashi Watanabe
Tohoku University

シンポジウム

1．はじめに

脊髄損傷や脳血管障害などにより運動機能の麻痺が生じた場合に，訓練だけでは麻痺肢の機能回復が困難な場合がある．これに対し，機能的電気刺激（FES）は，麻痺した運動機能を再建する手法としての有効性が臨床的に示されてきた[1]．しかし，様々な FES システムがこれまでに開発されてきたが，その臨床的実用性に関しては解決すべき課題が多く，FES システムの臨床応用は限られた対象にとどまっている．ここでは，FES システムを臨床で実用的に使用していくための課題について考え，我々の研究グループで取り組んでいる研究の一例を紹介する．

2．FES システムの課題

FES システムは，中枢的役割を担うコンピュータと，末梢の筋を制御する情報を伝える神経的役割を担う電極とから基本的に構成される．この他，使用者からの命令や制御対象の状態を検出するセンサや電極，麻痺肢やシステムの状態を使用者に伝える情報提示も必要になる．これまでに，中枢部のハードウェアや，神経束単位で扱われる刺激用電極については，概ね実現されてきた．しかし，1本の神経としての電極，制御命令や状態検出のセンサや情報提示などに関するハードウェア，得られた情報の処理や麻痺肢の制御方法などのソフトウェアは十分であるとはいえない．

所望の動作を再現性良く，安定に再建するためには，閉ループ制御を含めることが望まれるが，状態を検出するセンサや多チャネル閉ループ制御アルゴリズムの実現が課題であって，現在も研究段階である．そのため，現在の臨床用 FES システムのほとんどは，開ループ制御を採用している．また，患者は残存機能が制限されているため，FES システムを操作する命令の検出や伝達を行うユーザインターフェイスも実用性に大きく関係するが，BCI のような方法が確立されなければ大幅な改善は見込めないと思われる．

3．高機能化と実用性の改善に向けて

最近の FES システムの研究開発は，ハードウェアを中心に，完全体内埋め込み化と，機能を制限し使い勝手を改善した簡便な表面電極を用いたシステム化の異なった観点で進められているといえる．体内埋め込み型システムが多チャネル制御アルゴリズムとともに実現されれば，インターフェイスを除いて，高機能で実用的なシステムへ展開できると思われる．そこで我々は，中枢部を担う多チャネル閉ループ制御法や，学習型の制御法の開発を進めている[2]．FES 制御法の開発では，被験者での実験的検討が不可欠であるが，それを実施する前にできる限りの問題解決を図ることが望まれる．これに関して，計算機によるモデルシミュレーションの利用が，特に下肢の制御においては，効果的であると考えられる．一方，使用の簡便さから，表面電気刺激システムについても，臨床的実用性の改善に着手している．

FES により運動機能を再建する際には，運動や把持などの感覚の情報とともに，制御装置の情報を使用者である麻痺者に伝えることも望まれている．我々は，視覚や聴覚など，日常生活で頻繁に使用する身体機能を用いるのではなく，皮膚表面電気刺激により生じる感覚を利用して情報提示を行う方法の開発を進めている．より多くの情報を，使用者が直感的に理解できるようにするために，形状パターンのようなものまで伝達できるようにすることを検討している[3]．

4．おわりに

FES システムを情報技術と関連させて考え，実用的利用のための課題と我々のグループでの研究の一例を紹介した．一般的に知られているような情報技術が麻痺肢再建で応用されているとは限らないが，電子的システムと生体との双方向のコミュニケーションを実現するハードウェアとソフトウェアの開発が FES システムにおける課題であり，それらは生体と電子機器とをつなぐ神経工学的な情報技術により実現される一種の人工臓器であると考えることもできる．

謝辞　ここで紹介した研究の一部は，厚生労働科学研究費補助金の補助を受けた．記して感謝する．

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フィードバック誤差学習を用いたFES制御の臨床応用のための実験的検討

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An Experimental Study of FES Control Using Feedback Error Learning for Clinical Application

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1. はじめに

これまでに，フィードバック誤差学習（FEL）を用いた機能的電気刺激（FES）制御法について，2筋の電気刺激による手関節1自由度運動制御の実験[1]や，4筋刺激による手関節2自由度運動制御の計算機シミュレーション[2]により検討し，その実現可能性を示してきた。本報告では，学習の際の被験者への負担を軽減する方法を考慮しつつ，刺激実験により手関節2自由度運動制御を検討した。

2. 方法

Fig.1にFELを用いたFES制御系のブロック図を示す。系への入力は目標角度 $\boldsymbol{\theta}_d$，系の出力は実現角度 $\boldsymbol{\theta}$ である。フィードバック制御器であるPID制御器と，フィードフォワード制御器であるニューラルネットワーク（ANN）の出力の和に，リミッタで刺激最小値と最大値の制限が加えられて最終的な制御器出力となる。

PID制御器のアルゴリズムは式(1)で表される。

$$\boldsymbol{I}_{PID}(n)=\boldsymbol{I}_{th}+\boldsymbol{K}_P e(n)+\boldsymbol{K}_I\sum_{i=0}^{n}e(i)+\boldsymbol{K}_D\left(e(n)-e(n-1)\right)$$

ここで，$\boldsymbol{I}_{PID}(n)$は時刻 n でのPID制御器の出力を，$\boldsymbol{I}_{th}$は筋の刺激最小値を，$e(n)$は関節角度の誤差を，$\boldsymbol{K}_P$，$\boldsymbol{K}_I$，$\boldsymbol{K}_D$はPIDパラメータを表している。ただし，$\boldsymbol{I}_{PID}(n)$が刺激最小値以下で，1時刻前からの変化分が負の場合，積分を行わないという条件を加えてある。ANNは，ニューロン数が入力層36個，中間層18個，出力層4個の3層パーセプトロンを用いた[2]。

被験者は健常者2名とした。目標軌道は，周期が3秒，振幅が掌背屈方向15度，橈尺屈方向10度の円軌道とし，1周期分を学習の1セットとして一括更新によりANNの学習を行った。

3. 結果および考察

Fig.2に学習前後の制御結果の一例を示す。角度軌跡のグラフより，学習前に目立っていた制御の遅れや誤差が，学習後には小さくなっていることが分かる。また，学習前にはPID制御器の出力が大きくANNの出力が小さいが，学習後にはPID制御器の出力が小さくANNの出力が増加している。これらの結果より，円軌道の1周期分を学習の1セットとして用いることでも，FELを用いたFES制御器による多自由度運動制御が実現可能であるといえる。

これまでは，6周期分を学習の1セットとしていたが，学習時の患者への負担を考えると，1周期分を1セットとする方が望ましく，本報告の結果は有意義であるといえる。今後は臨床応用を意識し，円軌道以外の目標軌道の学習や，学習時と異なる軌道への追従制御を検討する予定である。

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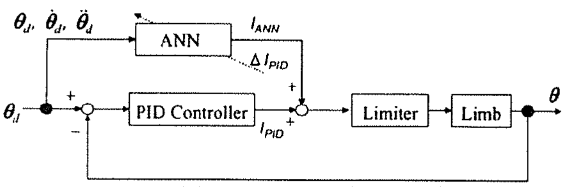
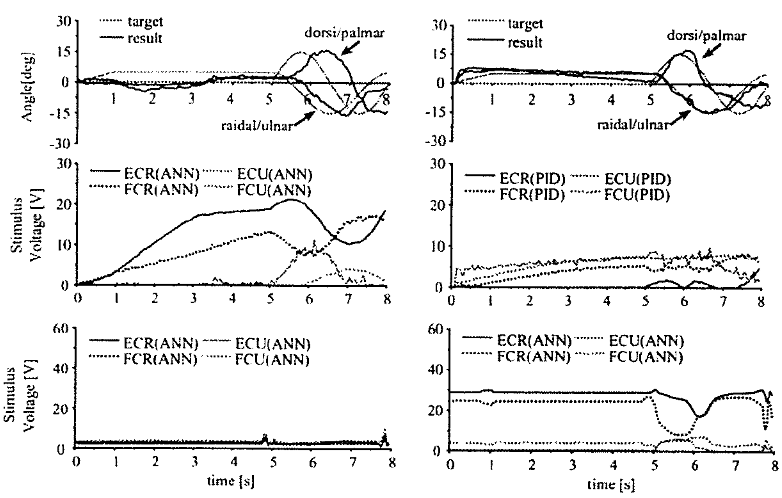


Fig.1 FELを用いたFES制御系のブロック図



(a) 学習1回目 (b) 学習10回目

Fig.2 FELによる学習前後の制御結果．上から角度軌跡，PID制御器の出力，ANNの出力を示す。

オーガナイズドセッション

電気刺激による皮膚感覚を用いたパターン提示のための電気刺激パラメータに関する検討

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A study on electrical stimulation parameters for pattern presentation using cutaneous sensation elicited by electrical stimulation

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1．はじめに

皮膚表面電気刺激により生じる感覚を利用して，義肢や機能的電気刺激（FES）使用者に対する情報提示などを行う方法の実現を目的としている．我々は，マトリクス状に配置した点電極を用いて移動感覚を生成させることで，複数の提示パターンの識別が可能になることを示唆してきた[1-2]．本報告では，不快感を生じないような刺激パルスや，電気刺激提示の時間間隔がパターン識別に及ぼす影響について検討した．

2．方法

(1)刺激波形の違いによる刺激位置の識別

周期10msの単極矩形波（0.2ms幅），両極矩形波（正負各0.2ms幅），バースト波（正負各0.02ms幅の両極矩形波0.4ms）の3種類について，2点の電気刺激位置の識別を健常被験者3名の前腕部内側で行った．5×5のマトリクス状電極（直径1.2mmステンレス線，電極中心間距離4.0mm）の縦4個，横4個を用いて，4つの電極のうち1つに電気刺激を1s間印加し，1sの休止時間を挟んで4つのうちのどれかの電極へ1s間刺激を印加し，刺激電極が同一かどうか回答させた．横方向と縦方向の各々で，電極間距離0, 4, 8, 12mmの組み合わせをランダムに行った．刺激振幅は，被験者が判断しやすい値に設定した．

(2)移動感覚における電気刺激感覚の残存の影響

図1(a)に示す2本の線の提示パターンの各々について，健常被験者4名で刺激感覚の残存の影響を調べた．被験者には提示するパターンを示し，2sの間隔で刺激を2回与え，1番目と2番目の提示パターンが同じかどうかを回答させた．また，それらがどのような形のパターンとして感じられたかも回答させた．1番目は，1本目と2本目の刺激の間隔（ISI）を0sとし，2番目は，1番目に提示したパターンのISIを0, 0.1, 0.33, 0.66, 0.99sのいずれかにランダムに設定した．刺激振幅は，被験者が判断しやすい値に設定した．

3．結果及び考察

(1)各波形での正答率は，単極矩形波84.0±14.3%，両極矩形波79.5±18.6%，バースト波79.2±14.2%であり，波形の違いによる刺激位置の識別結果に差は見られなかった．一方，全被験者から，バースト波がぼんやりとしてわかりにくいという意見があった．また，矩形波の場合は単極も両極もチクチクとした感覚であり，バースト波の場合は痛みや不快感が矩形波に比べ少ないように感じる被験者が多かった．バースト波を用いることで不快感が軽減する傾向があったが，受容感覚が不明確になる傾向もあり，皮膚電気刺激感覚による情報提示にはバースト波は適さない可能性があることも示唆される．

(2)図1(b)に，パターン(ii)に対する各ISIでの正答率を示す．被験者により差はあったが，ISI=0.1s, 0.33sの場合にはISI=0sの場合との違いが不明確になっており，1本目の刺激が2本目に影響を及ぼすことが示唆される．パターン(i)についてもほぼ同様であった．被験者からは，ISI=0sの場合，2本の線が1本の線に感じるとの意見があった．今回の実験では，1個の電極を0.33s間電気刺激して移動感覚を生成させており，その場合，0.33sより長いISIを用いることで，最初に提示された移動感覚の影響を受けなくなると推測される．

謝辞　本研究の一部は，厚生労働科学研究費補助金の補助を受けた．記して感謝する．

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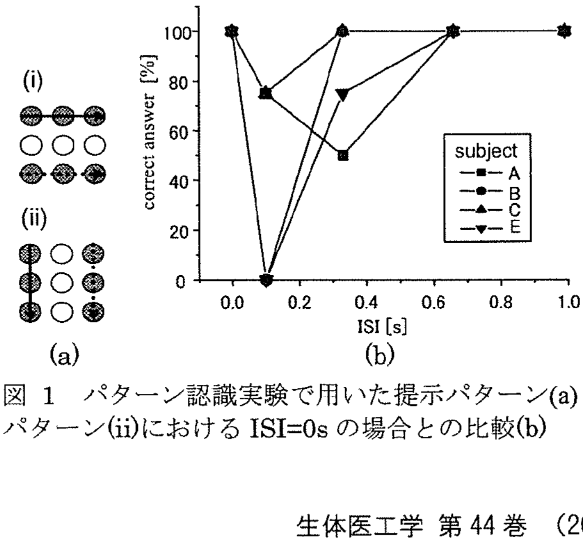

図1　パターン認識実験で用いた提示パターン(a)とパターン(ii)におけるISI=0sの場合との比較(b)

一般演題

足動作のイメージを脳波から検出する BCI システムの基礎的検討

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A basic study on brain-computer interface based on EEG during foot movement imagery

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1. Introduction

To provide the better communication to the severely paralyzed patients, the brain-computer interface (BCI), that extracts the patients' motor imagery from ongoing EEG was studied.

2. Method

Six healthy subjects took part in the experiments. From each subject, bipolar EEG (CZ-FCZ) was measured by Ag-AgCl electrodes. In the experiments, subjects were requested to imagine foot movement during the presence of the cue (0 - 6 s) onto the LCD display of the computer.

The beta EEG activity between 20-30 Hz was analyzed. To estimate the band power, the acquired signal was filtered, squared and smoothed (moving average: time window 1 s). The calculated power value was fed back to the subject in real time as the length of the bar displayed on the screen.

In previous BCI studies, it was common to take a mathematical method e.g. pattern separation, to classify features from preprocessed EEG data. But in this study, the following method [1] was taken for the event detection of motor imagery ("Brain Switch"): the event is detected if the band power in the upper beta range exceeds the threshold value for a certain period of time (dwell time). To avoid undesired detections (false positive), the successive event detection was suppressed by taking refractory period into account.

3. Results and Discussion

From two subjects out of six, it was shown by the time-frequency analysis of the EEG, the following ERS, ERD (event-related (de)synchronization [2]), i.e. increase/decrease of the band power on the specific frequency range, were observed: (a) ERD after onset of motor imagery on lower beta band (20 - 25 Hz), which was related to motor planning. (b) sustained ERS on upper beta band (25 - 30 Hz), (c) sustained ERD on mu band, and (d) ERS after offset of motor imagery (rebound) on lower beta band. Whereas the EEG was suppressed on the rest four subjects.

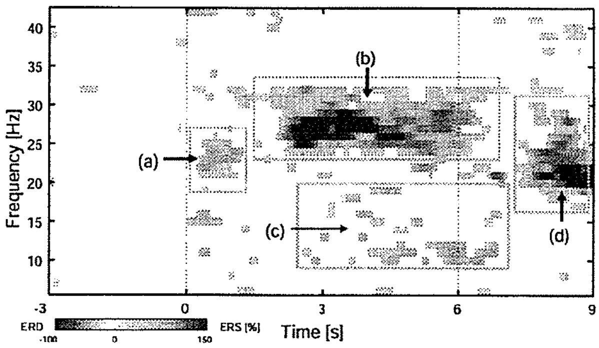

Fig.1 An example of averaged activity evoked by foot movement imagery. (a and c) ERD, (b and d) ERS

The two former subjects participated to further experiments with the aim to detect foot motor imagery, based on the proposed detection method above. By these experiments, it was shown that the true event (true positive) of foot motor imagery was detected with a provability of 60 - 90%.

Although the proposed method could detect only the presence or absence of foot movement imagery, its algorithm was very simple and only the three electrodes (one bipolar channel) were required for measurement. The "Brain Switch" method may be a BCI system that is easily applicable to the patients. The investigation of training effect and development of good training strategy to improve the S/N ratio of the EEG are left to the future study.

一般演題

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眼電図を用いたメニュー選択型インターフェースのための基礎的検討

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A basic study on human-computer interfacing system for menu selection using EOG

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1. はじめに

残存機能が著しく限られた四肢麻痺患者へコミュニケーション手段を提供するために，眼球運動に伴って生じる眼電図（EOG）から視線移動の方向や距離を検出する患者・コンピュータ間のインターフェースの検討を行った．EOGから視点を検出するには，EOGを直流増幅することが望ましい [1]．しかし微小信号の低周波増幅は，増幅器のドリフト特性などのために容易ではない．そこで本研究では，交流増幅した水平・垂直方向のEOGから8方向への長短の視線移動を検出するシステムを提案し，健常被験者に対する動作確認実験を行った．

2. 実験方法

3名の健常被験者に対して，視線移動時のEOG計測実験を行った．被験者は，キューの呈示とともに液晶ディスプレイ上に表示されるターゲットに，注視線から視線を速やかに移動させ，移動後はターゲットを注視し続けることを求められた．ターゲットは，注視線から視野角5°，10°の円上の各8点（右，右上，上，左上，左，左下，下，右下）とした．

眼球の上下，左右にAg-AgCl電極を設置し，生体信号増幅器を用いてEOGを双極誘導で計測した．増幅後のデータは0.5〜100Hzに帯域制限した後，標本化周波数250Hzでパーソナルコンピュータに取り込み，解析を行った．

得られた垂直・水平方向のEOG（それぞれ V_v, V_h）から計算されるノルムが，試行中の最大値の80%以上となるサンプルデータから重心を計算した．重心の2次元ベクトルをk近傍決定則によって分類し，EOGから被験者の視線移動の方向（8通り）と距離の長短（視野角の大小）の組み合わせを検出できるかを検討した．

3. 結果

試行中に計測したデータ点 (V_h, V_v) の時間経過による軌跡を観察した．視線の移動方向や距離によって異なる軌跡が得られ，原点からターゲットの位置に対応する点との間で軌跡が直線状になる場合が一般的であった．被験者や視線移動方向によっては，軌跡が直線状にならない場合もあった（図1）．これは視線の移動が直線的ではなかったためと推測される．しかしこのようなケースでも，本研究で提案した方法によって選択されたデータ（図1中の丸印）から重心（同・三角印）が算出されるため，視線の移動軌道のばらつきによらず安定した検出結果が得られた．

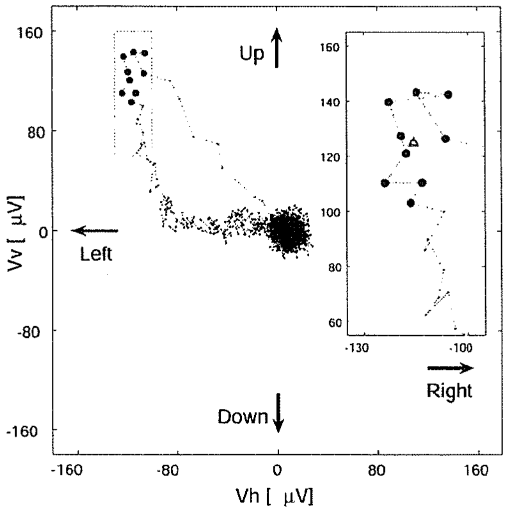

図1：左上への視線移動の際の (V_h, V_v) の軌跡の例

視野角10°の円上8点をターゲットとしたもの，および視野角を2通りとして視線移動距離の異なる計16点としたものの2種類の実験を行った．両者の実験とも，概ね90%以上の正答率を達成した．判別の誤りは，視線移動距離の長短によるものが主であり，隣接する方向への誤判別がそれに続いた．

本研究で提案した手法は厳密なEOGを直流増幅することなく，簡易な計測で高い正答率を得られるものである．四肢麻痺患者に対するメニュー選択型のヒト・コンピュータ間インターフェースへの応用が可能である．

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一般演題

Nonlinear FES Control of Knee Joint by Inversely Compensated Feedback System

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Abstract: The aim of applying Functional Electrical Stimulation (FES) is to restore a person's motor function by directly supplying the controlled electrical currents to the site of the paralyzed muscles. However, most clinically utilized FES systems have adapted an open-loop control scheme. Recently the closed-loop control scheme has been considered for setting up the FES system, but due to the inherent nonlinearities in the musculoskeletal system, the nonlinearities were not fully compensated and it caused the oscillatory responses for tracking the output variables. In this study, a nonlinear controller model that has two inverse compensation units is proposed with the compromising feedback linearization method and this will eventually be used to design the FES control system for stimulating a knee joint musculoskeletal system.

Keywords: Feedback linearization, FES, inverse compensation, knee joint musculoskeletal system, nonlinear control.

1. INTRODUCTION

Functional Electrical Stimulation (FES) is defined as "the electrical stimulation for assistance or reconstruction of biological functions, with clear purpose and understanding of the mechanism" [1-3]. It is an effective method for restoring motor functions to the limbs paralyzed by spinal cord injury (SCI) or cerebral apoplexy. It utilizes the controlled electrical currents to evoke a certain skeletal muscle contraction for the paralyzed patients by supplying the proper electrical pulse trains to the intact muscles. However, due to the highly nonlinear nature of the musculoskeletal system, most clinical FES systems have been considered as only open-loop control schemes to stimulate a specific pattern predetermined by the relevant medical experts. [4,5]. Also, the clinical oriented FES systems have employed the open-loop control scheme because the closed-loop feedback type is difficult to implement when attaching the proper sensors so that good reproducibility at every attachment is guaranteed [6]. Indeed, the closed-loop EES clinical system induces rapid muscle fatigue [7], spinal reflexes and spasticity [8]. Also, due to the time-varying and unstable characteristics of the muscle [9], more difficulties are imposed in identifying the musculoskeletal system [10,11]. For these reasons, the closed-loop feedback control approaches have been applied to the FES clinical system only recently [12,13]. However, these efforts were incapable of compensating the inherent nonlinearities contained in the musculoskeletal system and consequently the controllers often caused the unstable oscillatory responses because they were not suitable for the overall range of a patient's motions such as FES standing, walking or cycling. In [14,15], a neural network based inverse model system trained with the complex nonlinear mapping for the feed-forward control was proposed for establishing a PID feedback controller and yielded the better performance. However, the stability issue remained unresolved due to the black-box nature of the neural network. Moreover, its nature does not provide any intermediate variable or clue about the physiological process. Effort was also put forth to apply the

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