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),

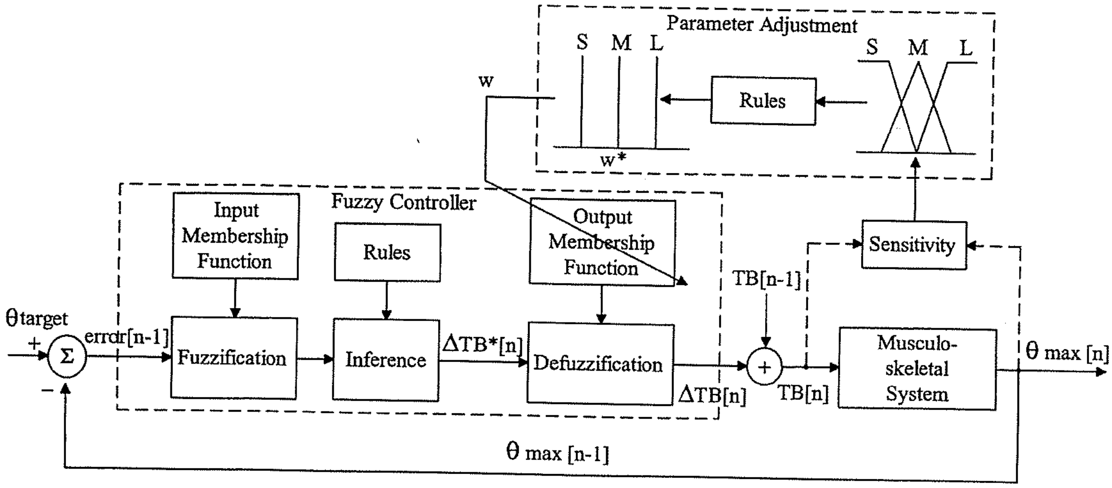

Fig. 3 Structure of the fuzzy controller with the parameter adjustment based on the fuzzy model. In this figure, the maximum joint angle is an example of the controlled joint angle.

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

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),

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

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).

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

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.

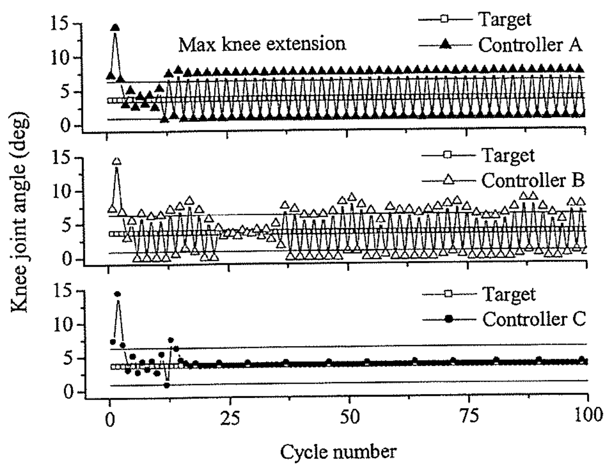

Fig. 4 Responses of the fixed fuzzy controller (controller A), the gradient descent based adaptive fuzzy controller (controller B), and the adaptive fuzzy controller realized based on the fuzzy model (controller C) in suppressing the oscillation of the maximum knee extension angle with the very strong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

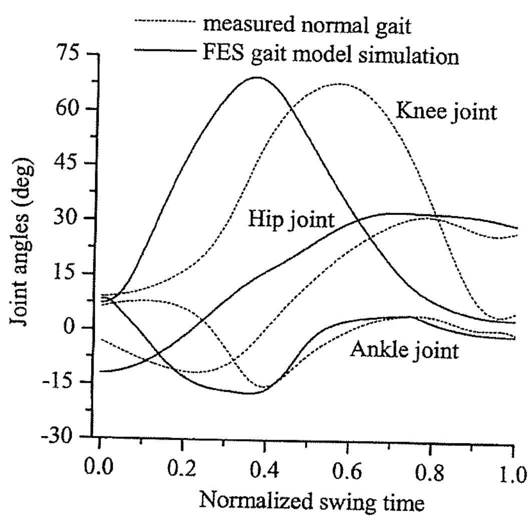

Fig. 5 Trajectories of the controlled joint angles of the very strong subject compared to the trajectories of the joint angles measured from the normal subject.

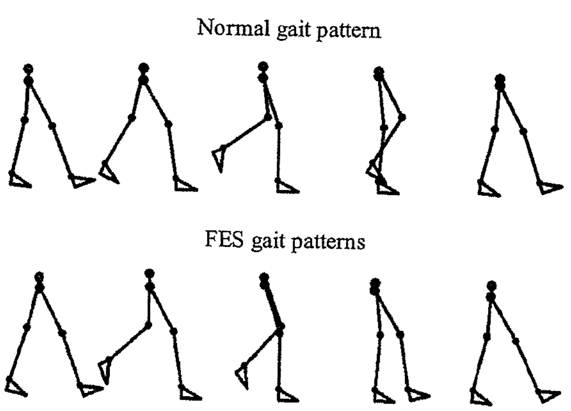

Fig. 6 Stick picture of the FES-induced gait pattern and the normal gait pattern. The black leg is the controlled swing leg and the gray leg is the stance leg.

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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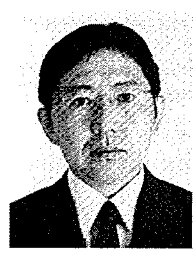

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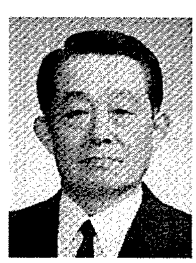

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情報技術の麻痺肢機能再建への応用

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。

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

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，$\boldsymbol{e}(n)$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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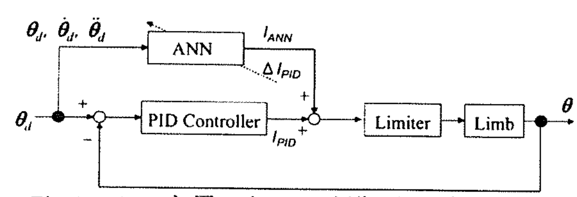

Fig.1　FEL を用いた FES 制御系のブロック図

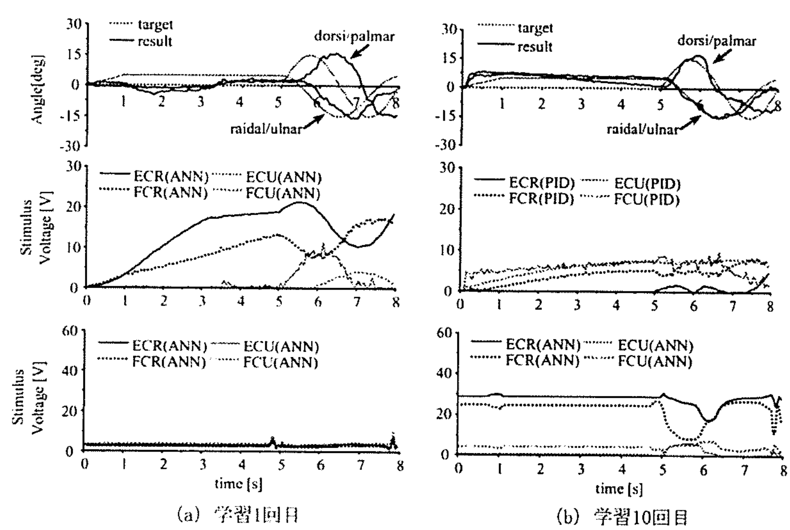

Fig.2　FEL による学習前後の制御結果．上から角度軌跡，PID 制御器の出力，ANN の出力を示す。

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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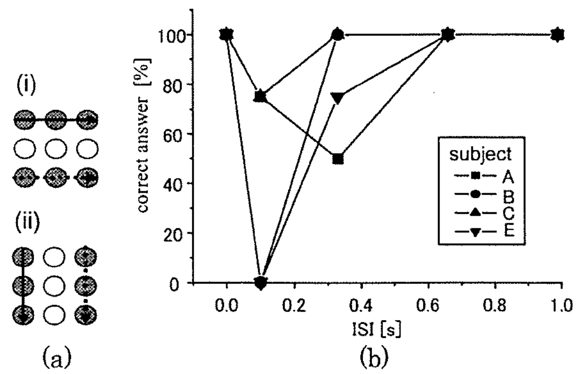

図 1　パターン認識実験で用いた提示パターン(a)とパターン(ii)における ISI=0s の場合との比較(b)

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.

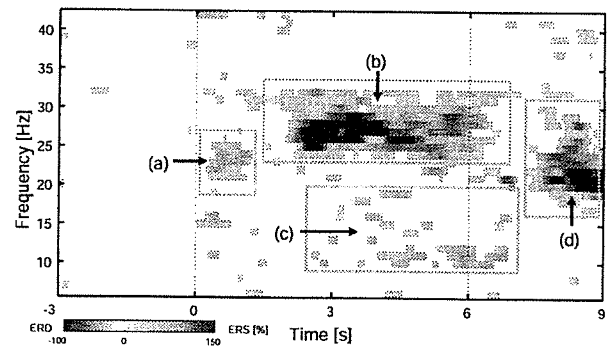

Fig.1 An example of averaged activity evoked by foot movement imagery. (a and c) ERD, (b and d) ERS

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.

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一般演題

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．

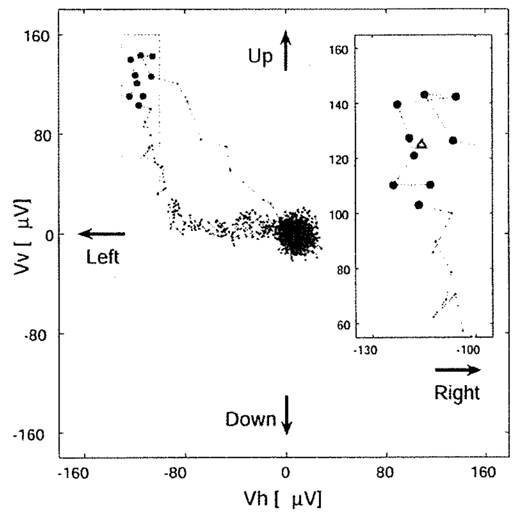

図1：左上への視線移動の際の (V_h, V_v) の軌跡の例

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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physiologically based model for modeling an inverse system [12]. However, only the partial aspects of the nonlinearities were compensated and consequently the output system still had some oscillatory responses if the extent of a certain posture of FES motions such as standing or walking exceeds the range of the compensated linearity. Thus, we try to completely compensate the nonlinearities of a knee joint musculoskeletal model through feedback linearization with inverse compensation scheme and to suggest a new linear control scheme for tracking the reference trajectory of FES posture related with knee joint movements.

2. KNEE JOINT MUSCULOSKELETAL MODEL

The musculoskeletal system model of a knee joint with quadriceps muscle, which has an important role in FES standing, walking and cycling, is depicted as three blocks as shown in Fig. 1.

In Fig. 1, the first block contains the recruitment feature of muscle fibers, $r(s)$ stimulated by the intensity s, which is formulated in (1).

$$r(s) = s_c \tanh(s_h(s - x_c)) + y_c . \quad (1)$$

Here, the parameters of a rising and a falling path are not identical and they represent the different hysteretic characteristics depending on the path, i.e.,

i) On a rising path: $s_c = s_{rc}, s_h = s_{rh}, x_c = x_{rc}, y_c = y_{rc}$,

ii) On a falling path: $s_c = s_{fc}, s_h = s_{fh}, x_c = x_{fc}, y_c = y_{fc}$.

The first block also includes the 1st order activation dynamics of the muscle's normalized active state corresponding to the calcium release. The activation dynamics, *a(t)* is described by

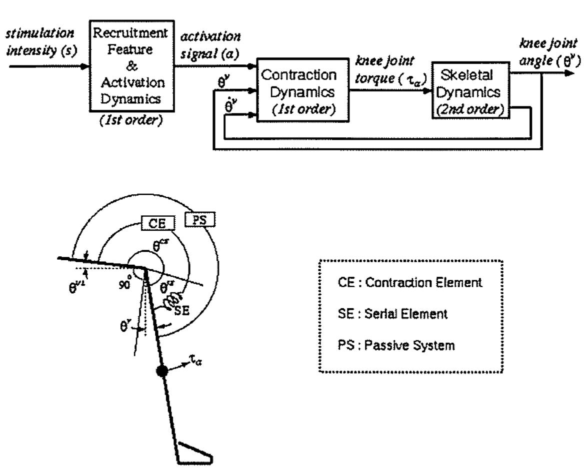

Fig. 1. A musculoskeletal system model of a knee joint.

$$\dot{a} = (r-a)r/\tau_r + (r-(a-a_{\min})-(r-a)r)/\tau_f . \quad (2)$$

Here, the time constants of τ_r and τ_f are the different rates of calcium release to and uptake from muscle fibers, respectively and $a_{\min}$ is the minimal active state.

The second block in Fig. 1 includes the 1st order muscle contraction dynamics. We adopted a lumped model of musculotendon and a moment arm as indicated in the lower schematics of Fig. 1 to refer the muscle to a torque generator [13]. The contraction dynamics of the musculotendon was stated as

$$\dot{\tau}_a = (k_2\tau_a + k_1k_2\tau_{\max})\bullet \left(-\dot{\theta}^V - \dot{\theta}^{CE}_{\max} g^{-1}_{CE}\left(\tau_a/\left(\tau_{\max}k_{CE}(\theta^{CE})a\right)\right)\right), \quad (3)$$

where

$$\theta^{CE} = 3\pi/2 - (\theta^V + \theta^{UL}) - \ln\left(\tau_a/\left(\tau_{\max}k_1\right)+1\right)/k_2 ,$$

$$k_1 = 1/\left(\exp(sh^{SE})-1\right), \quad k_2 = sh^{SE}/\theta^{SE}_{\max},$$

$$g^{-1}_{CE}(X) = \begin{cases} \dfrac{sh^{CEg}_{low}(X-1)}{X + sh^{CEg}_{low}} & (0 \le X < 1) \\ \dfrac{x_s sh^{CEg}_{up}(X-1)}{-X + y_s sh^{CEg}_{up} + y_s + 1} & (1 \le X < 1.3), \end{cases}$$

$$k_{CE}(Y) = \begin{cases} \exp(-(\dfrac{Y-\theta^{CE_L}_{\max}}{sh^{CEk_L}})^2) & (Y < \theta^{CE}_{\max}) \\ (1-c_1)\exp(-(\dfrac{Y-\theta^{CE_R}_{\max}}{sh^{CEk_R}})^2) \\ \quad + sl^{CEk}(Y-\theta^{CE_R}_{\max}) + c_1 & (\theta^{CE}_{\max} \le Y). \end{cases}$$

Here, τ_a is the active muscle torque, k_1 and k_2 are the parameters of series elastic element (SE). $\dot{\theta}^{CE}_{\max}$ is the maximum contraction velocity of the contractile element (CE), $g^{-1}_{CE}()$ is the inverse function of torque-angular velocity relationship of the CE, $\tau_{\max}$ is the maximum muscle torque and $k_{CE}()$ is the torque-angle relationship of the CE.

The third block of Fig. 1 can be described by the following 2nd order skeletal dynamics:

$$\ddot{\theta}^V = 1/I\left(\tau_a - G\sin(\theta^v) - D\dot{\theta}^V + s_1(e^{-s_2(\theta^V+\theta^{UL})}-1)\right). \quad (4)$$

Here, θ^v is the knee joint angle with reference to the vertical line, I and G are the moment of inertia and the gravity constant of the lower leg, respectively. The latter term of (4), $-D\dot{\theta}^V + s_1(e^{-s_2(\theta^V+\theta^{UL})}-1)$ represents the damping and elastic torque induced by the

Table 1. Knee joint, vastus lateralis muscle parameters used in our computer simulation.

s_{rc}	1.661	τ_r	0.02	$sh_{low}^{CE_s}$	0.25	sh^{CEk_R}	0.2
s_{rh}	2.346	τ_f	0.2	$sh_{up}^{CE_s}$	0.25	sl^{CEk}	-0.18
x_{rc}	1.143	$a_{\min}$	0.001	x_s	0.2	$\theta_{\max}^{CE}$	3.491
y_{rc}	1.538	sh^{SE}	2.7	y_s	0.3	I	0.44
s_{fc}	0.516	$\theta_{\max}^{SE}$	0.384	$\theta_{\max}^{CE_L}$	3.491	G	16.5
s_{fh}	7.054	θ^{UL}	0.154	sh^{CEk_L}	0.2	D	0.22
x_{fc}	0.792	$\tau_{\max}$	11.32	c_1	0.75	s_1	0.092
y_{fc}	0.536	$\dot{\theta}_{\max}^{CE}$	18	$\theta_{\max}^{CE_R}$	3.456	s_2	5.09

passive system, such that D, s_1, s_2 are the constants and θ^{UL} is the angle of the upper leg with reference to the horizontal line. The muscle contraction and skeletal dynamics stated in (3) and (4) can be alternatively expressed by the state space representation form as stated in (5) if we define $x_1 \equiv \tau_a$, $x_2 \equiv \theta^V$, $x_3 \equiv \dot{\theta}^V$ and the output variable, $y \equiv \theta^V$.

$$\begin{aligned}
\dot{x}_1 &= (k_2 x_1 + k_1 k_2 \tau_{\max})\cdot \\
&\quad \left(-x_3 - \dot{\theta}_{\max}^{CE} g_{CE}^{-1}\left(x_1/\left(\tau_{\max} k_{CE}(x_1,x_2)a\right)\right)\right), \\
\dot{x}_2 &= x_3, \\
\dot{x}_3 &= 1/I\left(x_1 - G\sin(x_2) - Dx_3 + s_1\left(e^{-s_2(x_2+\theta^{UL})} - 1\right)\right), \\
y &= x_2.
\end{aligned} \tag{5}$$

Thus the overall plant (musculoskeletal system) dynamics can be interpreted as a 4th order system composing of i) one activation dynamic unit, ii) one contraction dynamics unit, iii) two skeletal dynamics units. For our computer simulations, we adopted the estimated knee joint and vastus lateralis muscle parameters that are identified by the lump parameters for knee extensors for the voluntary contraction and for knee joints with the vastus lateralis muscle, respectively as stated in Table 1.

The identification procedures determining muscle parameters shown in Table 1 are listed in [13,16] and the accuracy of the model parameters are evaluated by tracking the model-predicted joint angle trajectories compared with experimental data [13].

3. NONLINEAR CONTROLLER DESIGN

The main control task of our study is to force a certain knee joint angle to track the reference trajectory. The overall control system structure is shown in Fig. 2.

The central idea of our suggested controller is utilizing a feedback linearization. In other words, we approximate a nonlinear system dynamics into a linear one with applying the algebraic transformation. With this scheme, the inherent nonlinearities contained in a musculoskeletal system model of the knee joint are vanished, and consequently the relatively simple linear control scheme can be applied. In Fig. 2, two inverse compensation units for the feedback linearization are described [17]. The first inverse compensation unit as stated in the following equation is to eliminate the nonlinearity of a term, $g_{CE}^{-1}\left(x_1/\left(\tau_{\max} k_{CE}(x_1,x_2)a\right)\right)$, which represents the muscle contraction dynamics.

$$r_{in} = \begin{cases}
\dfrac{-x_1 u_{in} + sh_{low}^{CE_g}\dot{\theta}_{\max}^{CE} x_1}{sh_{low}^{CE_g}\tau_{\max} k_{CE}(x_1,x_2)u_{in} + sh_{low}^{CE_g}\dot{\theta}_{\max}^{CE}\tau_{\max} k_{CE}(x_1,x_2)} \\
\qquad \left(0 \le \dfrac{x_1}{\tau_{\max} k_{CE}(x_1,x_2)a} < 1\right) \\
\dfrac{x_1 u_{in} + x_s sh_{up}^{CE_g}\dot{\theta}_{\max}^{CE} x_1}{((y_s sh_{up}^{CE_g} + y_s + 1)\tau_{\max} u_{in} + x_s sh_{up}^{CE_g}\dot{\theta}_{\max}^{CE}\tau_{\max})k_{CE}(x_1,x_2)} \\
\qquad \left(1 \le \dfrac{x_1}{\tau_{\max} k_{CE}(x_1,x_2)a} < 1.3\right)
\end{cases} \tag{6}$$

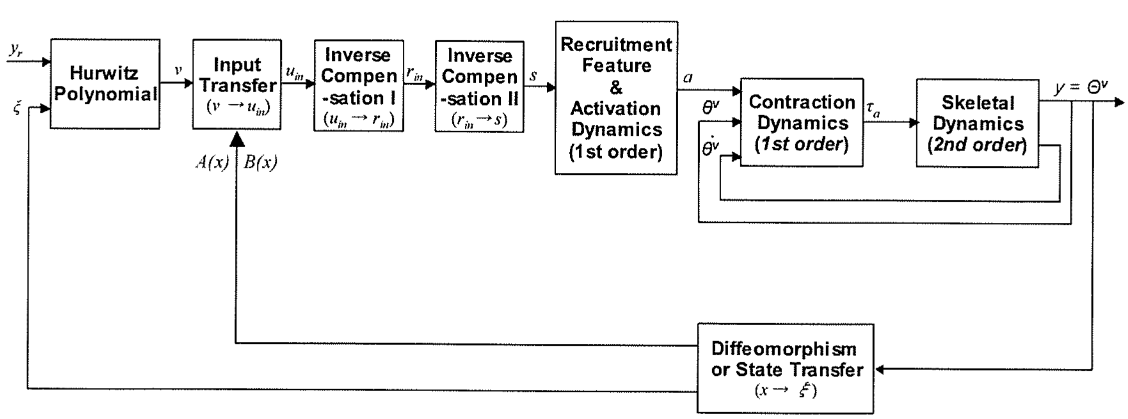

Fig. 2. The proposed control system structure.

In (6), u_{in} is the control input and required recruitment r_{in} is the output. The second inverse compensation unit as stated in the following equation is to remove the nonlinearity of the recruitment.

$$s = \ln\left(\left(r_{in} + (s_c - y_c)\right)/\left(-r_{in} + (s_c + y_c)\right)\right)/2s_h + x_c. \quad (7)$$

In (7), r_{in} is the input and the normalized stimulation intensity, s is the output. Since the normalized active state a in (2) is not measurable, it is regarded as a bounded and structured uncertainty. Then, the diffeomorphism (state transfer) [18] is defined as in (8) and the control law (input transfer) u_{in} is defined as in (9), respectively.

$$\begin{aligned} \xi_1 &\equiv y = x_2, \\ \xi_2 &\equiv \dot{y} = x_3, \\ \xi_3 &\equiv \ddot{y} = x_1/I - G\sin(x_2)/I - Dx_3/I \\ &\quad + s_1(\exp(-s_2(x_2 + \theta^{UL})) - 1)/I, \end{aligned} \quad (8)$$

$$u_{in} = (v - B(x))/A(x), \quad (9)$$

where

$$\begin{aligned} A(x) &\equiv -(k_2 x_1 + k_1 k_2 \tau_{\max})/I, \\ B(x) &\equiv (k_2 x_1 + k_1 k_2 \tau_{\max})(-x_3 + 1)/I - G\cos(x_2)x_3/I \\ &\quad - D(x_1/I - G\sin(x_2)/I - Dx_3/I \\ &\quad + s_1\left(\exp(-s_2(x_2 + \theta^{UL})) - 1\right)/I)/I \\ &\quad + s_1(-s_2\exp(-s_2(x_2 + \theta^{UL}))x_3)/I. \end{aligned}$$

(9) is derived by combining the diffeomorphism of (8) and the control law. The input to the control law v is selected as stated in (10) and (11),

$$\dddot{y} = v, \quad (10)$$

$$v = \dddot{y}_r + \alpha_2(\ddot{y}_r - \xi_3) + \alpha_1(\dot{y}_r - \xi_2) + \alpha_0(y_r - \xi_1), \quad (11)$$

where $\alpha_2, \alpha_1, \alpha_0$ are the coefficients of the Hurwitz polynomial $s^3 + \alpha_2 s^2 + \alpha_1 s + \alpha_0$. The Hurwitz polynomial coefficients are chosen with considering the behavior and response of the closed-loop system in terms of overshoot, rising time and settling time. Since the degree of the control system is three, we need to differentiate the output three times to get the control input.

The reference trajectory $y_r(t)$ was set as a sinusoidal function, which is useful in clinical practice because it is often used for paralyzed muscle training. The simulation result is shown in Fig. 3. With the inverse compensations of nonlinearities and the feedback linearization stated above, the stable output tracking of the overall closed-loop system was successfully achieved.

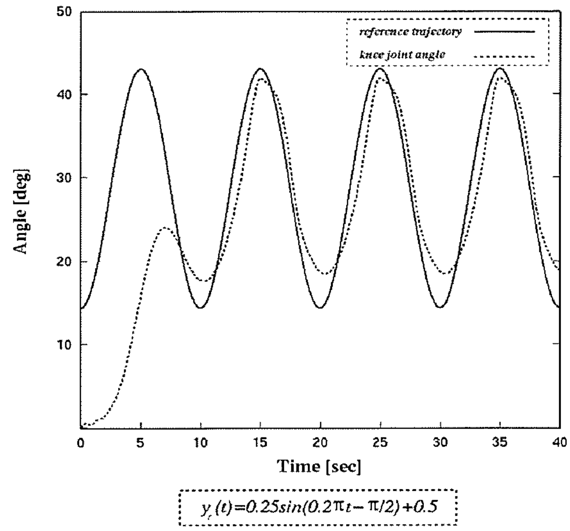

Fig. 3. The simulation result of FES nonlinear control.

4. CONCLUSIONS

As we mentioned earlier, the recent closed-loop control schemes for FES do not completely compensate the nonlinearities of the plant and consequently often lead to the oscillatory responses. Our control plant shows better performance in terms of output tracking. Also, the output of the control system achieves the stable conditions by utilizing a feedback linearization with considering inverse compensations. Thus, we believe that we have suggested a new promising application especially for controlling the inherent nonlinearity of the FES system. In other words, we propose a new nonlinear control scheme to track a referenced angle trajectory of the paralyzed knee joint with implementing a controller composing two linearization units.

For the real clinical application of our control scheme, the following two issues must be considered. Firstly, the model parameters must be easily identified so that we can minimize the physical and psychological burdens of a patient. Secondly, the control scheme must be improved to be adaptive to time varying factors in the musculoskeletal system such as muscle fatigue.

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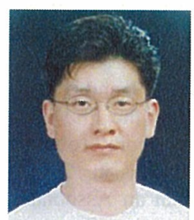

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A Basic Study of Fuzzy Controller for Cycle-to-Cycle Control of Knee Joint Movements of FES Swing: First Experimental test with a Normal Subject

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Abstract

A feasibility of the fuzzy controller using cycle-to-cycle control for multi-joint movements of swing phase of FES-induced hemiplegic gait had been shown in our research group by computer simulation with musculoskeletal model. It was required to show the effectiveness of the fuzzy controller experimentally. In this study, a fuzzy controller using the cycle-to-cycle control method was tested experimentally in controlling the knee extension stimulating the Vastus using surface electrode stimulation. Experimental results with a healthy subject suggested that the fuzzy controller could realize the target maximum knee extension angle during walking.

1. INTRODUCTION

In the human motor functions, the movements of lower limbs during gait are complex multi-joint movements. So, an excellent FES control strategy is required to restore the paralyzed gait using FES. Our research group has studied an FES controller to develop the swing phase of hemiplegic gait in computer simulation towards improving gait ability.

We proposed the fuzzy controller based on the cycle-to-cycle control method for controlling swing phase of hemiplegic gait induced by FES [1]. Computer simulation study showed that the controller was feasible to control multi-joint (hip, knee, and ankle) movements of swing phase of FES-induced hemiplegic gait.

In this study, we focused on testing an effectiveness of the controller experimentally. As the basic study, experimental tests of the cycle-to-cycle control method were performed in controlling the knee extension angle with a normal subject.

2. METHODS

2.1 Control Algorithm

In the cycle-to-cycle control, stimulation burst duration TB is adjusted, while pulse amplitude, pulse width, and frequency of stimulation pulse are fixed. The adjustment of the burst duration TB is based on the performance of the previous cycle. Basic algorithm of the cycle-to-cycle control is shown in equation (1),

$$TB[n] = TB[n-1] + \Delta TB[n] \qquad (1)$$

where $TB[n]$ is the stimulation burst duration for the current cycle, $TB[n-1]$ is the stimulation burst duration of the cycle just before the current one, and $\Delta TB[n]$ is the output of the controller. Taking a controlling of the maximum knee extension angle for example, maximum knee joint angle obtained in last control is used as a feedback signal. Error is defined as difference between target maximum joint angle and the obtained maximum joint angle.

2.2 Experimental Method

Knee joint angle of the left leg of a healthy subject (22 years old, male) was controlled by stimulating the Vastus (Vastus medialis, Vastus lateralis). The pulse width and the pulse frequency were 200μs and 20Hz, respectively. Electrical stimulation was applied to the muscles through the isolator and surface electrodes (F-150, Nihon Koden). The knee joint angle was measured with the goniometer (M180, Penny & Giles).

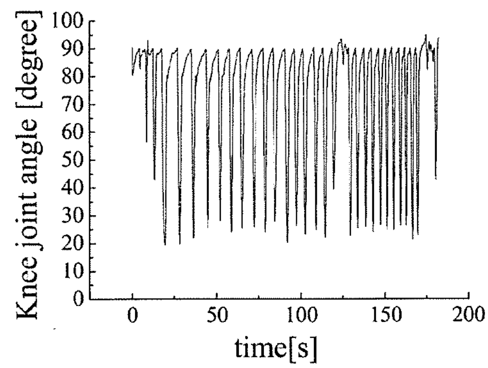
The subject sat on the chair that his leg didn't reach the ground and relaxed his legs during experiments. Before the controlling of the knee joint angle, maximum pulse amplitude was determined in order to get enough control range without pain, which was 50V in this case. The desired value of the maximum knee extension angle was set to 25 degree. A knee joint angle of 0 degree was defined as full extension. The knee joint angle at rest was about 90 degree. The cycle-to-cycle control was performed three times.

3. RESULTS

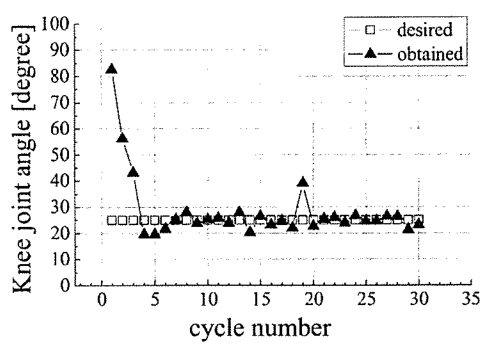
Figure 1 shows the result of the first trial of controlling the knee joint angle. The entire angle trajectory of the knee joint is given in Figure 1(a) with time on the x-axis and the knee joint angle on the y-axis. The obtained value of maximum joint angle at each cycle is plotted in Figure1 (b) with cycle number on the x-axis. Figure1(c) represents stimulation burst duration of each cycle.

In the first trial, although the result of the 19th cycle was not good, the maximum knee extension angle was controlled well most of the controlled cycles (Figure 1(b)).

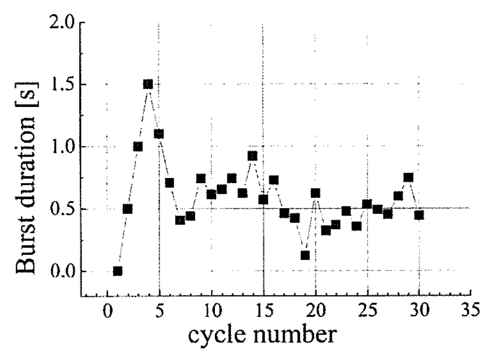
Figure 2 shows result of the second trial of controlling the knee joint angle. In the second and the third trial, there were a few uncontrolled



(a) entire trajectory of joint angle

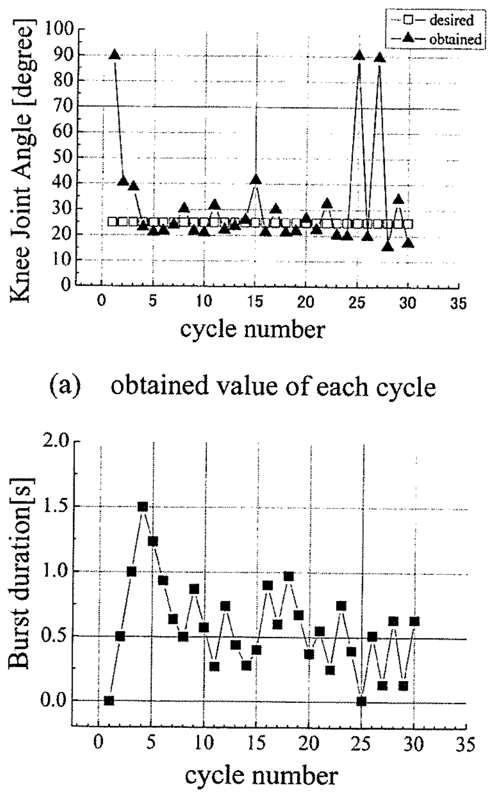


(b) obtained value of each cycle



(c) burst duration

Figure 1 Results of the first trial of the cycle-to-cycle control of the knee joint angle. The maximum knee extension angle was controlled.



(a) obtained value of each cycle

(b) burst duration

Figure 2 Results of the second trial of the cycle-to-cycle control of the knee joint angle.

cycles. They appeared in the cycle over 20th control as shown in Figure 2(a). However, the fuzzy controller recovered this uncontrolled situation by adjusting stimulation burst duration.

4. DISCUSSION AND CONCLUSION

Results of this paper suggested that the fuzzy controller using the cycle-to-cycle control would be effective in clinically. In this study, knee joint angle was controlled by stimulating single muscle (the Vastus). Our previous work using computer simulation showed that the controller could control multi-joint (hip, knee, and ankle) movements [1]. So, it is necessary to test the controller in controlling multi-joint movements experimentally.

In the experimental results, the knee joint didn't move sometimes (Figure 2(a)). The reason that the knee joint didn't move at the 25th cycle was considered that burst duration was very small. The maximum knee joint angle of the 24th cycle reached the target maximum joint angle, resulting in large reduction of the burst duration of the 25th cycle by the fuzzy controller. It is necessary to adjust the parameters of the fuzzy membership function for each subject.

On the other hand, muscle fatigue was considered to be a reason of the uncontrolled situation at the 27th cycle. Gradual decrease of muscle force caused by muscle fatigue was found to be compensated by the fuzzy controller in our previous simulation study [1]. However, sharp decrease of muscle force was not tested. It is necessary to study a method of dealing with such sudden change.

In this paper, the first experimental test was performed to study the effectiveness of the fuzzy controller. Methods of detecting the condition to start control and the maximum knee extension angle in each cycle were not considered strictly. The way of detecting maximum knee extension angle used in this paper was to compare each sampled data of joint angle after the end of electrical stimulation to the Vastus. There is a possibility that the detection method doesn't perform well because of disturbance, and so on. Development of practical methods will be necessary for clinical use.

Reference

[1] A.Arifin, T.Watanabe, N.Hoshimiya: Design of Fuzzy Controller of the Cycle-to-Cycle Control for Swing Phase of Hemiplegic Gait Induced by FES. IEICE Trans. Inf. and Syst, vol.E89-D, no.4: 1525-1533, 2006

Acknowledgements

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