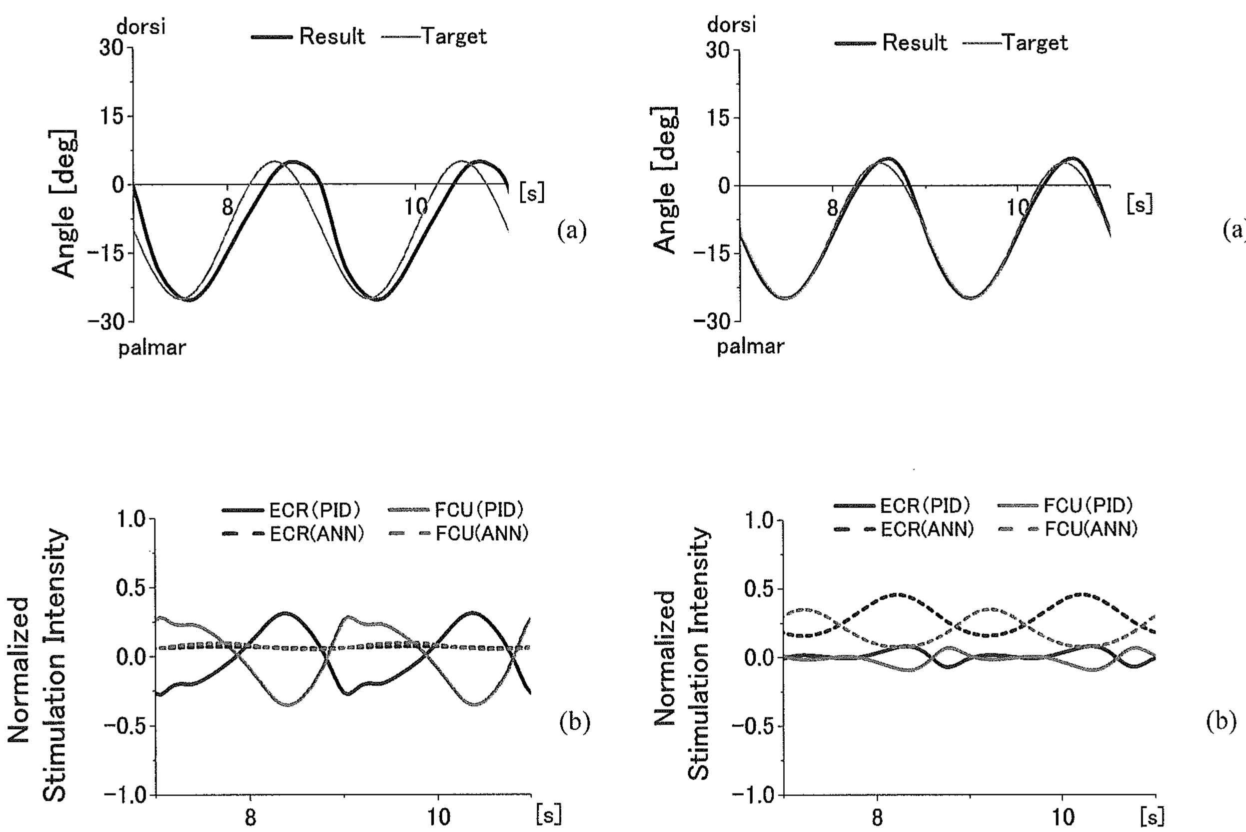
Similar tendency was also observed through the same simulations conducted changing the ANN's initial value of weights. These results suggest that the computer simulation by the musculoskeletal model is useful.

The forward dynamics model used in the previous study had difficulties in modification to express different muscle properties and in expansion to multi-degrees of freedom of movement control. Since the musculoskeletal model used in this study can be modified easily, it is considered to be suitable for this research work compared to the forward dynamics model. It is expected that control capabilities of the FEL controller will be revealed in more detail by performing computer simulations under various conditions modifying properties of this musculoskeletal model.

The ANN trained with fast movements could control the slow movements with good performance. Therefore, training with fast movement may only be required for restoring movements at the clinical site. This will lead to reduce the burden of patients. However, the ANN sometimes failed to learn the inverse dynamics of controlled object. When there was large difference in the output level between the PID controllers for the ECR and the FCU, only one output of the ANN became larger and the training didn't succeed. Modification of the PID controller or the structure of the ANN will be required to solve this problem.

Conclusions

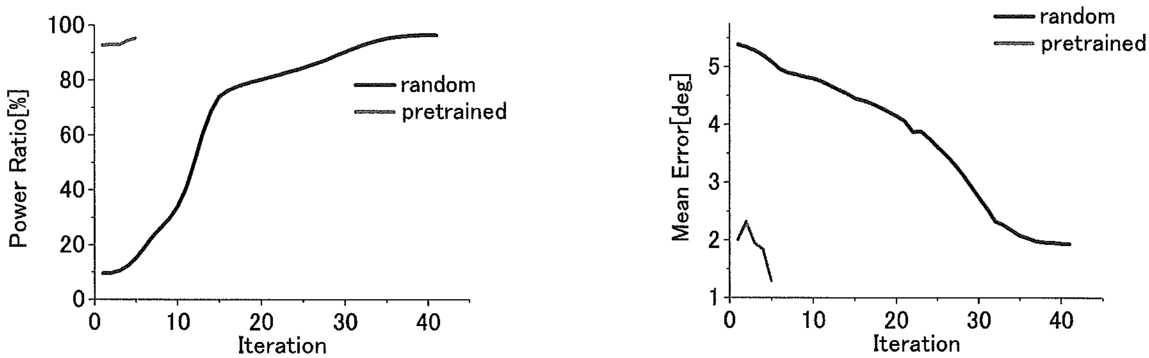
The FES controller using the FEL was found to be a promising controller from the results of computer simulations. However, the PID controller or the structure of the ANN has to be discussed in more detail for further improvement, because the ANN wasn't trained appropriately in some cases. Then it is expected to extend the FEL controller to controlling multiple joint movements. Since the musculoskeletal model was found out to be a useful tool for these studies, further study of the FEL controller can be performed by computer simulation.



(A) iteration 1

(B) iteration 24

Figure 4. An example of simulation results before and after the FEL
(a) angle trajectory (b) outputs of each controller



(a) mean error

(b) power ratio

Figure 5. An example of changes in mean error and power ratio (PR)

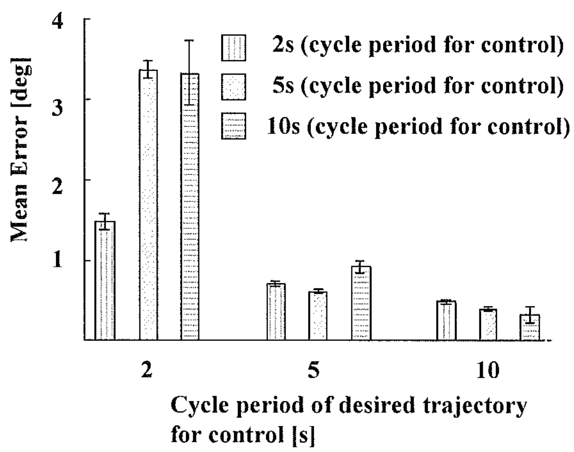


Figure 6. Mean error when the cycle period of the desired trajectory during control is different from it during training

Acknowledgment

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A Test of Stimulation Schedules for the Cycle-to-Cycle Control of Three-joint Movements of Swing Phase of FES-induced Hemiplegic Gait

A. Arifin*, T. Watanabe**, M. Yoshizawa**, and N. Hoshimiya***

* Graduate School of Engineering, Tohoku University, Sendai, Japan
** Information Synergy Center, Tohoku University, Sendai, Japan
*** Tohoku Gakuin University, Sendai, Japan

arifin@yoshizawa.ecei.tohoku.ac.jp

Abstract: Our previous computer simulation study showed effectiveness of the cycle-to-cycle control implemented in fuzzy controllers for controlling swing phase of FES-induced gait. There was difference in the knee joint angle trajectory between the FES-induced gait and the normal gait. In this study, we performed computer simulation to test six different stimulation schedules for the cycle-to-cycle control. The purpose of this test was to attempt to generate the joint angle trajectories that were similar to the normal gait trajectories. Computer simulation result showed that the stimulation schedule with co-activation of the hamstrings and the vastus muscles at the beginning of swing phase was effective in improving the knee joint angle trajectory with the hip and knee joint angle trajectories that were not far from the normal gait trajectories. The stimulation schedule designed based on EMG data could not realize target joint angle of swing phase control. Utilizing of the understanding of the joint movement and the muscle function was found to be necessary for design of the stimulation schedule for cycle-to-cycle control.

Introduction

The cycle-to-cycle control implemented in a set of fuzzy controllers was found to be effective in FES control of swing phase movements through computer simulation [1]. The trajectories of the hip and ankle joint angles were similar to the trajectories of the normal gait. However, the knee joint flexed faster than that of the human normal gait in joint angle trajectory.

In the cycle-to-cycle control, sequence of the muscle stimulation is arranged in a stimulation schedule. Basically, the stimulation schedule was designed based on joint movement and muscle function to generate relevant joint movement during a certain gait phase in the previous study. Various stimulation schedules can be designed to achieve the purpose of the cycle-to-cycle control.

In this paper, six different stimulation schedules were tested to attempt to generate the joint angle trajectories that were similar to the normal gait trajectories through computer simulation of FES-induced hemiplegic gait. Trajectories of the controlled joint angles obtained by each stimulation schedule were evaluated by comparing to the normal gait trajectories.

Materials and Methods

Stimulation Schedule for the Cycle-to-Cycle-Control of Three-joint Movements: The cycle-to-cycle control regulates the stimulation burst duration, while amplitude, pulse width and frequency of stimulation pulse are fixed. Six different stimulation schedules tested in the cycle-to-cycle control are shown in Figure 1. Stimulation schedule A was used in the previous study [1] and other five stimulation schedules (stimulation schedules B-F) were designed in this study. In all stimulation schedules, each muscle or muscle group is stimulated to induce the related joint movement to reach the control objective. Beginnings of the muscle stimulation were at the maximum hip extension, maximum knee extension or maximum ankle dorsiflexion angles at the end of stance phase. In normal gait, those maximum joint angles usually occur at different time in a cycle of gait. In order to facilitate the computer simulation, we assumed these maximum joint angles occurred simultaneously.

In the stimulation schedule A, the iliopsoas, the hasmtrings (biceps femoris short head and bicep femoris long head), the quadriceps (vastus muscles and rectus femoris), the gastrocnemius medialis and the tibialis anterior muscles were stimulated 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 reach the target of hip joint angle at initial contact. The tibialis anterior and the soleus were also stimulated simultaneously to prepare a good initial contact.

Stimulation schedules B-E were variants of the stimulation schedule A. In order to reduce excessive knee flexion caused by simultaneous stimulation of the hamstrings and the gastrocnemius medialis at the beginning of swing phase control, the hamstrings were stimulated after the ankle joint angle reached the target

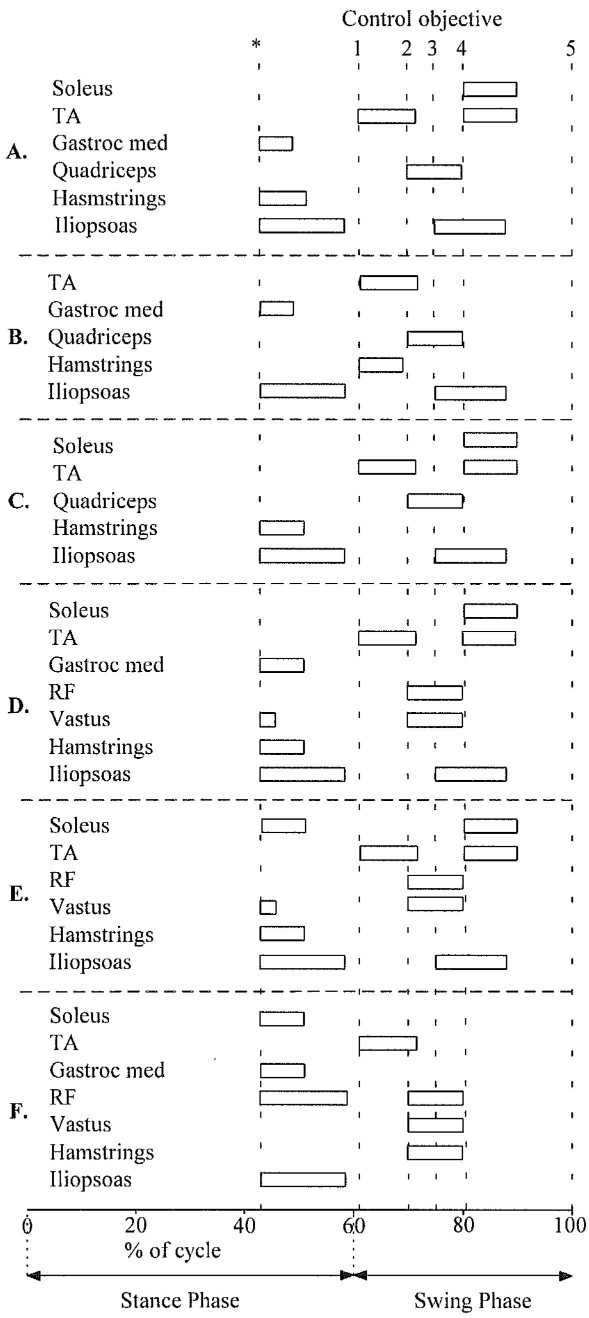


Figure 1: Six different stimulation schedules for three-joint control. *: beginning of stimulation (maximum hip extension angle, maximum knee extension angle, and maximum ankle dorsiflexion angle at the end of stance phase). Control objective: 1: maximum ankle plantar flexion angle, 2: maximum knee flexion angle, 3: maximum hip flexion angle, 4: maximum ankle dorsiflexion angle, and 5: maximum knee extension angle and hip and ankle angles at initial contact.

joint angle of maximum ankle plantar flexion in the stimulation schedule B. Additionally, co-activation of the tibialis anterior and the soleus was omitted in order to test the significance of the simultaneous stimulation of these muscles. Stimulation C was aimed to test effect of omitting the stimulation of the gastrocnemius medialis at the beginning of control on knee flexion. Purpose of the stimulation schedule D was to test effect of co-activation of the vastus muscles with the hamstrings in flexion of knee joint at the beginning of swing phase. The stimulation schedule E was to test effect of stimulation schedule D if the ankle plantar flexion activated by the soleus and gastrocnemius muscle was not stimulated. Stimulation schedule F was aimed to test possibility of using stimulation schedule based on the EMG data of the lower limb muscles of the normal gait. This stimulation schedule was qualitatively designed based on the EMG data [5].

Musculo-skeletal Model and Computer Simulation: In the present study, we designed musculo-skeletal model for FES-inuduced hemiplegic gait based on literature [2]. The model consisted of a paralyzed leg and a normal leg. The paralyzed leg model was activated by the electrically stimulated muscle model. While the normal leg was activated by joint angle trajectories measured from a normal subject. Parameters values of musculo-skeletal model were obtained from literature [3].

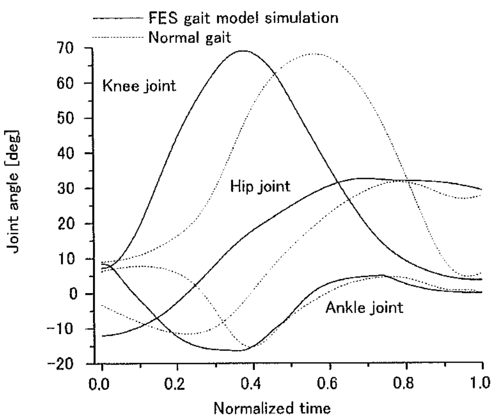
Computer simulation of the cycle-to-cycle control was performed using the fuzzy controllers [1]. In the test of the designed stimulation schedules, the cycle-to-cycle control was initiated with zero burst durations of stimulation pulses for 200-cycles stimulation courses of swing gait.

The burst duration of the vastus muscles at the beginning of control of stimulation (co-activation with the hamstings) of stimulation schedule D was determined by a ratio of it to the stimulation burst duration of the hamstrings. We tested the stimulation schedule D using five different burst durations of the stimulation of the vastus muscles: 0.1, 0.2, 0.3, 0.4 and 0.5 of the burst duration ratio. A well controlled gait was defined as a condition when all the controlled joint angles reached their targets satisfying error criterion [1]. The trajectories of the controlled joint angles of each stimulation schedule obtained under the condition of the well controlled gait were used to evaluate the designed stimulation schedule.

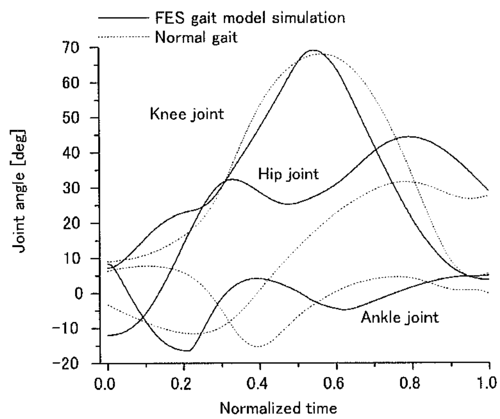
Results

Figure 2 shows the trajectories of the controlled joint angles compared to the trajectories of the normal gait. Stick figure in Figure 3 represents the movements of the FES-induced gait (black line) and normal gait (grey line). Criterion of evaluation of the stimulation schedules were RMS error, stride length and minimum foot clearance. The RMS error was the RMS value of difference between the trajectory of the controlled joint angle and that of the normal gait. The stride length was defined as distance of horizontal displacement of the heel from the beginning of the swing gait to the end of that. The minimum foot clearance was defined as the minimum height of the toe during the swing phase. Result of performance evaluation of the stimulation schedules is summarized in Table 1.

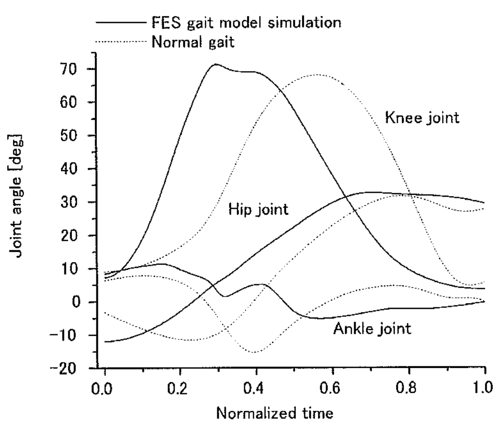
The stimulation schedule A resulted in fast knee flexion at the beginning of swing phase. However, the stick figure in the left top of Figure 3 shows the pattern of the swing gait that was not entirely different from the normal gait pattern. Stimulation of the hamstrings of the stimulation schedule B generated gait pattern that was obviously different from the normal gait (right top of Figure 3). In case of the stimulation schedule C that eliminated stimulation of the gastrocnemius medialis, the hamstrings was stimulated with longer burst duration resulted in fast knee flexion. Additionally, the target joint angle of the maximum ankle plantar flexion was not realized (Figure 2(c)) and minimum foot clearance was very small (Table 1). Longer stimulation burst duration of the vastus muscles in stimulation schedule D (co-activation with the hamstrings) increased the number of cycle required to reach the well controlled gait. In case of the burst duration ratio was 0.3 (Figure 2(d)), the error of the knee joint angle was smaller than that of the stimulation schedule A and the



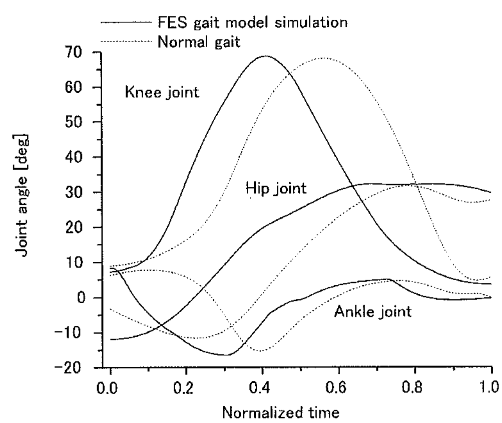
(a) Stimulation schedule A



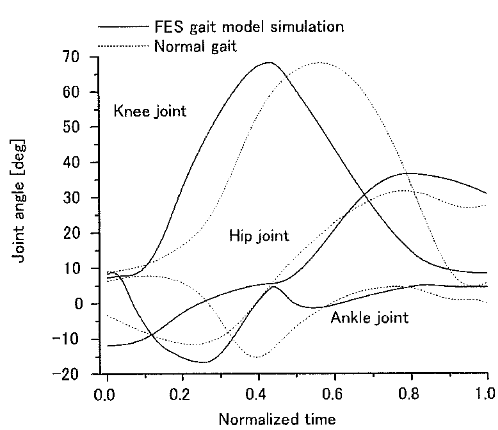
(b) Stimulation schedule B



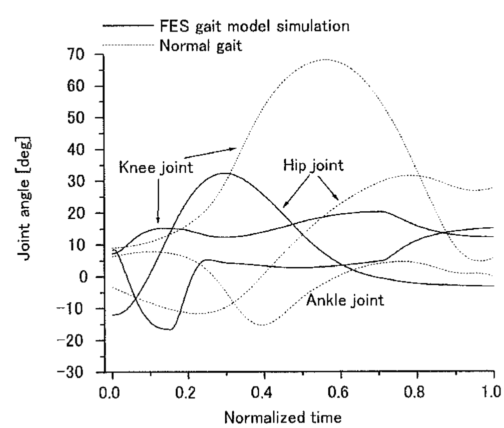
(c) Stimulation schedule C



(d) Stimulation schedule D



(e) Stimulation schedule E



(f) Stimulation schedule F

Figure 2: Trajectories of the controlled joint angles of each stimulation schedule. The trajectories of the joint angles obtained form normal gait are also shown in each figure.

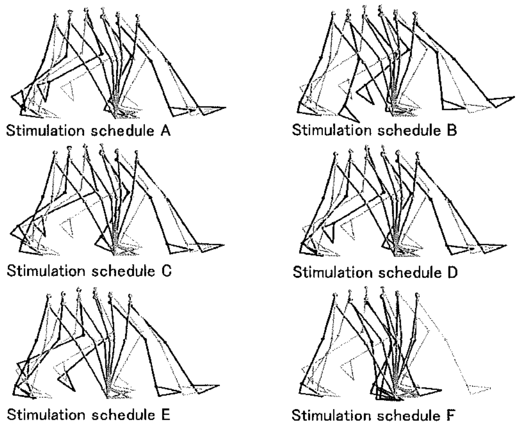


Figure 3: Stick figure of gait pattern generated by stimulation schedule (black) and normal gait (grey).

Table 1: Performance evaluation of stimulation schedule

Stim Schd	RMS error [deg]			Stride length [cm]	Min foot clearance [cm]
	Hip angle	Knee angle	Ankle angle		
A	9.7	22.9	7.4	120.3	2.0
B	19.3	6.8	11.1	120.3	3.5
C	8.1	24.1	8.0	120.3	0.1
D	10.2	19.4	8.6	120.7	2.2
E	6.2	16.7	10.3	119.3	0.3
F	27.4	28.5	12.2	67.3	-2.6

well controlled gait was reached in a few cycles. Stimulation of the soleus to induce ankle plantar flexion at the beginning of swing phase of stimulation schedule E caused unstable movements of the hip and ankle (Figure 2(e)). Stimulation schedule F generated unstable hip joint movement and insufficient knee flexion angle as shown in Figure 2(f). The foot impacted to the ground due to inappropriate hip and knee joint angles.

Discussion

Stimulation of the hamstrings after the ankle joint reached the maximum ankle plantar flexion angle in the stimulation schedule B resulted in a slow flexion of knee joint. Although this method was effective in improving the knee joint angle trajectory, the unstable movement of the hip joint was caused. On the other hand, co-activation of the vastus muscles and the hamstrings in the stimulation schedule D could improve knee joint angle trajectory without deterioration in controlling the hip and ankle joints. Considering values of the criterions of evaluation of the stimulation schedules and capability to reach the well controlled gait, the stimulation schedule D is preferable to the other stimulation schedules. The minimum foot clearance was also close in value to the average value of the normal gait (2.19 cm) in [4]. In this study, the test of the stimulation schedules was focused on improvement of the knee joint angle trajectory. Modulation of stimulation intensity is considered to be an alternative to generate all the controlled joint angle trajectories highly similar to the trajectory of the normal gait.

The results showed that EMG-based stimulation schedule generated unsuccessful gait pattern. On the other hand, the effectiveness of the stimulation schedule based on understanding of the joint movements and function of the muscles was shown. In order to generate the appropriate gait pattern, it is necessary to combine the EMG pattern with understanding of the joint movements and function of the muscles in design of stimulation pattern for FES-induced gait.

Conclusions

Six different stimulation schedules for the cycle-to-cycle control of swing phase of FES-induced hemiplegic gait were tested to attempt to generate the joint angle trajectories that were similar to those of the normal gait through computer simulation. Co-activation of the vastus muscles and the hamstrings at the beginning of swing phase was effective to improve the knee joint angle trajectory. Utilizing of the understanding of the joint movement and the muscle function was necessary for design of the stimulation schedule for cycle-to-cycle control.

Acknowledgement

This study was partly supported by the Ministry of Education, Culture, Sports, Science and Technology of Japan under a Grant-in-Aid for Scientific Research, and the Ministry of Health, Labour and Welfare under the Health and Labour Sciences Research Grants (Comprehensive Research on Disability Health and Welfare).

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第5回計測自動制御学会制御部門大会
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機能的電気刺激（FES）制御におけるフィードバック誤差学習の適用方法の検討

○帖佐征一*，　渡邉高志**，　吉澤誠**，　星宮望***
*東北大学大学院工学研究科　**東北大学情報シナジーセンター　東北学院大学***

A Study of application method of Feedback Error Learning on Functional Electrical Stimulation (FES) Control

S. Chosa*, T. Watanabe**, M.Yoshizawa** and N.Hoshimiya***
* Graduate School of Engineering, Tohoku University
** Information Synergy Center, Tohoku University, *** Tohoku Gakuin University

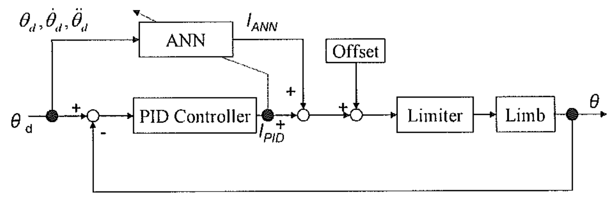
Abstract— In this study, we proposed a new method of adopting Feedback Error Learning (FEL) for FES controller. Although the feasibility of the FEL FES controller was shown in the previous study, there was the problem that the training of Artificial Neural Network (ANN) failed in some cases. In this study, a new FEL control system was proposed. Computer simulation results showed that the new method improved the training process of ANN.

1　はじめに

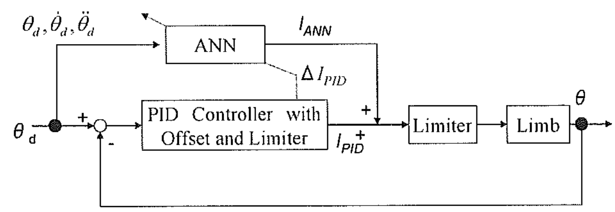
機能的電気刺激（FES）は、脊髄損傷や脳梗塞などによる麻痺患者の運動機能を再建する手段として期待される技術である。FES の実用化における課題の 1 つは、正確で安定な制御器を開発することである。現在、臨床で多く用いられている開ループ制御器は刺激データの初期設定や微調整が必要であり、患者や医療スタッフにとって負担が大きいという問題点がある。一方、我々の研究グループでは、多チャネル PID 制御器を設計し、4 筋を電気刺激することによる手関節の 2 自由度運動について、良好な追従制御が行えることを示してきた[1]。しかしながら、制御する動作の速度が大きい場合には制御誤差が大きいという問題点も指摘されていた。

フィードバック誤差学習（FEL）[2]を用いた制御器は、開ループ制御器と閉ループ制御器を組み合わせたハイブリッド型の制御器であり、遅れなく制御できるという開ループ制御器の長所と、外乱を補償するという閉ループ制御器の長所を併せ持つと考えられる。そこで我々の研究グループでは、FEL を FES 制御器に適用することを検討してきた。閉ループ制御器として PID 制御器を、開ループ制御器としてニューラルネットワーク（ANN）を用いた FEL 制御器の制御能力を確認し、PID 制御器と比較して遅れや誤差の小さな制御が可能であることを示してきた[3]。しかし、PID 制御器の出力レベルが 2 つの筋で大きく異なる場合、学習が適切に行われないという問題点も報告された[4]。

本報告では、筋骨格モデルシミュレーション[5]を用いて、FEL を用いた FES 制御器において ANN の学習に用いる誤差信号と PID 制御器を変更することを検討した。その結果、手関節 1 自由度運動制御において、これまで学習が適切に行えなかった条件においても適切な学習を行えるという結果が得られたので報告する。



(a) Previous method (FEL1)



(b) Proposed method (FEL2)

Fig1. The block diagrams of the FEL control system for FES

2　方法

2.1 フィードバック誤差学習を用いた FES 制御器

本報告では、今回新たに提案する FEL の FES 制御への適用方法について、筋骨格モデルによるシミュレーション実験により比較検討を行った。Fig.1(a)に従来の FEL 制御系（FEL1)のブロック図を示した。ANN および PID 制御器からの出力の和に、筋の応答の刺激閾値に相当するオフセットが加算され、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した。この方法では、PID 制御器の出力 I_{PID} は正負両方の値とする。Fig.1(b)に、本報告で提案する FEL 制御系（FEL2）のブロック図を示した。Fig.1(a) との違いは、オフセットの加算とリミッタによる刺激最小値と最大値の制限が PID 制御器内部で行われており、PID 制御器の出力 I_{PID} は

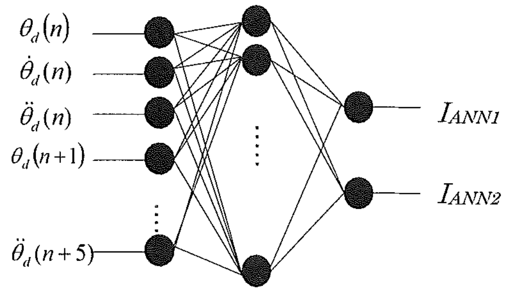


Fig2. Structure of 3 layers ANN used for the FES controller

常に刺激閾値以上、刺激最大値以下の値になることである。

ANNは、ニューロン数が入力層18個、中間層18個、出力層2個の3層パーセプトロンを用いた(Fig.2)。ANNへの入力は、現時刻から5時刻先までの目標角度、角速度、角加速度とした。本報告では、２つの筋を刺激して動作を制御するので、ANNの出力は各筋への刺激強度 I_{ANN1}、I_{ANN2} である。ANNの中間層と出力層の出力関数は、０から１までを出力するシグモイド関数を用いた。また、PID制御器のアルゴリズムは式(1)、(2)で表される。

$$\Delta I_{PID,n} = K_p(e_n - e_{n-1}) + K_I e_n + K_D(e_n - 2e_{n-1} + e_{n-2}) \qquad (1)$$

$$I_{PID,n} = I_{PID,n-1} + \Delta I_{PID,n} = K_p e_n + K_I \sum_{i=0}^{n} e_i + K_D(e_n - e_{n-1}) \qquad (2)$$

ここで、$\Delta I_{PID,n}$ は時刻 n での時刻 n-1 からのPID制御器の出力変化分を、$I_{PID,n}$ は時刻 n でのPID制御器の出力を表している。また、e_n は目標関節角度と実現関節角度の誤差を、K_P、K_I、K_D はPIDパラメータを表している。ここでFEL2では、PID制御器の中にリミッタが含まれているため、$I_{PID,n}$ が刺激最大値より大きい、または刺激最小値より小さい場合、$I_{PID,n}$ がそれぞれ刺激最大値、刺激最小値に更新されて次の時刻で用いられる。一方で、FEL 1では、$I_{PID,n}$ は更新されずにそのまま次の時刻で用いられる。

2.2 ANNの学習

ANNの学習は、式(3)で表される学習則に従って行われた。

$$\Delta w_{ij} = \varepsilon\left(\partial I_{ANN,j} / \partial w_{ij}\right) \times ES \qquad (3)$$

ここで、Δw_{ij} はANNの結合係数の変化分、ε は学習係数、$I_{ANN,j}$ はANNの出力を、ES は誤差信号を表している。従来のFEL制御系（FEL1）の場合、誤差信号としてPID制御器の出力 I_{PID} を用いていた。しかしながら、今回提案するFEL制御系（FEL2）では、PID制御器の出力 I_{PID} が常に正であるため誤差信号としては使用できない。そこで、誤差信号としてPID制御器の出力変化分である ΔI_{PID} を用いた。よってANNは、式（1）で表される誤差信号 ΔI_{PID} を減少させる方向に学習が進むことになる。

2.3 シミュレーション方法

FEL1およびFEL2を用いて、同じ筋骨格系の応答特性を持つモデルを用いて手関節1自由度運動制御を行った。刺激した筋は橈側手根伸筋群（ECRL／ECRB、以下ECR）および尺側手根屈筋（FCU）である。目標軌道は、振幅30度（掌屈角25度、背屈角-5度）、周期2秒の正弦波軌道とし、6周期分を学習の1セットとして、学習は一括更新により行った。

3 結果

未学習のANNを用いた最初の試行を学習1回目と表現する。FEL 1の学習1回目および学習50回目の制御結果をFig.3に示した。角度軌跡のグラフより、学習1回目、学習50回目共に制御の遅れや誤差が大きくなっていることが分かる。また、各制御器の出力のグラフより、学習50回目のPID制御器の出力が小さくなっていない。これらのことから、FEL 1では学習が適切に行われなかったと考えられる。

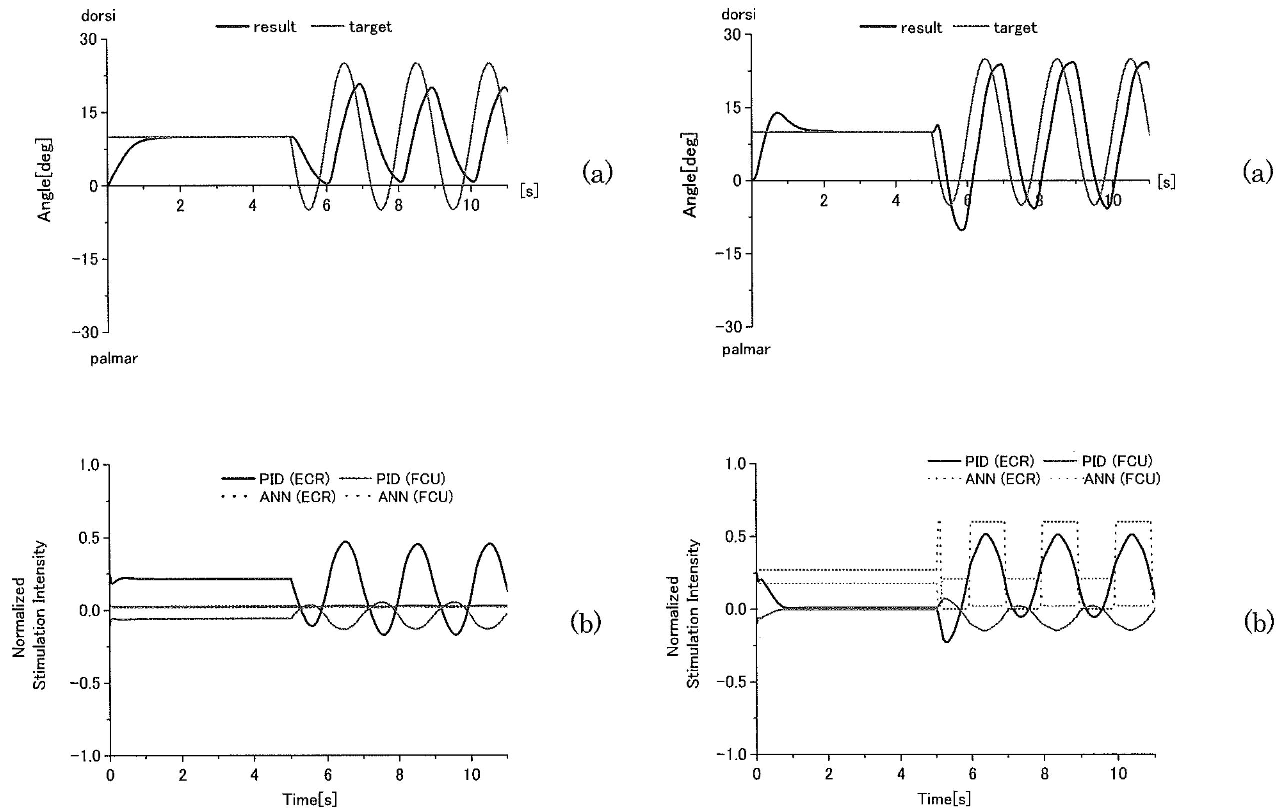
FEL2の学習1回目および学習40回目の制御結果をFig.4に示した。角度軌跡のグラフより、学習1回目に見られた制御の誤差や遅れが、学習40回目には改善していることが分かる。各制御器の出力のグラフより、学習1回目にはPID制御器中心の制御が行われていたが、学習40回目にはANN中心の制御に推移していることが確認できる。これより、FEL 2では学習が適切に行われたといえる。

FEL 1および、FEL 2の制御中の平均誤差の学習回数に対する推移をFig.5に示した。FEL 1については学習初期段階での誤差が7deg.と大きく、学習を重ねても誤差が減少していない。一方で、FEL 2については学習初期段階での誤差が3.4deg.と比較的小さく、学習によってさらに1.5deg.程度まで誤差が減少していることが分かる。

FEL 1、FEL 2の制御器全体の出力に対するANNの出力のパワーの比（Power Ratio、以下PR）の学習回数に対する推移をFig.6に示した。PRの計算方法を式(4)に示した。

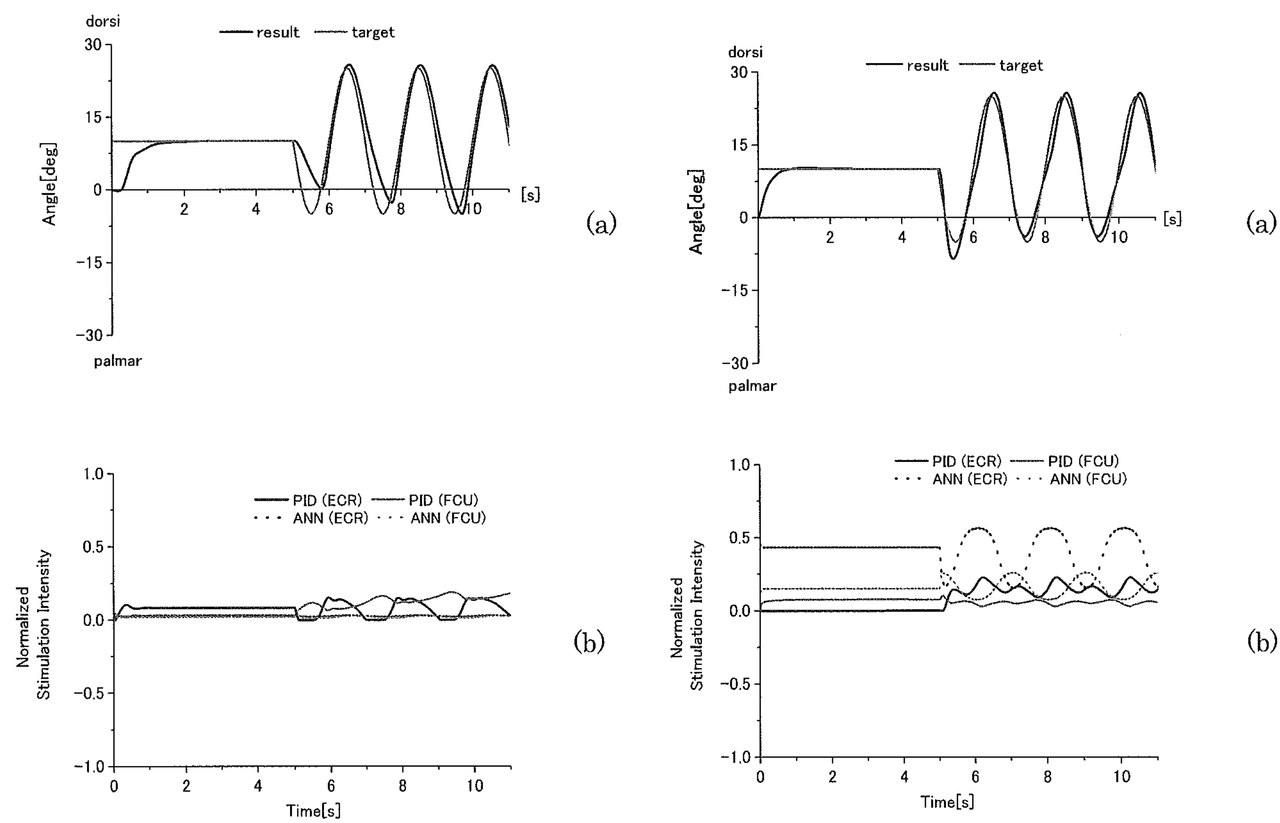
$$PowerRatio(PR) = \frac{\sum P_{ANN}(t)}{\sum P_{PID}(t) + \sum P_{ANN}(t)} \times 100 \qquad (4)$$

FEL1ではPRが70%強で飽和しているのに対して、FEL 2では学習回数20回程で90%近くまで上がり、ほぼANN中心の制御が行われていることが分かる。



(A)iteration1 (B)iteration50

Fig.3 Control results before and after the FEL using the previous method (FEL1)
(a) Angle trajectory (b) Outputs of each controllers



(A)iteration1 (B) iteration40

Fig.4 Control results before and after the FEL using the proposed method (FEL2)
(a) Angle trajectory (b) Outputs of each controllers

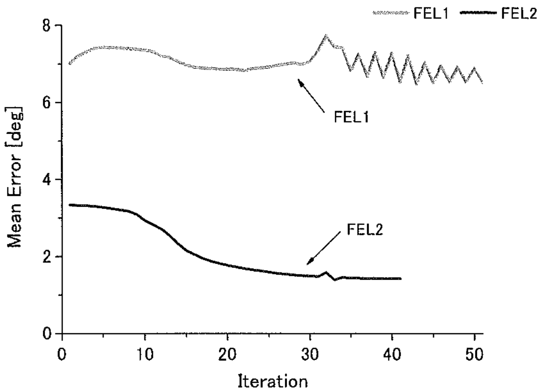


Fig.5 Change of the mean tracking error

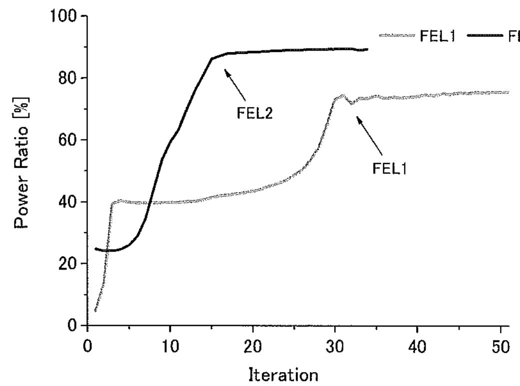


Fig.6 Change of the Power Ratio

4. 考察

FEL 1 、FEL 2 を用いてシミュレーション実験を行ったところ、FEL2 の学習 1 回目の制御結果が FEL1 の制御結果より誤差が小さく良好な結果であった。これは PID 制御器が、FEL1 では負の値を出力しているのに対して、FEL2 ではリミッタが刺激閾値で制限を加えているため、正の値を出力していることによるものと考えられる。例えば、時刻 n で PID 制御器の出力 $I_{PID,n}$ が刺激閾値を下回った場合、FEL1 では、時刻 n+1 で $I_{PID,n}$ に出力変化分 $\Delta I_{PID,n+1}$ が加算される。一方で、FEL2 では、$I_{PID,n}$ が刺激閾値に更新されてから、出力変化分 $\Delta I_{PID,n+1}$ が加算されるため、FEL2 では FEL 1 よりも大きな値が出力されることとなる。その結果、FEL1 では PID 制御器が負の値を出力する場合にも、FEL2 では刺激閾値以上の値を出力し、制御結果が良かったと考えられる。

また、FEL 1 では学習が適切に行われなかったが、FEL2 では適切な学習が行われるという結果が得られた。FEL 1 で学習が適切に行われなかった原因は、誤差信号である PID 制御器の出力にあると考えられる。Fig.3(A)(b)のグラフに見られるように、学習 1 回目の PID 制御器の ECR に対する出力が正に大きく、FCU に対する出力が負に大きかった。式(3)より、結合係数の変化分 Δw は誤差信号である PID 制御器の出力に比例するため、学習を繰り返すことで、ECR への出力に関係する ANN の結合荷重は大きくなり、FCU の出力に関係する結合荷重は小さくなる。したがって、ANN の刺激出力も ECR に対しては増加し、FCU に対しては減少することになり、FEL 1 では学習が適切に行われなかったと考えられる。一方、Fig.4(A)(b)のグラフより、FEL 2 では PID 制御器の出力が常に正となる。また、誤差信号として PID 制御器の出力 I_{PID} ではなく出力変化分 ΔI_{PID} を用いているため、ANN の出力が ECR、FCU のどちらかに偏ることは無く、FEL 2 では学習が適切に行われたと考えられる。

5 まとめ

フィードバック誤差学習（FEL）の FES 制御への新しい適用方法について検討し、計算機シミュレーションにより従来の方法と比較した。新方法では、ANN の学習に用いる誤差信号を、PID 制御器の出力変化分 ΔI_{PID} とし、PID 制御器の内部にリミッタを設置し、刺激閾値と刺激最大値の制限を加えた。今回提案した FEL の FES 制御器への適用方法により、学習が適切に行われないという問題点を解決する可能性が示された。今後は、モデルの筋特性を変えるなどして、異なる条件下でのさらなる検討が必要である。また、手関節多自由度運動制御への FEL 制御器の拡張が課題である。

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LETTER

Computer Simulation Test of Fuzzy Controller for the Cycle-to-Cycle Control of Knee Joint Movements of Swing Phase of FES Gait

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

SUMMARY We proposed a fuzzy control scheme to implement the cycle-to-cycle control for restoring swing phase of gait using functional electrical stimulation (FES). We designed two fuzzy controllers for the biceps femoris short head (BFS) and the vastus muscles to control flexion and extension of the knee joint during the swing phase. Control capabilities of the designed fuzzy controllers were tested and compared to proportional-integral-derivative (PID) and adaptive PID controllers in automatic generation of stimulation burst duration and compensation of muscle fatigue through computer simulations using a musculo-skeletal model. Parameter adaptations in the adaptive PID controllers did not significantly improve the control performance of the PID controllers. The fuzzy controllers were superior to the PID and adaptive PID controllers under several subject conditions and different fatigue levels. These results showed the fuzzy controller would be suitable to implement the cycle-to-cycle control of FES-induced gait.
key words: *cycle-to-cycle control, functional electrical stimulation (FES), fuzzy controller, PID controller, adaptive PID controller*

1. Introduction

The cycle-to-cycle control is a control method for restoring paralysed gait using functional electrical stimulation (FES) [1], [2]. The cycle-to-cycle control implemented in a proportional-integral-derivative (PID) controller was experimentally tested in controlling the knee extension angle [1] or the hip flexion angle range [2]. However, the PID controller showed deterioration in compensating fatigue of the hip flexors [2].

Controlling paralysed limb using FES is a difficult problem because of nonlienarity and time varying properties of the musculo-skeletal system. Fuzzy controller is nonlinear in nature. Main advantage of the fuzzy controller is the capability of utilizing the human expert knowledge to calculate control action. Additionally, simple design procedure of the fuzzy controller leads to the successful applications of a variety of engineering systems [3]. Therefore, the fuzzy controller has high potential to implement the control method for restoration of functional movement of paralysed limb using FES.

This letter presents computer simulation tests of the fuzzy controllers for the biceps femoris short (BFS) and the vastus muscles to control flexion and extension of the knee joint during the swing phase. The capabilities of the fuzzy controllers were tested and compared with the PID and the adaptive PID controllers in two crucial items of FES control: automatic generation of stimulation burst durations and compensation of muscle fatigue, under several different subject conditions and levels of fatigue.

2. Methods

2.1 Outline of the Cycle-to-Cycle Control

Gait is a cyclical movement. In a certain sub phase of gait, movements of the knee joint reach certain joint angles (i.e. a maximum knee flexion angle of swing phase, a maximum knee extension angle of swing phase, etc.). In the cycle-to-cycle control, each muscle is stimulated by single burst duration with constant intensity of stimulation pulses to induce the joint movement reaching the target joint angle (such as the maximum joint angle of normal gait). The controlled maximum joint angle of the previous cycle is delivered as feedback signal. Error is difference between the target and feedback signal. The burst duration of stimulation pulses of a current cycle is regulated based on the error of the previous cycle to ensure the joint angle reaching the target joint at every cycle.

2.2 Determination of Target Joint Angles and Stimulation Schedule

Sequence of the muscle stimulation for the cycle-to-cycle control is arranged in a stimulation schedule. We developed the stimulation schedule to generate those joint movements as shown in Fig. 1 based on the joint movements during swing phase and muscle functions to generate relevant joint movements. Beginning of the muscle stimulation is at the maximum knee extension of the end of stance phase. In this stimulation schedule, the stimulation of every cycle was started with the stimulation to the BFS. The stimulations of the BFS and the vastus muscles were controlled to induce the knee movements reaching the targets of the maximum

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[†]The author is with Graduate School of Engineering, Tohoku University, Sendai-shi, 980–8579 Japan.
[††]The author is with Information Synergy Center, Tohoku University, Sendai-shi, 980–8579 Japan.
[†††]The author is with Tohoku Gakuin University, Sendai-shi, 980–8511 Japan.
a) E-mail: arifin@yoshizawa.ecei.tohoku.ac.jp
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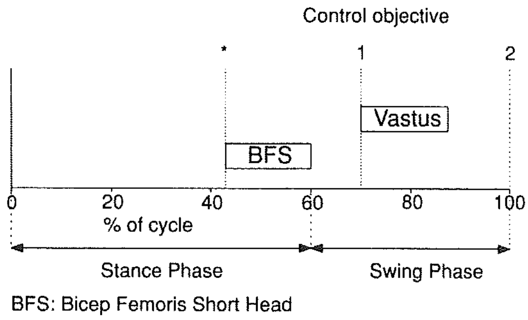

Fig. 1 The stimulation schedule. *: the beginning of the stimulation (the maximum knee extension angle of the end of the stance phase). The control objective: 1: the maximum knee flexion angle, 2: the maximum knee extension angle.

knee flexion angle and the maximum knee extension angle, respectively. In normalized scale, the stimulation intensities of the muscles were 1.0.

We performed an experiment to measure the knee joint angle of human subjects. Five neurologically intact subjects (males, average age: 24 ± 2.9 years, average body height: 170.2 ± 4.5 cm, average body weight: 60.6 ± 5.4 kg) participated in the experiment. Purpose of the experiment was explained to each subject, and consent was obtained. Gait analysis of the normal subject data from the experiment was performed to obtain the target joint angles. Parameter $\Delta\theta$ was introduced in order to evaluate whether the target joint angle was reached or not. We set the $\Delta\theta$ of each controlled joint angle as the average value of intra-subject standard deviation of each target joint angle resulted from the gait analysis. Values of the target joint angles with $\Delta\theta$ in parentheses are 69.0°(1.9°) and 3.6°(2.7°) for the maximum knee flexion angle and the maximum knee extension angle, respectively.

2.3 Control Scheme

The fuzzy controller for the BFS and the vastus muscles were designed based on the knowledge about function of the muscles of the knee joint and the knee joint movements during the swing phase. Structure, membership function and rule sets of the fuzzy controller were based on the Arifin et al. [4]. Control algorithm of regulation of the stimulation burst duration (TB) of current cycle is shown in Eq. (1),

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

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 structure of the PID controller for the cycle-to-cycle control was adopted from Franken et al. [2]. Method of setting of parameters of the PID controller was based on Arifin et al. [4]. The adaptive PID controller was developed from the PID controller by adding a parameter adaptation algorithm [5] (the adaptation constant: 0.001).

2.4 Computer Simulation Test

We designed a musculo-skeletal model for FES-inuduced gait based on Eom et al. [6]. Parameters values of the musculo-skeletal model were obtained from Ogihara et al. [7]. The motion equation was derived from the skeletal system model using the Lagrange function as shown in Eq. (2),

$$\mathbf{M}\ddot{\theta} + \mathbf{C}\dot{\theta} + \mathbf{G} = \tau \tag{2}$$

where θ, $\dot{\theta}$, $\ddot{\theta}$ 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 τ is vector of the joint torque. The joint movements activated by the electrical stimulation of the muscles were calculated by integrating the motion equation in Eq. (2) by the fourth order Runge-Kutta method with 10 μs integration time step.

Computer simulations to test capabilities of the designed controllers in controlling the knee joint movements were performed using a reference and twenty different subject models. The twenty different subject models were expressed by changing values of maximum muscle forces (50%-150%), mass of the shank and the foot (50%-150%), and/or length of the shank and the foot (75%-125%) of the reference subject model. The computer simulation test was divided into two parts: automatic generation of the stimulation burst durations and compensating muscle fatigue.

In test of automatic generation of the stimulation burst duration, the standard burst durations of the stimulation pulses were determined by the computer simulation initiated with zero burst durations. The standard burst duration of each muscle was obtained by averaging its burst durations of five cycles after all the controlled joint angles reached their targets with absolute errors were less than or equal to the $\Delta\theta$. In order to test possibility of using the standard burst duration of one subject as initial burst duration to other subjects, the standard burst durations of the reference (TBstd1), the strong3 (TBstd2), and the weak3 (TBstd3) subjects were applied to different subject models as initial burst durations in separate computer simulations. The computer simulation tests were performed in the stimulation course of 200 cycles of swing gait.

We modeled the muscle fatigue in the context of the cycle-to-cycle control as an exponential decreasing of the maximum muscle force, Fmax, to 50% of its original value as a function of the cycle number with a decay-constant [8]. We chose three values of the decay-constant, 5, 50, and 200 to represent a sudden, a moderate and a gradual fatigue, respectively. Each muscle was assumed to be fatigue after the 50th cycle. The test of the fatigue compensation was performed in an independent computer simulation of fatiguing of each muscle. In the muscle fatigue compensation test, the electrical stimulation of each subject was started with its own standard stimulation burst duration obtained in the automatic generation of the stimulation burst duration test.

3. Results

Figure 2 shows an example of the knee joint movements control result of the reference subject in automatic generation of the stimulation burst duration. The maximum knee flexion and extension angles in the swing phase were controlled by the cycle-to-cycle control. The settling index was defined as the number of cycles that were required to reach target joint angle with absolute error that was less than or equal to $\Delta\theta$. Average values of settling index of each controller with different initial burst durations are shown in Table 1. The settling indexes of the fuzzy controllers were significantly smaller than those of the PID and adaptive PID controllers. (t-test, significance level: 0.005). Adaptation of the controller parameters of the adaptive PID controllers reduced settling indexes about 8.3%-40% of the PID controllers. Using the standard burst durations of one subject as initial burst durations for different subjects could reduce settling index. However, this method resulted in occasionally inappropriate maximum knee angles at the beginning of stimulation. The fuzzy controllers could recover this condition faster than the PID and the adaptive PID controllers.

Because of the muscle fatigue, the controlled joint angles declined from their targets. The fuzzy controller could regulate stimulation burst duration fast, so the controlled joint angles reached to their targets again. The recovery index was defined as the number of cycles that were required to compensate muscle fatigue. The fatigue compensation was achieved when the absolute error decreased to be less than or equal to the $\Delta\theta$. The average values of recovery index and maximum error are shown in Table 2. The recovery index of the maximum knee flexion angle of the fuzzy controller was significantly smaller than those of PID controllers (t-test with significance level: 0.005). In the case of the maximum knee extension angle control, there was no significant difference among the recovery indexes of the three controllers. Although the maximum errors of the controlled joint angles obtained from the three controllers were not significantly different, the errors of the fuzzy controllers were smallest among the three controllers.

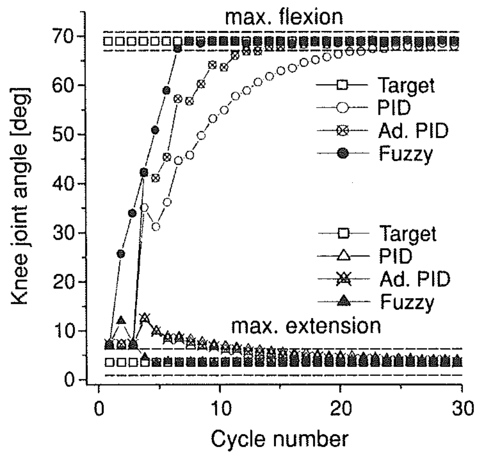

Fig. 2 A control result of the knee joint movements of the reference subject in the automatic generation of the stimulation burst duration. The maximum knee flexion and extension angles in the swing phase per cycle controlled by three controllers are posed together.

Table 1 Average settling index (cycles).

(a) Maximum knee flexion (BFS)

Initial TB	Controller Fuzzy	PID	Ad. PID
Zero	8 ± 3	25 ± 5	15 ± 2
TBstd1	3 ± 2	14 ± 5	8 ± 6
TBstd2	5 ± 2	19 ± 6	13 ± 4
TBstd3	8 ± 2	23 ± 5	16 ± 5

(b) Maximum knee extension (Vastus)

Initial TB	Controller Fuzzy	PID	Ad. PID
Zero	4 ± 1	12 ± 8	11 ± 4
TBstd1	2 ± 1	6 ± 6	5 ± 4
TBstd2	2 ± 1	7 ± 7	6 ± 6
TBstd3	4 ± 3	8 ± 6	7 ± 5

Table 2 Average recovery index and maximum error.

(a) Recovery index (cycles)

Knee joint angle	Controller Fuzzy	PID	Ad. PID
max. flexion (BFS)	6 ± 9	27 ± 40	18 ± 34
max. extension(Vastus)	0 ± 0	2 ± 6	2 ± 5

(b) Maximum error (deg)

Knee joint angle	Controller Fuzzy	PID	Ad. PID
max. flexion (BFS)	2.6 ± 2.5	4.2 ± 4.0	3.7 ± 3.7
max. extension(Vastus)	0.6 ± 0.5	1.1 ± 1.1	1.2 ± 1.2

4. Discussions

The computer simulation results showed that among three designed controllers the fuzzy controllers had best response in automatic generation of the stimulation burst durations and in compensating the muscle fatigue. The nonlinear mapping in the fuzzy controller accomplished by a number of fuzzy if-then rules could find the appropriate stimulation burst duration fast. The results of computer simulation tests suggest limitation of the PID controller that regulated control action in the way of a linear control algorithm. The adaptation of controller parameters in the adaptive PID controllers did not show a significant contribution to improve the control performances in all controlled joint angles. The adaptive PID controller showed the relatively improved control performances only in the maximum knee flexion angle control. Additionally, its control performance was inferior to the fuzzy controllers. Contribution of the controller parameter adaptation to the control performance depends on the adaptation constant. In the preliminary study, we found that a too small adaptation constant had no significant effect in improving the control performance. Otherwise, a too large adaptation constant resulted in the oscillating maximum joint angle.

We designed the fuzzy controller for the cycle-to-cycle control by a heuristic approach using the qualitative knowledge about the controlled system. The knowledge was incorporated in the controller structure, the fuzzy memberships, and the fuzzy rules. By using the knowledge, system identification could be eliminated and the fuzzy controller could be implemented with simple trial and error design procedure. The fuzzy Lyapunov-based approach was proposed as a systematic design of a fuzzy controller [9]. Using the systematic approach, the input variables and the fuzzy rule can be derived systematically. However, in our preliminary study, we found that the systematically designed fuzzy controller for the BSF muscle was inferior to the heuristically designed fuzzy controller in compensating the muscle fatigue.

Considering a clinical application of the cycle-to-cycle control, the standard burst duration obtained from a subject was applied to the other subject models as their initial burst durations. We observed the inappropriate joint angles in the strong subjects and thin subjects with the initial stimulation burst duration obtained from the reference subject and in all the subjects with the initial stimulation burst duration obtained from weak3 subject. This fact shows that an excessively long initial stimulation burst duration obtained from other subject causes inappropriate response at the beginning of stimulation. Therefore, using a standard stimulation burst duration of a subject as the initial stimulation burst duration stimulation to the other subjects will be effective if the initial stimulation burst duration is not excessively long.

In order to develop the cycle-to-cycle control as a practical method for clinical use, this control method has to be tested in controlling multi-joint movements, such as the hip, the knee and the ankle joint movements during the swing phase. The superiority of the fuzzy controller to the PID and the adaptive PID controllers showed in the present study and possibility to implement the fuzzy controller with a simple design procedure suggest that the fuzzy controller would be suitable to implement the cycle-to-cycle control for the multi-joint movements.

5. Conclusions

We tested the control capabilities of the fuzzy, the PID and the adaptive PID controllers for the cycle-to-cycle control in automatic generation of the stimulation burst duration and compensating the muscle fatigue through the computer simulations. In controlling the knee joint movements, the fuzzy controllers were superior to the PID and the adaptive PID controllers in all different subject models under several different fatigue levels. Considering the superiority of the fuzzy controller shown in this study and simplicity of the design procedure, the implementation of the cycle-to-cycle control for multi-joint control would be done in the fuzzy controller.

Acknowledgement

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A test of multichannel closed-loop FES control on the wrist joint of a hemiplegic patient

Watanabe T [1], Matsudaira T [2], Hoshimiya N [3], Handa Y [4]

[1] Information Synergy Center, Tohoku University, Sendai, Japan
[2] Graduate School of Engineering, Tohoku University, Sendai, Japan
[3] Tohoku Gakuin University, Sendai, Japan
[4] Tohoku University Graduate School of Medicine, Sendai, Japan
nabe@isc.tohoku.ac.jp

Abstract

The multichannel closed-loop FES control method using PID controllers developed in our research group was found to perform well with neurologically intact subjects in our previous studies. The PID controller could solve the ill-posed problem in the stimulation intensity determination for multichannel control of multi-degrees of freedom of movement. In this paper, the PID control method was examined with a hemiplegic patient. The dorsi/palmar and the radial/ulnar flexions were controlled by stimulating the ECR, the ECU, the FCR and the FCU using surface electrode stimulation. The tracking control of the wrist joints were achieved reasonably at vertical and horizontal positions of the upper limb. The multichannel PID control method would provide a basic technique for multichannel closed-loop FES control.

1. INTRODUCTION

A closed-loop control of paralyzed limbs using FES has been desired for clinical application. However, there are problems on a sensor to measure a feedback signal and a multichannel control algorithm to determine stimulation parameters to muscles. We focused on the algorithm for multichannel closed-loop FES control and have developed a method using the PID controller for the redundant musculoskeletal system that involved an ill-posed problem in stimulus intensity determination [1, 2].

The method was found to be effective on the tracking control of two degrees of freedom of movement of the wrist joint stimulating four muscles through experiments with neurologically intact subjects. The method could solve the ill-posed problem in calculation of stimulus intensities. However, good tracking control was not achieved with a hemiplegic patient at the first clinical test because of the small range of motion of the wrist radial flexion and increased reflex function [3].

In this paper, a clinical test of the closed-loop FES control was performed again with another hemiplegic patient who had experiences in FES and TES.

2. METHODS

2.1. Control Algorithm

The PID control algorithm was described by the following equation:

$$\boldsymbol{S}_n = \boldsymbol{S}_{th} + \boldsymbol{K}_{\mathrm{P}} \boldsymbol{e}_n + \boldsymbol{K}_{\mathrm{I}} \sum_{i=0}^{n} \boldsymbol{e}_i + \boldsymbol{K}_{\mathrm{D}} (\boldsymbol{e}_n - \boldsymbol{e}_{n-1})$$

where $\boldsymbol{S}_n$, $\boldsymbol{S}_{th}$ and $\boldsymbol{e}_n$ are stimulation intensity vector at present time n, the minimum stimulation intensity vector for FES control, and error vector at present time n that is the difference between targets and measured joint angles, respectively. The elements of PID parameter matrices $\boldsymbol{K}_{\mathrm{P}}$, $\boldsymbol{K}_{\mathrm{I}}$, $\boldsymbol{K}_{\mathrm{D}}$ were calculated by the followings [1]:

$$K_{Pij} = \frac{0.6T_i}{L_i} m_{ij}^-, \quad K_{Iij} = \frac{0.6\Delta t}{L_i} m_{ij}^-, \quad K_{Dij} = \frac{0.3T_i}{\Delta t} m_{ij}^-$$

L_i and T_i are delay time and time constant of step response, respectively, when the muscle i is stimulated separately. In case of a muscle has two or more degrees of freedom of movement, the delay time and the time constant obtained from every movement are averaged respectively. Δt is the sampling interval. m_{ij}^- is the element of the generalized inverse matrix $\boldsymbol{M}^-$ of the matrix $\boldsymbol{M}$. Elements of matrix $\boldsymbol{M}$ are slopes of approximated linear lines of the input-output (stimulus intensity-joint angle) characteristics calculated by the least squares method between the minimum and the maximum stimulation intensities. The inverse matrix of $\boldsymbol{M}$ does not exist in general because $\boldsymbol{M}$ is not the square matrix (i.e. usually, the number of muscle stimulated is larger than that of degree-of-freedom of movement controlled). Therefore, we introduced the generalized inverse matrix of $\boldsymbol{M}$ that was calculated uniquely under the limitation of the sign of the elements and the least square condition [4].

2.2. Experimental Method

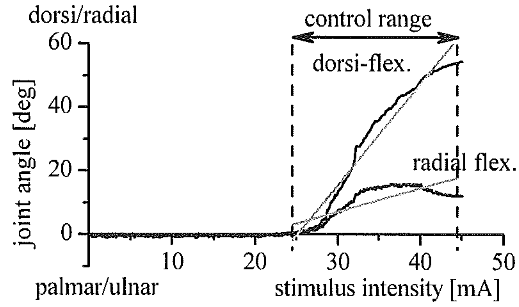
Wrist joint angles of the dorsi/palmar- and the radial/ulnar-flexions of the paretic side of a right hemiplegia (63 years old, male) were controlled by stimulating the flexor carpi radialis (FCR), the flexor carpi ulnaris (FCU), the extensor carpi radialis longus/brevis (ECR) and the extensor carpi ulnaris (ECU). Stimulus current pulse amplitudes (pulse frequency: 20Hz, pulse width: 0.2ms) were regulated by the controller and applied to the muscles through isolator (5384, NEC Medical Systems) and surface electrodes (F-150M, Nihon Koden). Maximum pulse amplitude was determined in order to get enough control range without pain. The wrist joint angles were measured with the magnetic 3-D position and orientation sensor (FASTRAK, Polhemus) with 20Hz of sampling frequency [2].

The subject sat on a chair and relaxed his right upper extremity during experiments. The closed-loop control was performed under two different upper limb positions: in the direction of the gravity with the neutral position of the forearm in the pronation/supination angle (vertical position) and in the horizontal plane with almost full extended elbow joint and 90deg pronation of the forearm (horizontal position). The horizontal position was maintained by supporting the forearm and the upper arm with wooden pedestal. Parameter values of the PID controller were determined at the vertical position, which were fixed for all control experiments. Target joint trajectory was circle on the joint angle plane with 10s or 5s of cycle period.

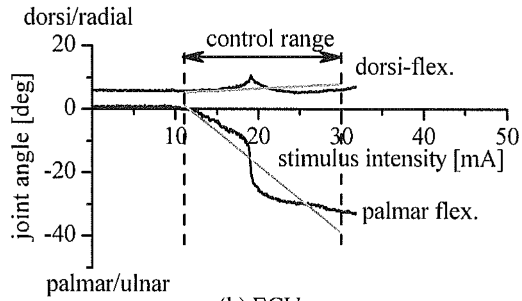
3. RESULTS

Examples of the input-output characteristics of stimulated muscles are shown in figure 1. The approximated linear lines between the minimum and maximum intensities for FES control are also shown in the figure.

Figure 2 shows examples of control results of tracking target joint angles. In the first 5s, joint positions were moved from the relaxed position to the starting position on the target trajectory. In the experiments, parameters of the PID controller were not optimized after the first determination from the input-output characteristics and the responses to step-shaped stimulations. In the case of controlling at the vertical position, tracking control was almost achieved although small amplitude of oscillation of joint angles was observed. Even in the case of controlling at the horizontal position, the tracking was not so severely deteriorated. Especially, little oscillation was caused in the first half of the control. These results were similar to those of the previous experiments with neurologically intact subjects. The PID controller was considered to perform well with the hemiplegic patient.



(a) ECR

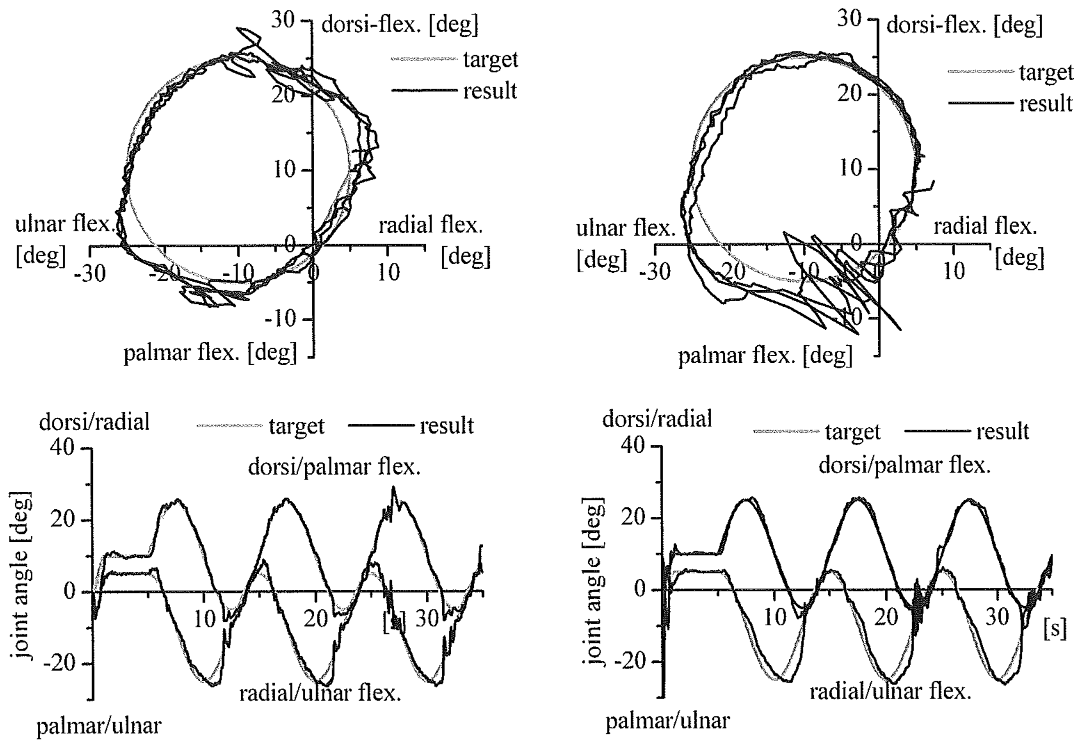


(b) ECU

figure 1 Examples of measured input-output characteristics of stimulated muscles (black line) and approximated linear lines for PID parameter determination (gray line). The minimum and the maximum stimulation intensities for FES control are shown by the control range.

4. DISCUSSION AND CONCLUSIONS

The PID controller could regulate stimulus currents properly after the PID parameter determination by simple measurements. In the experiment of this paper, fine tuning of the PID parameters was not performed considering clinical application since the trial and error tuning of the parameters causes burdens on patients. Simplifying parameter determination process is one of important factors for clinical



(a) vertical position (b) horizontal position

figure 2 Examples of tracking control results with a hemiplegic patient. Target and measured joint angle trajectories on the joint angle plane (upper figures) and time responses of them (bottom figures) are shown.

applications of FES controllers.

In the experimental results, the ECU had steep response in the input-output characteristic and little contribution to the dorsiflexion. These were considered to be a reason of the undesirable responses such as the oscillation of joint angles. As seen in our previous clinical test with a hemiplegic patient, the small range of motion and reflex like response caused oscillating responses [3]. It will be necessary to vary controller parameters considering stimulation intensity and/or joint angles and so on during control.

Results of this paper showed that the multichannel PID control method would be effective in FES control. Development of fine tuning method of the controller parameters and expansion of the PID controller to multijoint movement control will be necessary for practical clinical use.

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Application of Local EMG-Driven FES to Incompletely Paralyzed Lower Extremities

R.Futami [1], **K.Seki** [2], **T.Kawanishi** [3], **T.Sugiyama** [3], **I.Cikajlo** [3] **and Y.Handa** [2]

[1] Faculty of Symbiotic Systems Science, Fukushima University
[2] Graduate School of Engineering, Tohoku University
[3] Graduate School of Medicine, Tohoku University
email: futami@sss.fukushima-u.ac.jp

Abstract

For FES control of incompletely paralyzed lower extremities, various rule-based methods have been proposed. However, the necessity for adequate reliability of sensors and walking phase detection have made such systems difficult to be widely used clinically. In this report, a simple "local EMG-driven FES" system is proposed and shown to be effective in knee extension FES for incomplete paralyses. In this scheme, e.g., knee extensor is electrically stimulated according to the magnitude of measured voluntary EMG from the same muscle, which is partially paralysed. Apparently this scheme is applicable only to incomplete paralyses, however, the number of corresponding patients is not small and the method is applicable not only to walking, but also to other daily activities such as transfer from bed to wheelchair or standing-up, without any control-mode selection for each consecutive activity to aid, because those EMG reflect the neural commands from the brain.

1. INTRODUCTION

For the incomplete paralyses of lower extremities, rule-based FES control which is based on gait phase detection with foot switches and/or acceleration sensors have been proposed and examined[1-5]. In those method, however, there seems to be a problem that the resulted gait improvement is not so satisfactory for the patients when compared to the expense of time and difficulty in setting-up sensors and stimulator accurately. Additionally, those control algorithms should include bothering mode-switching. Therefore, simpler, but still effective FES control method is needed to be developed for some kinds of paralyzed patients.

In this report, a very simple "local EMG-driven FES" is proposed and shown to be effective in knee extension FES for incomplete hemiplegics by stroke. In this scheme, e.g., knee extensor is electrically stimulated according to the magnitude of measured voluntary EMG of the same muscle. Resulted voluntary muscle force is expected to be reinforced naturally and if it is applied to a muscle of which the maximum tension is not sufficient, this FES can be expected to aid the patient's daily activities such as crutch-walking, transfer from bed to wheelchair, or standing-up, without any bothering mode selection. This method is a kind of EMG-driven FES[6-8] and the idea itself is trivial, however, successful clinical application has not been reported, although the circuit design has been reported for the case in which the EMG detection and electrical stimulation share a same pair of electrodes[9]. In the method shown below, two independent pairs of electrodes were used for EMG detection and FES, to make it easier to suppress the artifact from surface stimulation to voluntary EMG detection.

2. METHODS

2.1. Basic design of the controller

To realize the surface FES and EMG detection to/from a same muscle, control system shown in Figure 1 was designed and examined. The system includes EMG electrodes, gating (protection) circuit, EMG amplifier, note PC with AD/DA card, surface stimulator and stimulating electrodes. The timing chart of the control is shown in Figure 2. Each amplitude of repetitive (20Hz) electrical stimulation of 500 micro seconds width was modulated by the power of voluntary EMG in each preceding period. Each stimulation pulse was paired by opposite polarity pulse so as to reduce the harmful transient effect onto EMG detection and also to reduce the long term electro-chemical change in electrodes. The EMG power of each segment was calculated from the samples for 20ms after subtracting the offset

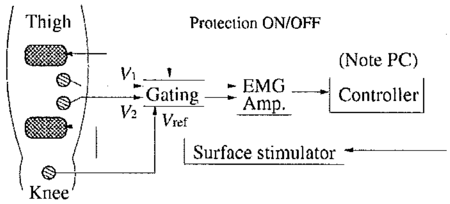

Figure 1. Local EMG-driven FES

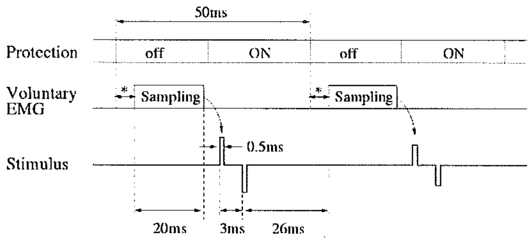

Figure 2. Timing chart of the control

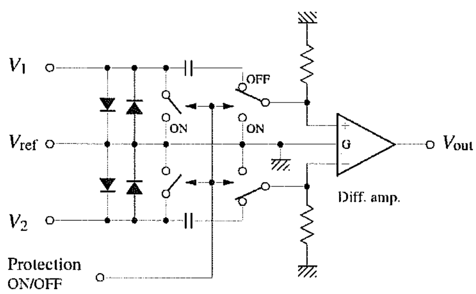

Figure 3. Gating (protection) circuit

level. Thus the resulted value corresponded to a "short-term standard deviation" of the EMG. A saturated threshold function was used for translating voluntary EMG power into FES amplitude. Sampling frequency of 1kHz was used for the RT-Linux PC as a controller.

2.2. Design to reduce artifacts

To protect the input stage of EMG amplifier from the damage by surface FES pulses, and to reduce the transient artifact, a gating circuit shown in Figure 3 was designed and examined. Due to the nonlinearity of silicon rectifiers, succeeding CMOS analog switches and amplifier can be protected from breakage. We also examined the effect of M- and H-waves. As a result of preliminary test for normal subject, it was confirmed that the time window design in Figure 2 was sufficient for use in lower extremity muscles (Significant H-wave could not be observed in knee extensor stimulation).

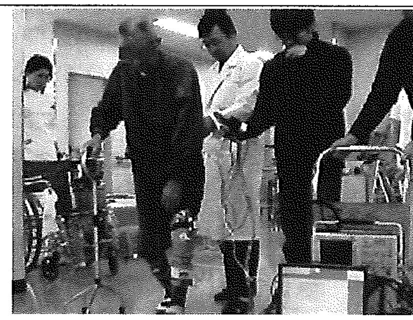

Figure 4. Clinical evaluation

2.3. Electrode arrangement

As the magnitude of artifacts from FES pulses to EMG measurement depends on the arrangement of each electrode pairs, we examined and compared several electrode arrangements experimentally. As as result, it was shown that in any arrangement including the case in which the EMG electrodes were sandwiched between FES electrodes, voluntary EMG could be clearly detected in the presence of adequate stimuli. Therefore, we adopted the arrangement to maximize the amplitude of voluntary EMG, by putting the electrodes in the direction of EMG propagation in muscle fibers.

2.4 Preliminary clinical tests

Three male subjects of 61 to 87 years old with hemiplegia by stroke participated in preliminary clinical evaluation of the system. Their voluntary knee extension forces were not sufficient to walk without crutch, and the speed of crutch-walking was very low. The subjects were applied the local EMG-driven FES described above to their quadriceps to improve the stability, speed and stride of their crutch-walking. The size of the surface FES electrode was 7cm x 9cm. EMG signal was measured at vastus medialis or rectus femoris, depending on the amplitude of voluntary EMG. Ankle foot orthoses were used for the paralyzed side.

3. RESULTS

For the two of three subjects, we could confirm that the walking stability was improved by local EMG-driven FES, and the subjective evaluation by patients were something like "this is helpful". For one other subject, noticeable improvement was not observed, and later it was confirmed that the noise from AC power line heavily affected EMG measurement because

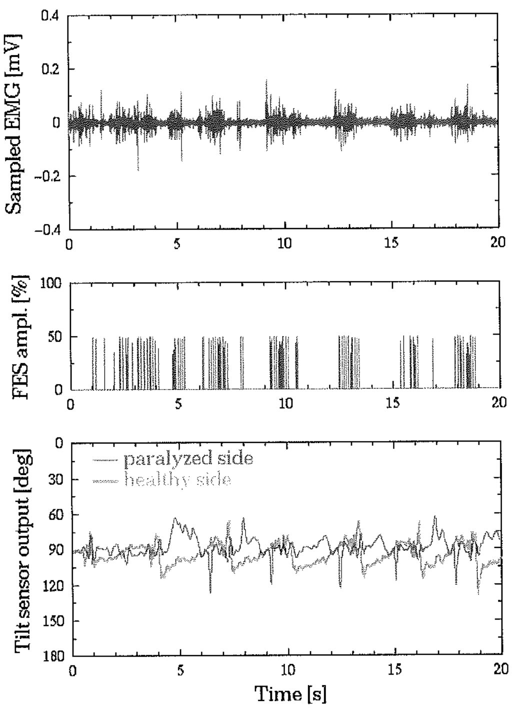

Figure 5. Example of walking with local EMG-driven FES (From the top, EMG, stimulation, and the outputs of tilt sensors at paralyzed and healthy side of shanks are shown.)

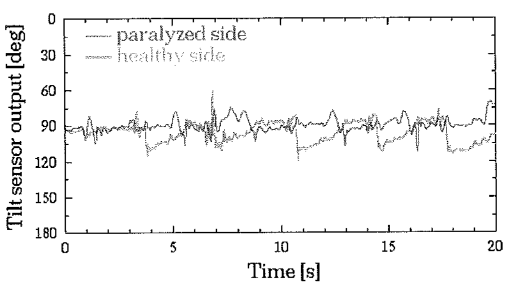

Figure 6. Example of walking without FES (Subject is the same as for Figure 5)

there happened attachment failure between EMG electrode and skin.

One of the results is shown Figures 5 and 6, from which we can confirm that the electrical stimulation was appropriately controlled by the sampled voluntary EMG signal. From the comparison of Figures 5 and 6, it can be also confirmed that the walking symmetry of the paralyzed and healthy side lower extremities was improved to some extent.

4. DISCUSSION AND CONCLUSIONS

A simple and easy-to-use local EMG-driven FES system was developed and examined. Although the clinical tests were subjective and limited for a small number of patients, the basic effectiveness of the idea was confirmed for the hemiplegic patients' crutch-walking.

With further development, we believe that a method to treat the EMG electrode failure could be found and controller could be implemented as battery-operated portable box. Using those controllers, quantitative evaluation of long-term effect of this control method should be carried out for various types of paralyses and objective motions in the near future. Application for completely implanted FES device should also be done.

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