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Development of a Novel Gait Rehabilitation System Based on FES and Treadmill-Walk for Convalescent Hemiplegic Stroke Survivors

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Abstract— **Recently, a large amount of stroke survivors are suffering from motor impairment. However, existed therapy interventions have limited effects to restore normal motor function. Thus, we proposed a novel control strategy for gait rehabilitation of hemiplegic patients. The whole system consists of a Functional Electrical Stimulation (FES) device and Treadmill-Walk system. FES contributes to improve the quality of the gait based on real-time adjustment of gait pattern. During gait, the electrical stimuli from separate output channels of an FES device are launched to stimulate two lower extremity muscles (Tibialis Anterior (TA) and Hamstrings). Stimulus launching procedure is based on identifying subject`s gait state (stance and swing phases). According to the current variation of treadmill motor, gait phase and muscle activation of lower limbs can be determined during walking on Treadmill-Walk. Three able-bodied subjects simulated hemiplegic patients in the experiment. The results indicated that the proposed method is a safe, feasible and promising intervention.**

I. Introduction

Currently, more and more people with mild or severe stroke are suffering from motor impairment. The recovery of lost motor function is hardly conducted by the biomedical treatments [1]. Physiotherapy only has typically limited effects to motor function restoration [2]-[4]. A large proportion of stroke survivors have persistent deficits in their physical mobility, but conventional interventions don't make a big effort toward the motor function recovery. Novel therapies are therefore in necessity.

Biomechanical solution has been considered a considerably effective method, such as FES device. It can realize the restoration of the motor function by stimulating the paralyzed muscles. While FES is applied to the dorsiflexor muscles of stroke survivors, it is commonly used to address foot drop [5]. It is also crucial to their walking ability recovery while it is applied to the quadriceps muscles as they swing the leg forward into the ensuing step [6]. Because high neural plasticity and repair mechanisms for restoring motor functions could be obtained by FES, it has been proved that the effectiveness of FES can be maintained for at least 24 months [7]. The purpose of utilizing FES in this research was to improve gait pattern and quality.

Treadmill training also has become an established rehabilitation method for hemiplegic patients after stroke. The positive effects of this task-specific therapy have been shown in various studies [8] [9]. The aim for hemiplegic patient to use the treadmill is to correct the asymmetric physical ability, because physical workload can be modified with independent operation of the left and right treadmill [10]. As for measuring the gait phase of hemiplegic patients, conventional methods, such as force plate, require a large amount of preparation. Also they are cumbersome to apply because of imposing a burden on both patient and therapist. We have proposed a novel method in previous research for measuring the walk phase via the only two split belt treadmill called Treadmill-Walk (shown in Fig. 1). It developed an original algorithm capable of properly estimating walking phase of healthy subjects by observing treadmill motor current [11].

With Treadmill-walk and FES method, the research aimed to contribute to the lower limb motor function recovery of stroke survivors. We investigated and verified the feasibility of a novel therapy intervention for gait rehabilitation. By combining FES with a split belt treadmill system, a novel FES control system was comprised. Stimuli from FES device were automatically triggered by their own gait phase estimation based on the treadmill motor current.

This study was to validate that the proposal for gait rehabilitation is a safe, feasible and promising intervention.

This paper is organized as the following. Section II explains the proposed gait phase estimation method involved in these experiments. Section III analyzes gait modal on treadmill. Section IV depicts FES control algorithm based on gait phase estimation. Section V contains experiment

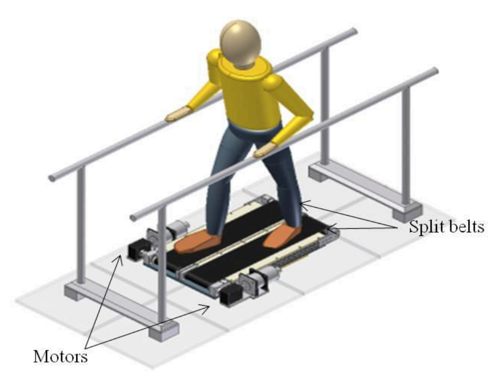

Figure 1. Treadmill-Walk for gait phase measurement.

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conduction, its results and discussion. Section VI shows a conclusion and future work.

II. Gait Phase Estimation Based on Treadmill Motor Current

Before establishing the FES control algorithm, it is necessary to understand the relation between treadmill motor current and estimation algorithm of gait phase. As shown in Fig. 1, for providing electrical power to each belt rotation, two DC motors were connected to the two split belts via gearboxes respectively. The handrails were set up to keep subjects from falling down accidentally. Motor current value and walk phase in both sides of the treadmill were measured and estimated. And the qualitative relation between them can be recognized. Motor current value was measured and recorded by a current sensor installed in each motor driver box. The current value fluctuated regularly up and down during gait phase. While no load was put on the belt during the swing phase, the motor current value existed almost constantly. During the stance phase, it increased to a peak and then decreased. By experiments, estimation algorithm of gait phase was applicable with acceptable 0.2 (s) error [11].

III. Gait Modal Analysis on Treadmill

A. Total Motor Current I_{Total}

The mechanical model of walking on the treadmill belt with a constant rotation velocity is specified in Fig. 2. While force F is applied onto the belt, motor current I will change correspondingly, and torque T generated from DC motor will be also altered automatically due to a velocity control feedback system controlling on the motor. The relation among the force F, motor current I and torque T are directly proportional to each other as formulated in (1).

$$Fv = UI = T\omega \tag{1}$$

Here v is the belt velocity, ω is rotation velocity, U is voltage.

While belts rotate without any extra load on it, motor current I is caused mainly by the torque loss T_{loss} in the gearbox. Other factors of causing current I consist of various forces applied to the belt, including friction force F_f between belt and friction reduction sheet on the walk board, and anteroposterior force F_y which is the kicking or braking force exerted by the subject's leg during walking procedure.

Integrating the factors mentioned above, the total motor current value I_{Total}, which could be detected by the current sensor in the motor driver box, is formulated as (2).

$$I_{Total} = I_{Tloss} + I_{Fy} + I_{F_f} \tag{2}$$

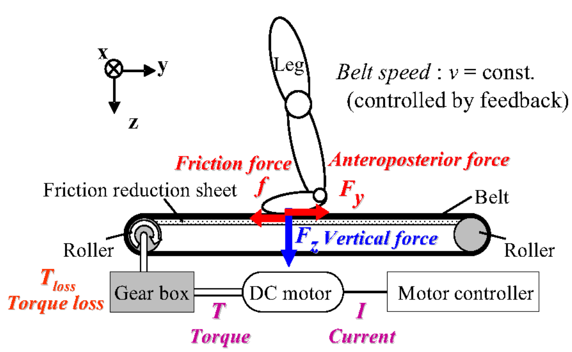


Figure 2. Mechanical model of walking on the treadmill belt.

where I_{Tloss} is the current value caused by torque loss T_{loss} in the gearbox, I_{Ff} is the current value caused by friction force F_f and I_{Fy} is the current value caused by anteroposterior force F_y.

B. Current I_{Tloss} Caused by Torque Loss in the Gearbox

The constant positive current value existed throughout the entire walk phase. Since T_{loss} occurs irrespective of other extra forces onto the belt, the constant positive current value, is considered as I_{Tloss}. T_{loss} can be formulated in terms of the treadmill belt velocity v, because T_{loss} mainly occurs in gearbox, the torque loss of the gear is mainly concerned with rotation velocity [12]. T_{loss} is related to non-reproducible factors, such as gear attrition, grease temperature and treadmill belt tension. In the previous study, I_{Tloss} was approximated with the current value I and the belt velocity v by experiments [13]. From the observed characteristic of I_i in this experimental result, I_{Tloss} could be approximated in second-order least squares formulation as (3).

$$I_{Tloss} = a_2 v^2 + a_1 v + a_0 \tag{3}$$

where a_i (i=0,1,2) is coefficient.

Because the friction force F_f is proportional to the normal force, I_{Ff} arises only when vertical force F_z from the subject is loaded on the belt during the stance phase. Because the direction of F_f is constantly opposite to the direction of the belt movement, in other words, it acts against to the driving movement of motor during the stance phase, increased part of motor current during the stance phase is considered as I_{Ff}. Therefore, if the time of increasing I_{Ff} can be measured, it is possible to identify and estimate the stance phase and swing phase of a patient's walk phase.

C. Current I_{Ff} Caused by Treadmill Belt Friction Force

The effect that, the friction loss I_{Ff} between the treadmill belt and the friction reduction sheet imposes on the current value, could be formulated as (4).

$$I_{Ff} = F_z \times P \tag{4}$$

$$P = \mu` / K_t \times n / r \tag{5}$$

where I_{Ff} is the current value caused by friction force F_f which occurs between the treadmill belt and the friction reduction sheet. F_f is computed through multiplying the dynamical friction coefficient μ' between the treadmill belt and the friction reduction sheet by the floor reaction force of the vertical direction F_z. With the constants determined by the characteristics of the treadmill, including torque fixed number of the motor K_t, axis shaft diameter r, the slowdown ratio of the gearbox n, P can be assumed and calculated as a constant shown in (5).

D. Current I_{Fy} Caused by Anteroposterior Force of Subject

Since anteroposterior force F_y is imposed onto the belt only when the subject's foot contacts it, I_{Fy} can be observed during the stance phase. More attention should be paid to the direction of F_y because it varies during the stance phase which is mainly a period from heel-contact to toe-off of gait. In the earlier part of stance phase, i.e., the heel-contact period, F_y acts forward (opposite to the direction of the belt movement, ie the negative direction on the y-axis in Fig. 2), causing the load on the motor increases and I_{Fy} has a positive value. However, in the later part of stance phase, i.e. the toe-off period, F_y acts backward (same as the direction of the belt movement, ie the positive direction on the y-axis in Fig. 2), causing the load on the motor decreases and I_{Fy} presents a negative value. When F_y acts backward strongly, I_{Fy} presents a largely negative value correspondingly, and the value of I_{Ff} is therefore partially offset by I_{Fy}. This procedure intervenes appropriately and precisely in estimating of the stance phase. Moreover, F_y tends to be wider in the positive and negative directions with heavy subject and fast walking velocity [14].

E. Threshold Current $I_{Threshold1}$ for Gait Phase Estimation

We have previously proposed an method for deciding the threshold current for estimating the gait phase.

Firstly, the algorithm approximates I_{Tloss} by belt velocity v, and can be formulated as $I_{Tloss}(v)$, which depends on the belt condition such as belt material, temperature and humidity. Therefore, $I_{Tloss}(v)$ should be calculated before the usage of the treadmill each time.

Secondly, the algorithm establishes a motor current threshold $I_{Threshold1}(v)$ by adding an offset to $I_{Tloss}(v)$ so as to decrease the affect of noise, as formulated in (6).

$$I_{Threshhold1}(v) = I_{Tloss}(v) + offset \tag{6}$$

As shown in (6), before determining the motor current threshold $I_{Threshold1}$, it is necessary to assure the adjustment of the *offset*. If the *offset* is too small, chattering noise will appear in the estimated walk phase; if it is too large, arising I_{Ff} will be buried partially within the offset and as a result, the walk phase estimation will be not accurate enough. Thus a function that can adjust appropriately the *offset* has been constructed in the system, $I_{Threshold1}(v)$ was determined based on measurement of motor current value of treadmill belt without load on it. It was approximated by the second-order least squares method, as more details found in [14].

Finally, through detecting and observing the motor current I_{Total}, the algorithm differentiates the stance phase and swing phase by determining whether I_{Total} exceeds $I_{Threshold1}(v)$ or not, as formulated in (7).

$$Ph = \begin{cases} Ph_{st},\ (I_{Total} \geq I_{Threshold1}(v)); \\ Ph_{sw},\ (I_{Total} < I_{Threshold1}(v)); \end{cases} \tag{7}$$

where Ph is walk phase of lower limb, Ph_{st} is stance phase and Ph_{sw} is swing phase.

During the periodic variation of I_{Total}`s value, once the $I_{Total}$ exceeds $I_{Threshold1}(v)$ in one side of the two split belts, the system will be starting to define the subject`s gait state as in stance phase in the contralateral belt side.

IV. FES Control Algorithm based on Gait Phase Estimation

Due to lower limb muscles functioning in different time during walking procedure, stimulation timing should be also considered in this way. The FES device facilitates two pairs of electrodes in two separate channels which are connected to TA and hamstrings to send electrical stimuli for gait assistance and rehabilitation. The strategies of triggering stimuli from the two channels are based on different gait state and conditions.

A. FES Triggered to TA

By connecting the FES device with the treadmill walking system, launching stimuli from an FES device to TA could be controlled according to the gait phase estimation. That is, once the gait phase of the sound lower limb of a subject with hemiplegia is distinguished as stance phase, the electrical stimulus from one of the two channels of FES device will be sent to TA of the affected lower limb for a period of time. The ideal stimulation sequence of TA should be sustained during the entire swing phase of the affected side.

B. FES Triggered to Hamstrings

While on normal symmetric walking speed, the beginning of terminal stance of a lower extremity is corresponding to the middle of mid-swing of the contralateral lower extremity [15]. The beginning of terminal stance is at the point of 50% of the entire stance phase while the middle of mid-swing is also at the point of 50% of the entire swing phase [15] [16]. Thus, when one of lower limbs is at the middle stance phase, the other lower limb should be simultaneously at the middle swing phase. During latter half swing phase, the hamstrings muscles are decelerating the thigh and preventing knee hyperextension [17] [18]. Therefore, hamstrings of affected lower limb should begin to be activated from the middle of stance phase of unaffected side.

Timing of the stimulation sequence to simulate a gait cycle is as shown in the Table I.

For automatically controlling on triggering stimulus from the other FES channel to hamstrings of the affected side, a current threshold $I_{Threshold2}$ for unaffected side here is necessary to be set up. The $I_{Threshold2}$ should satisfy that, when the real-time motor current meets the $I_{Threshold2}$, the stimuli for hamstrings of unaffected side will be launched at the time of half stance phase. The detailed construction of the current threshold $I_{Threshold2}$ for triggering hamstrings activity could be realized by statistic method as the following.

TABLE I. Treadmill motor current based FES control system

Percentage in a stride		62%	38%	
Gait phase		Stance phase	Swing phase	
Muscle stimulation	TA	OFF	ON	
	Hamstrings	OFF	OFF	ON

$I_{Threshold2}$ can be calculated through motor current statistic analysis of subjects` unaffected lower limb. $I_{Threshold2}$ for every step is self-adaptive and calculated after 10 steps of unaffected side. The motor current values at the half stance phase in previous 10 steps are recorded to compute the mean value of them, as shown in (8),

$$I_{Threshold2}(i) = \frac{1}{10}\sum_{n=i-11}^{i-1} I_{St_{half}}(n),\quad i \geq 11 \tag{8}$$

where $I_{St_{half}}(n)$ is the motor current value at the half stance phase of unaffected lower limb; St_{half} is the half stance phase; i is step counting. The first stimuli will be launched from the 11^{th} step.

The flow chart of FES control algorithm could be presented as Fig. 3.

Fig. 4 shows a block diagram of FES control based on the gait phase estimation with the bilateral separated treadmill. While the subject walks on the split belts, DC motors receives disturbance F_f, and are controlled by the treadmill controller. Then the gait phase will be determined by (7) in real time.

Finally, muscle stimulation will be manipulated by the controller after gait phase is detected.

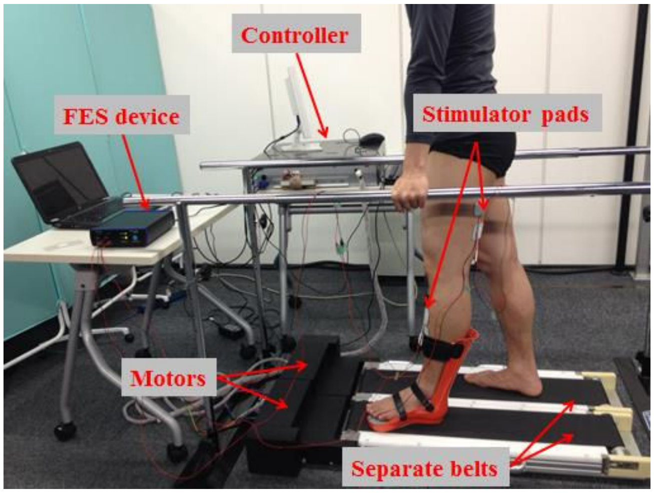


Figure 3. The experimental setup of gait training system

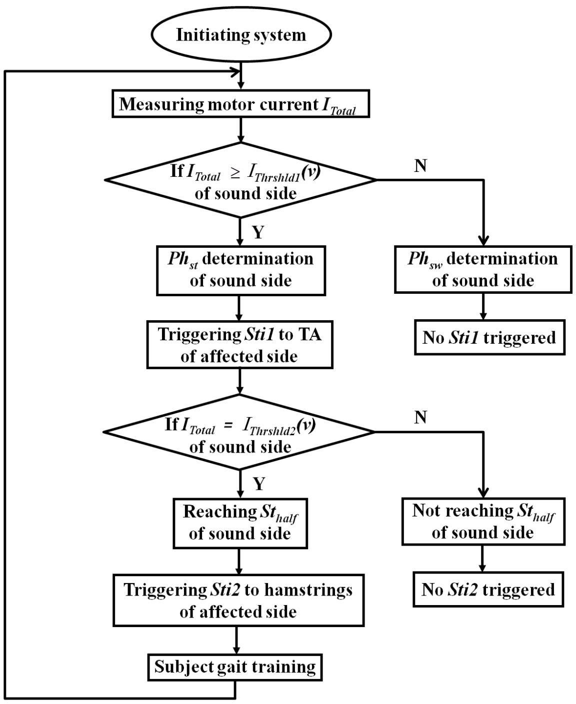


Figure 4. FES control algorithm for activating TA and hamstrings. Ph_{st} stands for stance phase, and *Sti1* is electrical stimulus to TA and *Sti2* to hamstrings.

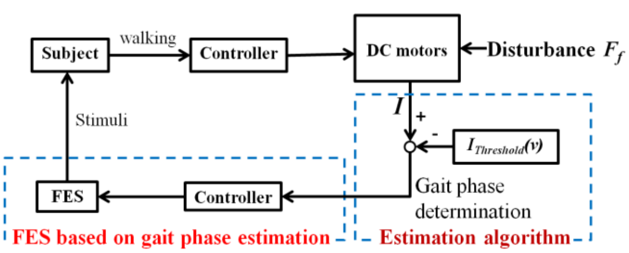


Figure 5. Block diagram of FES control based on the gait phase.

V. Experiment on FES Control Algorithm Based on Gait Phase Estimation

A. Objective

The objectives are to verify feasibility of the FES control algorithm based on the gait phase estimation. Also, it is to analyze characteristic gaits that affect the accuracy of the estimation and applicability for a variety of subjects without the need to tune the parameters individually for each patient.

B. Subjects

Three able-bodied subjects were recruited to simulate hemiplegic gait by wearing an ankle-foot orthosis (AFO), which was designed to function as a tool for simulation of hemiplegia. Because the F_z loading on the belt is related to subject`s body weight, the three subjects with considerately different weights from each other are chosen so as to analyze the application on various subjects. Before the experiments informed consent was obtained. During the experiment, handles were mainly used here for ensuring subjects` safety. Subjects walked without leaning on them as possible as they could. Their personal information is shown in Table II.

TABLE II. Information of Subjects

Subject	Personal Information			
	Gender	*Weight (Kg)*	*Simulated hemiplegia side*	*Age*
No.1	Male	55	left	28
No.2	Male	71	left	27
No.3	Male	87	left	23

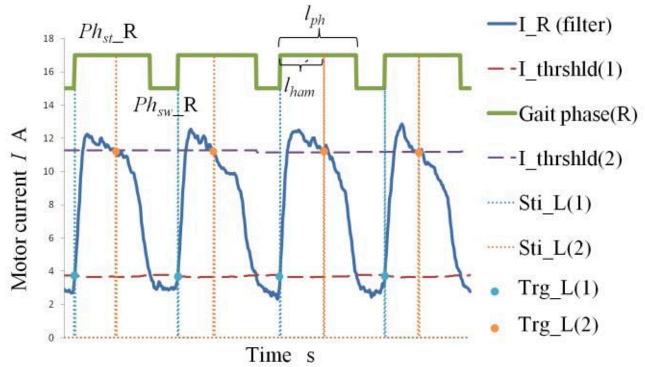
C. Methodology

Fig. 5 shows a subject wearing an AFO on the gait trainer treadmill with two pairs of separate electrodes attached to TA and hamstrings respectively.

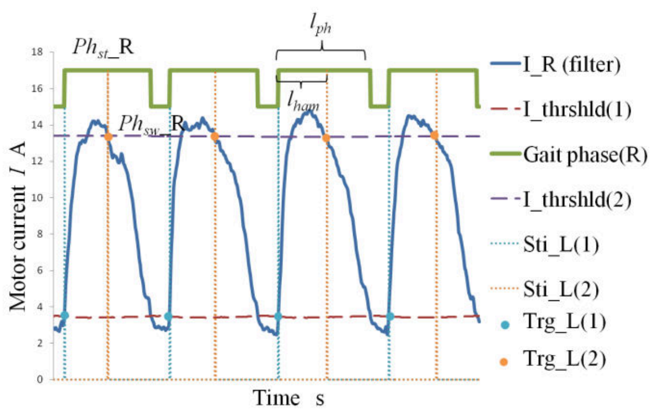
The DC motor of the treadmill was connected to a gearbox with a reduction ratio of 5 to 1, which drove the belt. A friction reduction sheet was placed under each belt. The velocity of the DC motor was controlled by a motor driver and could be set at a range from 0.0 to 4.0 (km/h). In this experiment, the left and right belt velocities were both set at 0.5, 1.0 and 1.5 (km/h). The three belt velocities were also set up to investigate influence of gait speed to the system application.

An FES device (STG4002, Multi Channel Systems MCS GmbH) was connected to the microcomputer of the treadmill with a trigger-in connector to receive commands of launching stimulus from treadmill controller. And on the other side, it was connected to the lower limb muscle with two pairs of non-invasive bipolar electrodes from two separate channels. Because lower limb muscles function differently during gait procedure, the quantity of stimulus to different muscles should be correspondingly treated with discriminatory attention. The stimulus from the FES device with a stimulation range from -8 to +8 (volt) was set at +7 (volt) and the pulse width was 400 (ms) for TA stimulation and 200 (ms) for hamstrings stimulation at each time of launch in this experiment. The waveform of the electrical stimuli is rectangular pulse.

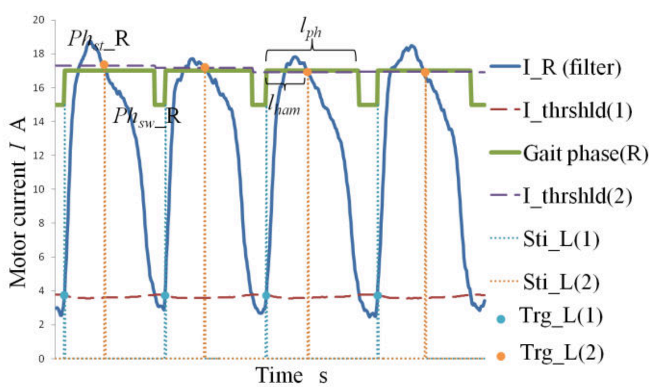
The left side of the subject was assumed as the affected side of hemiplegia, and the other side was considered the sound side. Since hamstrings are crucial to walking or running as it swings the leg forward into the ensuing step, and TA is the main muscle for foot dorsiflexion during walking, the electrical stimuli from the two channels of FES device will be sent to these muscles with surface stimulator pads respectively.



(a) One of FES control system experiment results, subject 1 at 1.0 (km/h).



(b) One of FES control system experiment results, subject 2 at 1.0 (km/h).



(c) One of FES control system experiment results, subject 3 at 1.0 (km/h).

Figure 6. I_R(filter) is the low-pass filtered motor current of the right(unaffected) belt side. I_thrshld(1) and I_thrshld(2) are the motor current threshold for stimulating TA and stimulating hamstrings respectively. Gait phase(R) is the walking phase of subject`s right side. Sti_L(1) and Sti_L(2) are the corresponding time point in gait phase when stimulus is launched. Trg_L(1) and Trg_L(2) are the FES timing for TA and hamstrings repectively. Ph_{st}_R and Ph_{sw}_R is stance phase and swing phase of right side. l_{ham} is the phase length from the beginning of stance phase to the phase point of hamstrings stimulation and l_{ph} is the entire stance phase length.

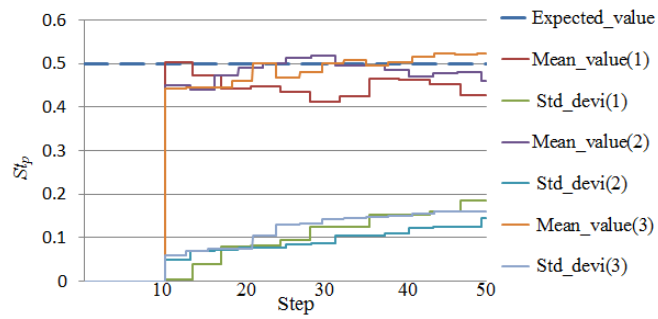


Figure 7. Part of experiment results: mean value and standard deviation of St_p from 3 subjects (1,2,3) during 50 steps at 1.0 (km/h) corresponding to Fig.6. Expected_value is 0.5, Mean_value is mean value of 50 St_p, Std_devi is standard deviation of of St_p.

Electrical stimulation timing was controlled by the FES control algorithm mentioned above.

The stimulation sites were determined via stimulation before experiments. While the subject was standing in an upright position, the electrode positions were found by trial and error until the best possible response to the stimulation was found.

Every subject tested ten times, 50 steps for each time, in three different belt velocities respectively.

D. Result and Discussion

Fig. 6 shows a classic group of stimulation results of three subjects at 1.0 (km/h). As the subject walking on the belts, the motor current presented periodicity and the gait phase including stance phase and swing phase could be estimated with our proposed estimation algorithm. At the moment that the gait phase was estimated to be stance phase during each step, a TTL signal from the treadmill controller was delivered to the FES device via it`s trigger-in input. That is, at the blue point in Fig.6 when the *I_R(filter)* just exceeded $I_{Threshold1}(v)$ of the unaffected lower extremity side, electrical stimulus was triggered to TA of the affected side.

As the motor current variation curve shows in Fig. 6, there could be two intersections between $I_{Threshold2}(v)$ and the real-time motor current. Though the $I_{Threshold2}(v)$ was intersected twice by *I_R(filter)* in each step, the FES control algorithm has been designed to identify the second intersection nearby the half stance phase as delivery time point of TTL signal. The orange point in Fig. 6 is the intersection of delivering TTL signal to FES device, and then the stimulus for hamstrings would be sent out. Also it indicates that the gait phase of left (affected) side is in half swing phase.

The three subjects` mean values of peak current are different from each other. This is caused by the different body weights. Fz loading on the belt is different, thus the amplitudes of motor current influenced by F_f are varied from each other.

The results also presented that, while walking at a higher belt speed, the current value curve in the middle area of stance phase was steeper and sharper than that while walking at a lower one. It is because the faster the subject walks, the more unstably steps occur. Therefore Fz varies more fiercely.

By previous research experiments, the gait phase estimation has an acceptable approximately 0.2 (s) error. So it is considerably acceptable for FES control. Here we mainly concentrate on analyzing the error of FES to hamstrings.

A parameter St_p for evaluating timing accuracy of stimulation is imported. St_p is the ratio of l_{ham} to l_{ph}. St_p could be calculated as (9).

$$St_p = l_{ham} / l_{ph} \tag{9}$$

where l_{ham} is the phase length from the beginning of stance phase to the phase point of hamstrings stimulation, and l_{ph} is the entire stance phase length .Obviously, if St_p is closer to 0.5, the stimulation timing is more precise.

Fig. 7 shows mean value of St_p and its standard deviation of 3 subjects at 1.0(km/h) corresponding to Fig.6. As walking proceeds, the mean value of St_p is approximating to 0.5, and the standard deviation is also around 0.2.

Fig. 8 shows the comparison of mean value of St_p and standard deviation from the experiment results of 3 subjects at 3 belt velocities. It indicates that while walking at 0.5km/h, the hamstrings stimulation timing presents a considerably lower accuracy comparing to that of other walking speeds. If body weight is heavier, the St_p is more possible to approach to 0.5 and the stimulation timing accuracy is higher while the standard deviation is larger.

Generally, it could be summarized that the proposed algorithm could be applied successfully to control FES for different patients without the need to tune parameters at a 1.0 or 1.5 (km/h) walking speed. However, results reveal that additional attentions should also be paid during walking procedure. Steps should avoid large gait fluctuation as much as possible, or the error of St_p would increase obviously and hamstrings stimulation timing prediction to next stride would also be affected negatively. During lower speed walking, *I_R(filter)* sometimes even did not intersect with the $I_{Threshold2}(v)$ and led to failure of trigger stimulation. This is resulted from the nearly flat current value curve in the middle area of stance phase. At this area, stance time is longer while walking slowly.

In summary, with the acceptable error, the proposed novel system is a safe, feasible and promising for gait rehabilitation.

VI. Conclusion and Future Work

This study describes a novel system combining FES with Treadmill-Walk to activate paralyzed muscles of lower limbs of hemiplegic patients. Experiments have preliminarily tested and verified feasibility of the proposed treadmill motor current based FES algorithm, nevertheless more accurate stimulation during walking and training results is expected to be obtained.

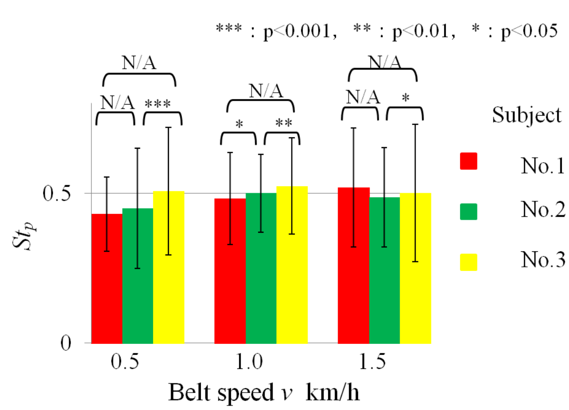


Figure 8. Comparison of mean value of St_p and standard deviation, 3 subjects at 3 belt velocities.

When the stimulus is triggered to the related muscles, the effectiveness of FES timing is also necessary to be assessed so as to obtain efficiently gait recovery of stroke survivors.

For restoring as close to normal gait pattern as possible by the end of rehabilitation programs, we will continue to investigate the accurate FES timing and electric quantity control of stimulation in the future. And the comparison of gait kinematics before and after experiments on hemiplegic patients also will be conducted to verify the effectiveness of the proposed method for gait rehabilitation.

References

[1] Dombovy M, The Evolving Field of Neurorehabilitation. Neurorehabilitation in Continuum Lifelong Learning in Neurology 2011, 17(3):443-448.
[2] Maynard FM, Reynolds GG, Fountain S, Wilmot C, Hamilton R, Neurological prognosis after traumatic quadriplegia. Three-year experience of California Regional Spinal Cord Injury Care System. J Neurosurg1979, 50(5):611-616
[3] Reding MJ, Potes E, Rehabilitation outcome following initial unilateral hemispheric stroke. Life table analysis approach.Stroke1988, 19(11):1354-1358.
[4] Jorgensen HS, Nakayama H, Raaschou HO, Olsen TS, Recovery of walking function in stroke patients: the Copenhagen Stroke Study. Arch Phys Med Rehabil1995, 76: 27-32.
[5] Kesar TM, Perumal R, Reisman DS, et al. Functional electrical stimulation of ankle plantar flexor and dorsiflexor muscles: effects on post stroke gait. Stroke 2009; 40:3821–7.
[6] Carolyn Kisner & Lynn A. Colby, Therapeutic Exercise: Foundations and Techniques. 5th ed. (2002) 692-693.
[7] Chantraine Alex, etc. "Shoulder Pain and Dysfunction in Hemiplegia: Effects of Functional Electrical Stimulation". Archives of Physical Medicine and Rehabilitation 80: 328–331. 1999.
[8] V. Dietz and J. Duysens, "Significance of load receptor input during locomotion: A review," Gait Posture, vol. 11, no. 2, pp. 102–110, 2000.
[9] Neckel, N., Wisman, W., Hidler, J., "Limb Alignment and Kinematics Inside a Lokomat Robotic Orthosis", IEEE EMBS, 2006, 2698 – 2701.
[10] M. Roerdink, C. JC Lamoth, G. Kwakkel et al.,"Gait Coordination After Stroke: Benefits of Acoustically Paced Treadmill Walking," Phys. Ther., 87(8), 2007, 1009-1022.
[11] Eiichi Ohki, etc. "Treadmill Motor Current Value Based Walk Phase Estimation", IEEE EMBS 2009, 7131- 7134.
[12] T. Yada, "Relation of Frictional Loose of Gear to Speed and Torque," Trans. Jap. soc. mech. eng., 38(313), 1972, 2396-2402.
[13] E. Ohki, Y. Nakashima, T. Ando, M. G. Fujie., "An Algorithm of Walk Phase Estimation with Only Treadmill Motor Current", the 2009 IEEE/RSJ International Conference on Intelligent Robots and Systems, pp. 4060–4066.
[14] Y. Ehara, S. Yamamoto, Introduction to Clinical Gait, Ishiyaku Publishers Inc., 2008, (in Japanese).
[15] Perry J. Gait Analysis: Normal and Pathological Function. Thorofare, NJ: Slack; 1992:149–167.
[16] Sivilingeniør Nils-Otto Negård, Controlled FES-assisted gait training for hemiplegic stroke patients based on inertial sensors, pp: 62-63, November, 2009.
[17] Michael G. Ciccotti, Robert K. Kerlan, Jacquelin Perry and Marilyn Pink, "An Electromyographic Analysis of the Knee During Functional Activities", American Journal of Sports Medicine, Vol.2, No.5, pp: 645-650, 1994.
[18] T. Yan, C. W. Y. Hui-Chan, L. S. W. Li, "Functional Electrical Stimulation Improves Motor Recovery of the Lower Extremity and Walking Ability of Subjects With First Acute Stroke-A Randomized Placebo-Controlled Trial", Stroke, pp:80-85 January, 2005.