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Development of a Novel Gait Rehabilitation System by integrating Functional Electrical Stimulation and a Split Belt Treadmill for Hemiparetic Patients after Stroke*

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***Abstract*—Nowadays, an increasing number of people with stroke are suffering considerably from a loss of physical mobility. Various traditional interventions have been developed to restore survivors' normal motor function following a stroke, but their effects are considerably limited. Many of these techniques require physical therapist's observation, specifically designed preparatory exercises and direct control of the lower limbs' position. Therefore, we propose a novel automatic gait training system for gait rehabilitation of hemiparetic patients. It integrates a split belt treadmill with a functional electrical stimulation (FES) device, which is used to improve gait quality by delivering electrical stimuli to the muscles. The delivery of the stimulus from the FES device is triggered automatically during gait cycle. As subjects walk on the separated treadmill, the gait phases are estimated by an algorithm that observes variation in the current values of the treadmill motors. Finally, we have preliminarily tested the feasibility of the proposed method through experiments on simulated hemiparetic subjects, by comparing with experimental results using force plates.**

I. INTRODUCTION

As many countries have entered an era of predominantly elderly societies, more and more people with mild or severe stroke are suffering considerably from a loss of physical mobility. The recovery of this lost motor function is not well addressed by biomedical treatments [1]. Typically, physiotherapy only has limited success in motor function restoration [2]–[4]. So far, many specific approaches require physical therapist's observation, and specifically designed preparatory exercises and direct control of the lower limbs position. Although much recent research has made great efforts toward improving motor function recovery, it still has a long way to go in enhancing the effectiveness of rehabilitation for a large number of stroke survivors with persistent deficits. Therefore, novel therapies and interventions are necessary.

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Currently, some biomechanical solutions are proving to be considerably effective. These include functional electrical stimulation (FES), which promotes restoration of motor function by stimulating the paralyzed muscles of stroke survivors. For example, FES is commonly used to address foot drop by applying it to the dorsiflexor muscles [5]. It has also been shown to improve walking ability recovery when applied to the quadriceps muscles as patients swing their leg forward for their next step [6]. High neural plasticity and repair mechanisms for restoring motor functions can be obtained by using FES, and the effectiveness of FES can be maintained for at least 24 months [7]. Therefore, in this paper we propose to use FES to improve the quality of gait by instantly influencing gait pattern. Precise timing of FES to muscles, however, is considerably difficult to control manually. Therefore, we use a special designed treadmill to accurately process and control the timing when FES is triggered.

Treadmill training has also become an established rehabilitation method for hemiparetic patients after stroke. The positive effects of this task-specific therapy have been shown in various studies [8]–[11]. The aim for hemiparetic patients using the treadmill is to correct for asymmetric physical ability, because the physical workload can be modified with the independent operation of the left and right treadmill [12]–[13]. In terms of the measurement of gait phases of hemiparesis, traditional methods, such as force plates and foot switches require long preparation. In addition, they place a burden on both patients and therapists, making their application considerably cumbersome. Thus, previously we proposed a novel method to measure gait phases via a gait-training robot, which consists of separated belt treadmills for each leg (shown as Fig. 1). To provide electrical power to each belt rotation, the two DC motors were connected to the split belt treadmills via gearboxes respectively. We developed a novel algorithm capable of estimating the walk phase of subjects by observing the treadmill motor current value [14].

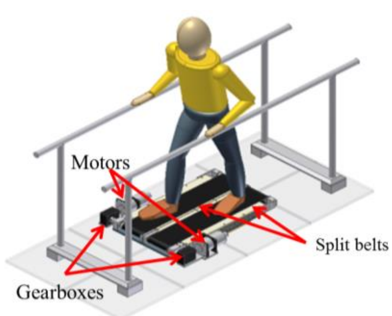


Figure 1. A split belt treadmill system for gait rehabilitation.

By combining FES with this treadmill system, we aim to promote more effectively the recovery of lower limb motor function in stroke survivors. Electrical stimuli from the FES device would be automatically calculated and applied to paralyzed muscles by a gait phase estimation algorithm based on the treadmill motor current. The feasibility of the novel gait training intervention, which combines FES to ankle dorsiflexor and quadriceps muscles with the treadmill system, was investigated and verified by comparison with gait results measured by force plates. The proposed approach is expected to have precisely controllable assistance, reliable repeatability with interactive feedback and quantifiable measures of subject`s performance. In addition, this approach reduces the amount of physical assistance required to walk.

The paper is organized as follows. Section II specifies the mechanical analysis of gait on the belt. Section III explains the proposed FES control algorithm based on gait phase estimation algorithm. Section IV presents the experimental setup for imitated hemiparetic subjects and the results. Section V contains the conclusion and future work.

II. Mechanical Model Analysis of Gait on Treadmill

A. Total Motor Current

The mechanical model of walking on the treadmill belt at a constant rotation velocity is shown in Fig. 2. The treadmill comprises two separate belts for the two lower limbs, two motors for rotating the belts, a pair of gearboxes, walk boards and friction reduction sheets placed between each belt and walk board. As the force F loaded onto the belt varies, the motor current I changes correspondingly. Meanwhile, because a velocity control feedback system controls the motor, the torque T generated from the DC motor also changes automatically. The force F, motor current I and torque T are directly proportional to one another as shown in equation (1).

$$Fv = UI = T\omega . \quad (1)$$

where v is the belt velocity, ω is the rotational velocity of the roller, and U is the voltage.

When the belts rotate without load, the motor current I is mainly caused by the torque loss (T_{loss}) in the gearbox of the treadmill's motor. Other factors that affect current I comprise various other forces applied to the belt. These forces include the anteroposterior force F_y, which is the kicking or braking force exerted by the subject's foot while walking on the belt, and the frictional force F_f between the belt and the friction reduction sheet on the walk board.

With the above factors taken into consideration, the total motor current I_{Total} can be formulated as:

$$I_{Total} = I_{Tloss} + I_{Fy} + I_{friction}, \quad (2)$$

where I_{Tloss} is the current value caused by the torque loss in the gearbox, I_{Fy} is the current value caused by the anteroposterior force exerted by the foot and $I_{friction}$ is the current value caused by friction between the belt and friction reduction sheet on the walk board. The value of I_{Total} is measured by the current sensor connected to the motor driver of the treadmill.

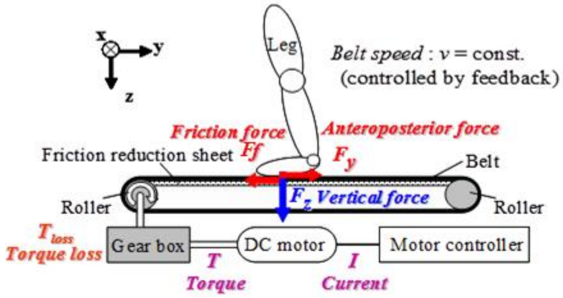


Figure 2. Mechanical model of walking on the treadmill belt.

B. Current Caused by Torque Loss in the Gearbox

The current I_{Tloss} necessary to compensate for T_{loss} is a constant positive value. Because T_{loss} occurs mainly in the gearbox and it is irrespective of other forces on the belt, I_{Tloss} exists throughout the entirety of the walk phases.

The torque loss of the gearbox is related to non-reproducible factors, such as gear attrition, grease temperature and treadmill belt tension, and it is also proportional to its rotational velocity, which can be formulated in terms of the treadmill's belt velocity v [15]. In our previous study [16], from the characteristic of I observed in the experimental results, I_{Tloss} can be approximated to a second-order polynomial using the least squares method:

$$I_{Tloss} = a_2 v^2 + a_1 v + a_0 , \quad (3)$$

where a_0, a_1, a_2 are constant coefficients. However, these coefficients depend on several variables such as belt material, temperature and humidity. Therefore, I_{Tloss} must be estimated before each use of the treadmill.

C. Current Caused by Treadmill Belt Friction Force

Because the frictional force F_f is proportional to the vertical force exerted by the subject, the current $I_{friction}$, necessary to compensate for the friction force, increases only when the vertical force F_z from the subject is loaded during the stance phase. Thus current $I_{friction}$ can be expressed as:

$$I_{friction} = F_z \times P , \quad (4)$$

$$P = \mu` / K_t \times n / r , \quad (5)$$

where F_f is computed by multiplying the dynamical friction coefficient μ' between the treadmill belt and the friction reduction sheet by the floor reaction force in the vertical direction (F_z). The remaining values are constants determined by the characteristics of the treadmill, including the torque of the motor K_t, roller shaft diameter r, and the reduction ratio of the gearbox n. P can be assumed constant.

The direction of F_f is the opposite to the direction of the belt movement; in other words, F_f acts against the driving movement of the motor during the stance phase. Therefore, if the time during which $I_{friction}$ increases can be measured, it is possible to estimate and recognize the stance phase and swing phase by observing current fluctuations during the subject's gait phase.

D. Current Caused by Subject's Anteroposterior Force

Considering that the anteroposterior force F_y is loaded onto the belt only during the stance phase, I_{Fy} can only be observed during this phase. However, more important is the direction of F_y, because it varies during the stance phase, which comprises the period of gait from heel-contact to toe-off. As shown in Fig. 2, in the earlier part of the stance phase, i.e., the heel-contact period, F_y acts opposite to the direction of the belt movement, causing the load on the motor to increase and I_{Fy} to have a positive value. However, in the later part of the stance phase, the toe-off period, F_y acts backwards (same as the direction of the belt movement), causing the load on the motor to decrease and I_{Fy} to have a negative value. When F_y acts backwards strongly, correspondingly I_{Fy} has a large negative value and the value of $I_{friction}$ is therefore partially offset by I_{Fy}. This procedure intervenes appropriately and precisely in the estimation of the stance phase. Moreover, F_y tends to be larger in the positive and negative directions with heavy subjects and fast walking velocity [17].

E. Threshold Current for Gait Phase Estimation

To establish the control algorithm for the FES stimulus, the relationship between the treadmill motor current and the gait estimation algorithm should be described first. For this, it is necessary to set up a motor current threshold $I_{Threshold}$. We propose an algorithm to estimate the gait phase of a subject walking on the belt as follows:

1. Approximate I_{Tloss} according to belt's velocity v
2. Establish a motor current threshold $I_{Threshold}$ by adding an offset to I_{Tloss}

so as to decrease the affecting of noise, as formulated in equation (6):

$$I_{Threshhold}(v) = I_{Tloss}(v) + offset\,. \tag{6}$$

As shown in equation (6), before determining the motor current threshold $I_{Threshold}$, it is necessary to fine-tune the adjustment of the offset. For a detailed explanation of the automatic offset setting function refer to [16].

Finally, via observing the motor current I_{Total}, the algorithm differentiates the walk phase of the lower limb between stance phase and swing phase by recognizing whether I_{Total} exceeds $I_{Threshold}$ or not:

$$Ph = \begin{cases} \text{stance;} & (if\ I_{Total} \geq I_{Threshold}) \\ \text{swing;} & (if\ I_{Total} < I_{Threshold}) \end{cases} \tag{7}$$

where Ph is the walking phase to be determined.

Motor current value was measured and recorded by a current sensor installed in each motor driver box, and the

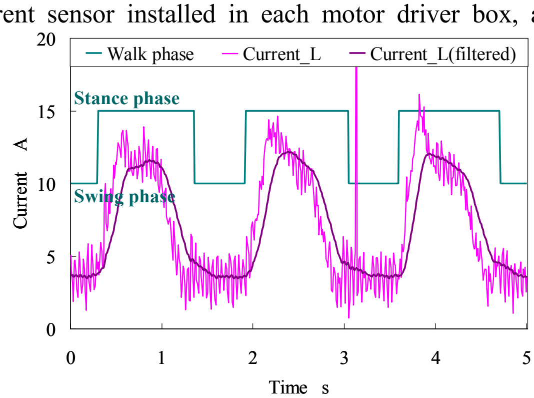


Figure 3. Walk phase estimation based on the motor current value.

qualitative relationship between them could be ascertained (as shown in Fig. 3). The current value fluctuated during each gait phase. During the periodic fluctuation of I_{Total}'s value, once I_{Total} exceeds $I_{Threshold}$ on one of the two belts, the system will define the subject's gait state as in the swing phase on the contralateral belt side. While no load was put on the belt during the swing phase, the motor current value was almost constantly non-zero because of the belt rotation. During the stance phase, it increased to a peak and then decreased.

III. FES Control Algorithm based on Gait Phase Estimation

To trigger automatically the stimulus, the FES control algorithm is clarified as below.

A. FES Triggered to Tibialis Anterior and Quadriceps

Tibialis Anterior (TA) is the main muscle for foot dorsiflexion and quadriceps are crucial to walk or run [18], as they swing the leg forward during walking, thus the electrical stimuli from the FES will be sent to these muscles. However, because lower limb muscles function at different times during walking, muscle stimulation timing is different. The FES device facilitates two pairs of non-invasive electrodes from two separate channels to send electrical stimuli to the two muscles respectively.

Because the gait phase estimation algorithm might fail to estimate the walk phase of hemiparetic patients when the affected side is used as reference [14], the FES control algorithm is based on the gait phase estimation of the unaffected side.

The stance phase of a lower limb is related to the swing phase of the other lower limb during gait cycle [19]. When heel contact of the unaffected side occurs, the pre-swing phase of the affected side will begin simultaneously. Because heel contact marks the beginning of the stance phase, electrical stimulus will be applied to the TA and quadriceps of the affected lower limb when the stance phase of the unaffected side is detected. The timing of the stimulation sequence is shown in Table I.

TABLE I. Muscle Activation Proportion During a Gait Cycle

Percentage in a stride		62%	38%
Gait phase		Stance phase	Swing phase
Muscle activation	**TA**	OFF	ON
	Quadriceps	OFF	ON

B. FES Control

The ideal stimulation sequence of the TA and quadriceps should be sustained during the entire swing phase of the

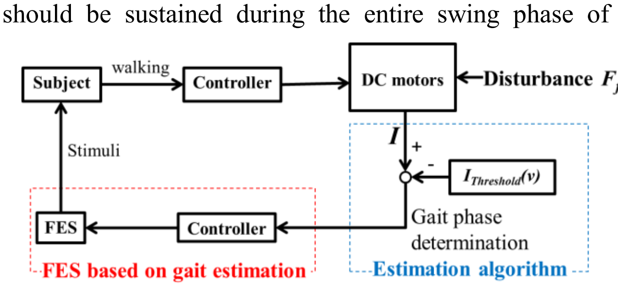


Figure 4. Block diagram of FES control based on gait phase estimation

affected side. Fig. 5 shows a block diagram of FES control based on gait phase estimation using the bilateral separated treadmill. While the subject walks on the split belts, the DC motors receive disturbance F_f, and are controlled by the treadmill controller. Then, the gait phase will be determined by equation (7) in real time. Finally, muscle stimulation will be manipulated by the controller after the gait phase is detected.

IV. EXPERIMENT ON FES CONTROL ALGORITHM

A. Objectives

The objectives of the present experiment were to verify the feasibility of control the FES rhythmically based on the estimated gait phase and to analyze the characteristic of the gaits that affect the accuracy of the estimation through comparing with the gait phase measured by plate forces.

B. Subjects

Three able-bodied subjects were recruited and walked on the treadmill while wearing an ankle-foot orthosis (AFO), which was designed to function as a tool for the simulation of hemiparesis. Because the vertical load on the belt is related to the subject's body weight, three subjects with considerably different weights were chosen so as to analyze the application across subjects. Before the experiments, informed consent was obtained. During the experiment handles were used for ensuring subjects' safety, but subjects walked without leaning on them if they possibly could. The physical characteristics of the subjects are shown in Table II.

TABLE II. SUBJECT INFORMATION

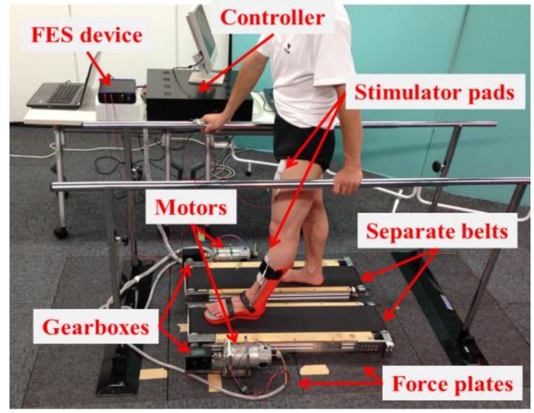
Subject	Personal Information			
	Gender	*Weight (Kg)*	*Simulated hemiparesis side*	*Age*
No.1	Male	55	left	28
No.2	Male	75	left	25
No.3	Male	87	left	23

C. Methodology

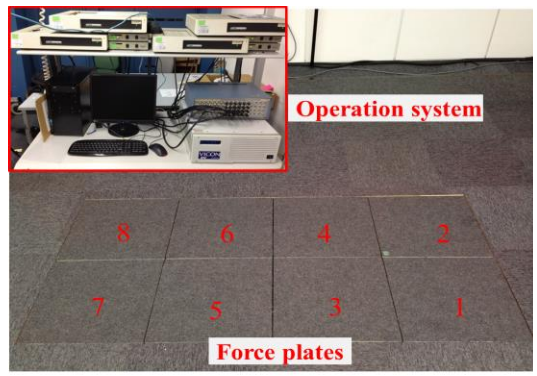
Fig. 5 (a) shows the FES system in integration with the split belt treadmill for gait rehabilitation. The DC motor of the treadmill was connected to a gearbox with a reduction ratio of 5 to 1. The belt speed could be set in the range of 0.0 to 4.0 km/h.

An FES device (STG4002, Multi-Channel Systems MCS GmbH, Reutlingen, Germany) was connected to the microcomputer of the treadmill with a trigger-in connector to receive trigger commands from the treadmill controller, and on the other side two output channels of it were connected to the lower limb muscles with two pairs of non-invasive bipolar electrodes. The stimulus from the FES device had stimulation amplitude in the range −8 to +8 V and was temporarily set at 7 V. The pulse width was 400 milliseconds with constant stimulation pulse.

The bilateral separated treadmill was placed on the central four of eight force plates (AMTI OR6-7 2000, Watertown, MA, USA), which measures the vertical load for the ideal FES timing to compare with the proposed FES timing method. There are totally 8 plates under the entire gait training system for detecting the F_z. As shown in Fig. 5 (b), The plate 3, 4, 5 and 6 are under the treadmill. The vertical force F_z to the belt was measured using the four force plates to. Force plate 3 and 5 are to detect the vertical force F_z of affected side. Force plate 4 and 6 are to detect F_z of the unaffected side. The data was sampled at a frequency of 100 Hz during experiments, the same as that of the current sensor of the treadmill motor.



(a) FES control system based on split belt treadmill



(b) Force plate system

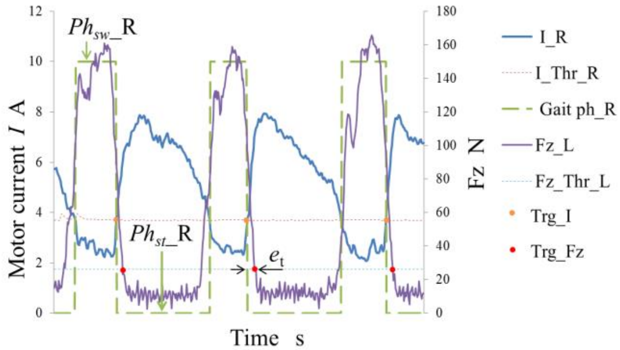
Figure 5. Experiment setup of FES control system

The optimal stimulation placement of the pads on the lower limb was determined by trial and error before the experiment took place. While the subject was standing in an upright position, the electrode positions were changed until the best possible response to the stimulation was found.

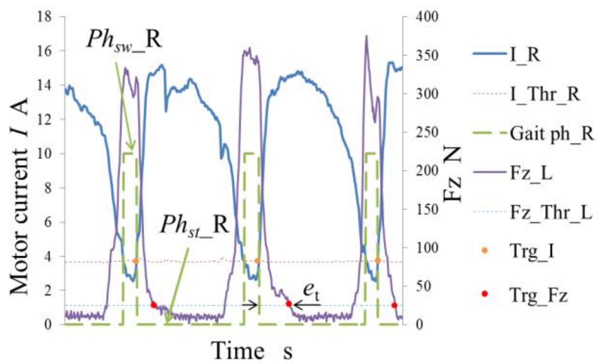
Every subject was tested 10 times, 40 steps each time at three different belt velocities (0.5, 1.0 and 1.5(km/h)) respectively, which is in consideration of appropriate acceptance of walking speed of patients after a stroke. There were at least 5 minutes intervals between each test of subjects.

D. Results and Discussion

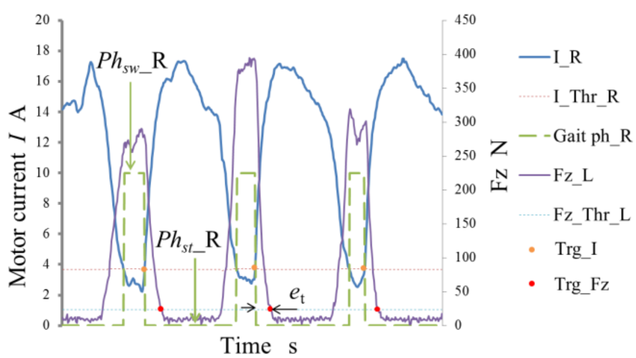
A representative results of the stimulation experiments of the three subjects at 1.0 (km/h) are shown in Fig. 6. During the subject`s walking on the belts, the motor current was periodic and the gait phase, including stance phase and swing phase, was successfully estimated. The orange points indicates that, as the motor current value varies at these time points, the value of I_R, which is the low-pass filtered current value, is just equal to the value of I_Thr_R on the sound side, which is the threshold of motor current value for recognizing the gait phase.



(a) One of the experimental results, subject 1 at 1.0 km/h.



(b) One of the experimental results, subject 2 at 1.0 km/h.



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

Figure 6. A part of the experimental results of FES control system. I_R is the low-pass filtered motor current on the right side of the two belts, i.e., the unaffected side. I_Thr_R is the threshold of motor current on the right side for gait phase determination. Gait ph_R is the walking phase of the subject's right side. Fz_L is the vertical force of the body weight on the left side measured by force plates. Fz_Thr_L is the threshold of recognizing the gait phase by the force plates. Trg_I is the FES stimulation trigger timing calculated by the motor current. Trg_Fz is the FES stimulation trigger timing calculated by the force plates. Ph_{st}_R is the stance phase of the right side and Ph_{sw}_R is the swing phase of right side.

They also indicate that the gait phase on the affected side is starting to be in the swing phase. Thus, a trigger signal from the treadmill controller was delivered to the FES device via its' trigger-in input. The FES device subsequently succeeded in sending voltage stimuli via electrode pads. The TA muscle of the affected side was stimulated to activate foot dorsiflexion, and the quadriceps of the affected was stimulated to achieve lifting of the leg simultaneously.

The three subjects' waves of current and vertical load values are different from each other. That is caused by their different body weights. Every subject's load on the belt is different in each step, thus the amplitudes of motor current and vertical force measured by force plates varies. Also, the three subjects' amplitudes and mean peak values of motor current differ from each other.

The results also show that, while walking at a higher belt speed, the current value curve in the middle area of stance phase was steeper and sharper. This is because the faster the subject walks, the more unstable steps are. Therefore F_z varies more rapidly.

There is a time difference of stimulation trigger between the trigger time point in estimated gait phase based on the motor current and the trigger time point in measured gait phase based on the force plates (red spots in Fig. 6). Here we denote the time difference as e_t in each step, which is the difference between the time when the stance phase is detected by the induced motor current based method, and the stance phase detection based on vertical load measurements:

$$e_t(n) = T_{Fz}(n) - T_I(n),\ 6 \le n \le 40 \tag{8}$$

where the $T_{Fz}(n)$ and $T_I(n)$ are the time of stimulation trigger of the two methods respectively and n is sequence number of steps in each single test. Because the subject needed to adjust to step on the belts at the beginning of walking, the first five steps were discarded, and n was counted from the sixth step in each test.

Fig. 7 shows the comparison of mean value of e_t and its standard deviation of 3 subjects at 3 belt velocities from the experiment results. While walking at 1.5km/h, the stimulation timing presented the smallest value of e_t comparing to that of other walking speeds. It implies that the FES timing by the proposed method is most accurate at 1.5km/h. This is a consequence of increasing belt's speed. When the belt's speed is increased, the step frequencies are increased correspondingly, then stance phase time and swing phase time are both decreased in every stride cycle. The gait phase time by the proposed method is therefore nearest to the measured one by force plates.

Before each test, the value of Fz_Thr_L was decided by selecting the maximum value of measured vertical force of the force plate 3 as the belt ran without any load on it. This is also for filtering the noise from the plate force itself. If Fz_Thr_L can mainly recognize the walk phase, the selection of its` value is not the decisive factor to exert influence on experiment results and e_t. It is because a small increase or decrease of Fz_Thr_L value under 40N just seldom influences the stimulation and e_t, as we also can see that in Fig. 6.

From Fig. 7, it can also be deduced that if weight is lighter, e_t is smaller. As a whole, the e_t is almost always smaller than 0.25 seconds, with the lightest subject having the smallest e_t, approximately lower than 0.08 seconds. On the other hand, there is a definition of pre-swing phase in normal gait cycle, which is from contralateral heel contact to ipsilateral toe-off (50% to 62% gait cycle) [20]–[21]. It implies that when the

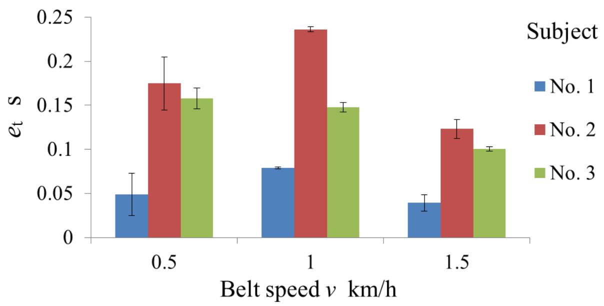


Figure 7. Timerror of stimulation control based on gait phase estimation and force plates for three subjects at three belt velocities.

contralateral heel contact occurs, the ipsilateral heel-off occurred or is oncoming in pre-swing phase. In other words, once heel contact occurs in the unaffected side, a percentage below 12% of a stride time in a gait cycle of the affected side should be in pre-swing phase. The red points, however, in the results are closer to the toe-off as shown in Fig. 6. The stimulation should be triggered at the heel-off time point and earlier than the time that the red dots determined by measured F_z. Therefore, the simulation time decided by our proposed system and e_t here is considerably acceptable and feasible for FES control.

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, because the threshold depends only on the treadmill but not on the subjects. However, results revealed that steps should avoid large gait fluctuation as much as possible, or the error would increase obviously, and the walking speed should be faster, but it might be more endurable and durative at a low walking speed for subject to conduct gait training. Thus, a medium walking speed might be the best choice for subject's gait training with the proposed system.

V. Conclusion and Future Work

In this paper we proposed a novel method to trigger muscle stimuli by FES in integration of a gait phase detection algorithm using a split belt treadmill system. The experiments tested and verified the feasibility of the proposed system. However, more accurate stimulations during walking and training results are expected to be obtained. When the stimulus is triggered and applied to the related muscles, it is also necessary to assess the effectiveness of the FES to more efficiently obtain gait recovery for stroke survivors. Moreover, because the lower limb muscles function differently during the walking procedure, the amount of stimulus to the different muscles should differ accordingly.

In the future, we will continue to investigate accurate FES timing and the electrical quantity control of stimulation based on kinematic analysis of the muscles involved in gait training.

Furthermore, the comparison of gait kinematics before and after experiments on patients after stroke will also be conducted to verify the effectiveness of the proposed method for gait rehabilitation.

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