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An Algorithm of Walk Phase Estimation with Only Treadmill Motor Current

Eiichi Ohki, Yasutaka Nakashima, Takeshi Ando, *Student Member, IEEE*,
Masakatsu G. Fujie, *Member*, *IEEE*

***Abstract* — To develop a gait rehabilitation robot for hemiplegic patients, quantitative evaluations of patient ability is needed. Patient's walk phase, which includes time balance of stance and swing legs, is one of the most useful indexes. However, conventional methods measuring walk phase require laborious preparations. In this paper, a novel algorithm estimating walk phase on a treadmill by observing DC motor current is proposed. In comparison of this algorithm and conventional methods, it was verified that the proposed algorithm had as the same accuracy as foot switch. Moreover, the proposed algorithm could estimate stance phase in 0.2 (s) errors between measurements of force plate mostly (4 out of 5 healthy subjects). However, result from the 5th subject showed that the proposed algorithm had erroneously identified stance phase as swing phase when little body weight loaded on leg. This characteristic is often observed in hemiplegic gait. Therefore, the proposed algorithm might need improvement of motor current threshold. However, this algorithm had capable of estimating the time of loading body weight on leg, and thus could be useful as a quantitative evaluation tool.**

I. Introduction

REHABIRITATION robots have become indispensable in aged society. Conventional rehabilitation methods require therapist's hard labor, and will become impossible to treat growing number of patients. In particular, strokes have been one of the main causes of physical disabilities and often produced hemiplegia [1]. For hemiplegic patients, mobility has strong correlation on quality of life [2]. Therefore, gait rehabilitation robot for hemiplegic patients is needed.

As gait rehabilitation machine which reduce therapists' labor, treadmills are often used [4]-[9]. Treadmill has merits such as ease of observing and analyzing patients, and feature of supporting severely disabled patients. One form of treadmill rehabilitation aims to correct the asymmetric physical ability of hemiplegic patients [6]-[9]. Among them, bilateral separated treadmill can be used in the treatment of mildly and severely disabled hemiplegic patients because

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Eiichi Ohki is with the Graduate School of Advanced Science and Engineering, Waseda University, Tokyo 169-8555, Japan (e-mail: e.ohki@asagi.waseda.jp). Yasutaka Nakashima is with the Graduate School of Advanced Science and Engineering, Waseda University, Japan. Takeshi Ando is with the Graduate School of Advanced Science and Engineering and the Faculty of Science and Engineering, Waseda University, Tokyo, Japan. Masakatsu. G. Fujie is with the Faculty of Science and Engineering, Waseda University, Tokyo, Japan.

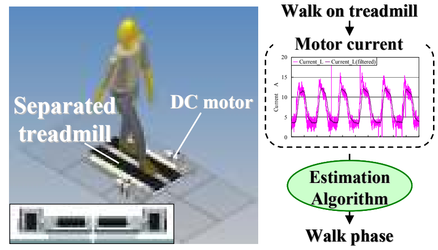


Fig. 1: Estimation of walk phase with only treadmill motor current.

physical workload can be modified with independent operation of left and right treadmill [8] [9]. Using a bilateral separated treadmill, we developed a rehabilitation robot capable of alleviating the asymmetry of body weight loading during walk of hemiplegic patients [10]. In a previous study, we concluded that asymmetry of loading body weight on simulated hemiplegia could be alleviated when velocity of affected side belt was reduced. In addition, we determined that symmetry of walk phase was correlated to ability of loading symmetric body weight to legs. As a result of our previous findings, we decided to develop a treadmill rehabilitation robot that could automatically adjust sound- and affected-side belt velocities depending on patient's walk phase. However, conventional methods to measure walk phase, such as foot switch and force plate, require a significant amount of preparation and impose a burden on both the patient and therapist [3].

The purpose of this paper is to propose an original method for measuring walk phase only with the rehabilitation robot based on separated treadmill, without wearing devices. This proposal utilizes a novel algorithm capable of estimating walk phase by observing current value of treadmill DC motor (Fig. 1). Motor current value to measure the interact force or torque has been widely used in robotics control studies [11]-[13], however our proposed algorithm has novelty in terms of applying for walk analysis and setting motor current threshold automatically with statistical approach in estimating walking condition.

As proposing the algorithm, this paper has 2 objectives. The first is to evaluate accuracy of walk phase estimation and the second is to investigate effects of various individual gaits in preparation of hemiplegic gait.

This paper is organized as follows. Section II explains the proposed the estimation algorithm. Section III describes an experiment of verifying the first objective with proving methodology, results and discussion. Section IV also relates experiment of verifying the second objective. Section V, the conclusion, provides a summary and outlines future work.

II. Walk Phase Estimation Experiment

A. Qualitative relation of walk phase and motor current

As a preparative of explaining the algorithm, this subsection shows the qualitative relationship of a patient's distinctive walk phase and the current value of a treadmill motor. In this study, a bilateral separated treadmill, which is a treadmill with completely separated left and right treadmill belts (as shown in Fig. 2), was used. The DC motor of the treadmill (SS60E6, Sawamura Denki Industrial Co., Ltd.) was connected to a gearbox with a reduction ratio of 5 to 1 (HY125R-005, Kyouiku Gear Mfg., Co., Ltd), which provided power to each belt of the treadmill. A friction reduction sheet was placed under each belt. The velocity of the DC motor was controlled by a motor driver (MS-100T15, Sawamura Denki Industrial Co., Ltd.) and could be set at target velocities ranging from zero to 4.0 (km/h).

In order to determine the qualitative relation between a subject's walk phase and the current value of the treadmill motor, it was necessary to measure both while the subject walked on the treadmill.

The experimental conditions were as follows. The treadmill was placed on a force plate (AMTI OR6-7 2000). The left and right belt velocities were set at 2.0 (km/h). The body weight of the subject was 80 (kg).

The walk phase was measured by the vertical force applied to the treadmill while the motor current value was measured by the current sensor installed in the motor driver. Sample walk phase measurements and motor current of the left treadmill belt are presented in Fig. 3. Figure 3 shows that constant positive motor current value existed throughout the walk phase and that motor current increased during the stance phase.

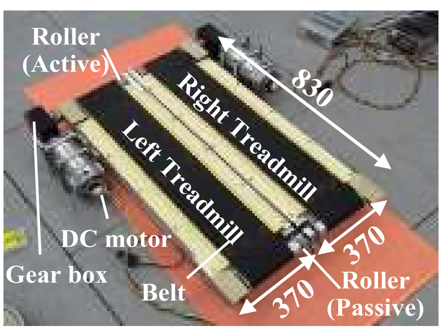


Fig. 2: Bilaterally separated treadmill.

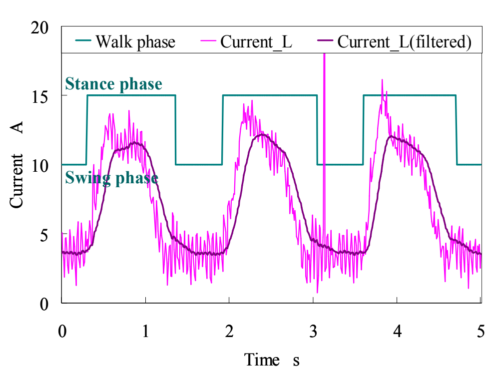


Fig. 3: Walk phase and motor current of left leg during subject walk.

B. Method to estimate walk phase from motor current

This subsection explains how to estimate a walk phase from a motor current value based on a mechanical model of the treadmill. Figure 4 shows the model and explains the variation of motor current that occurs when a subject walks on the treadmill. When force applied to the belt varies, generating torque T is automatically controlled and current I varies correspondingly. This is because the motor is controlled by a feedback velocity control system. According to the mechanical model, the torque loss T_{loss} that occurs in the gearbox is primarily the factor of raising I. As other factors of existing motor current I, the force applied to the belt is included. The force applied to the belt is classified into friction force between the walk board f, and into anteroposterior force F_y, which is the kicking and braking force exerted by the subject's leg(s) during walking movement. By considering the factors enumerated above, motor current I is formulated as (1).

$$I = I_{Tloss} + I_f + I_{Fy} . \tag{1}$$

where I_{Tloss} is current value that is caused by torque loss T_{loss}, I_f is current value that is caused by friction force f and I_{Fy} is current value that is caused by anteroposterior force F_y.

Because T_{loss} occurs irrespective of force to the belt, the constant positive current value that existed throughout walk phase shown in Fig. 3 is considered as I_{Tloss}. Supporting that T_{loss} mainly occurs in gearbox, I_{Tloss} can be formulated in terms of the belt velocity v, because the torque loss of the gear is mainly concerned with rotation velocity [14].

As the friction force f is proportional to normal force, I_f arises only when vertical force from a leg F_z is applied to the belt, which is during stance phase. Because the direction of f is forward (negative direction on the y-axis in Fig. 4) constantly and always becomes against to the motor during stance phase, motor current increase during stance phase in Fig. 3 is considered as I_f. Therefore, if the time of increasing I_f from I_{Tloss} can be measured, it is possible to identify and estimate the stance phase and swing phase of a patient's walk phase.

Because anteroposterior force F_y is only applied to the belt when the subject's foot is in contact, I_{Fy} can also be observed during stance phase. Attention has to be paid to the direction of the F_y, because the direction of F_y changes during a stance

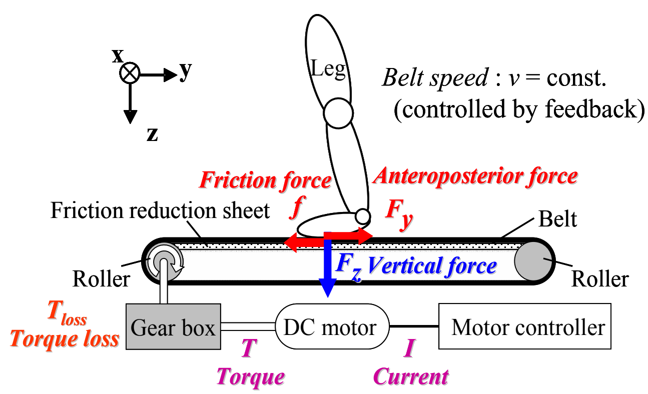


Fig. 4: Mechanical model of treadmill.

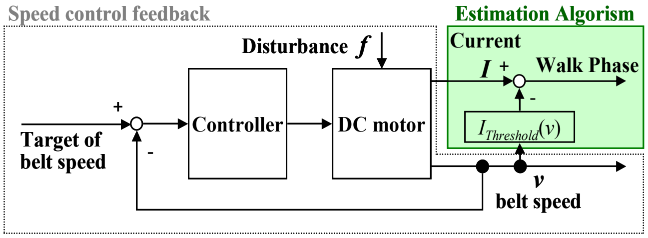


Fig. 5: Block diagram of treadmill with estimation algorithm.

phase [3]. In the earlier part of stance phase, F_y operates forward (negative direction of y-axis shown in Fig. 4). In this part, load on the motor increases and I_{Fy} has positive value. However, in the later part of stance phase F_y operates backward (positive direction of y-axis shown in Fig. 4). In this part, load on the motor decreases and I_{Fy} has negative value. When F_y operate backward strongly, I_{Fy} has too much negative value, and the value of I_f is canceled out by I_{Fy}. In this scene, appropriate estimation of stance phase is disturbed. This was not observed in Fig. 3. However verification for various gait types is possible because F_y tends to be wider in the positive and negative directions when the body weight of a subject is heavy, or the walking speed is fast [3].

Thus far, we have proposed an algorithm for estimating the walk phase of a subject as follows:

First, the algorithm approximates I_{Tloss} by belt velocity v, and formulates to $I_{Tloss}(v)$. Second, the algorithm constructs a motor current threshold $I_{Threshold}(v)$ by adding offset to $I_{Tloss}(v)$ in order to reduce affect of noise. Finally, the algorithm observes the motor current I and estimates walk phase by determining whether I exceeds $I_{Threshold}(v)$ or not, as in (2). Figure 5 shows a block diagram of the bilateral separated treadmill within the estimation algorithm.

$$I_{Threshold}(v) = I_{Tloss}(v) + \mathit{offset}\ . \qquad (2)$$

$$\begin{cases} if\ \ I \geq I_{Threshold} & then\ \ \textbf{\textit{Stance Phase}}. \\ if\ \ I < I_{Threshold} & then\ \ \textbf{\textit{Swing Phase}}. \end{cases}$$

C. Method to construct motor current threshold $I_{Threshold}$

In the proposed algorithm used to estimate walk phase, there are two problems that need to be solved in order to determine the motor current threshold $I_{Threshold}(v)$. The first problem is the change of approximation formula $I_{Tloss}(v)$ with the passage of time. Torque loss I_{Tloss} is related to non-reproducible factors such as gear attrition, grease temperature, and belt tension. As a result, when the state of the treadmill changes, $I_{Tloss}(v)$ will change. To solve this problem, we developed a system to determine $I_{Threshold}(v)$ automatically. By using this system, it was possible to determine $I_{Threshold}(v)$ just before measurement of walk phase . The second problem is the adjustment of the offset in (2). When the offset is too small, chattering noise can appear in estimated walk phase. However, when the offset is too large, arising I_f can be buried in the offset and walk phase estimation can be inaccurate. Because of this, a capability of adjusting the appropriate offset has been set in the system of determining $I_{Threshold}(v)$.

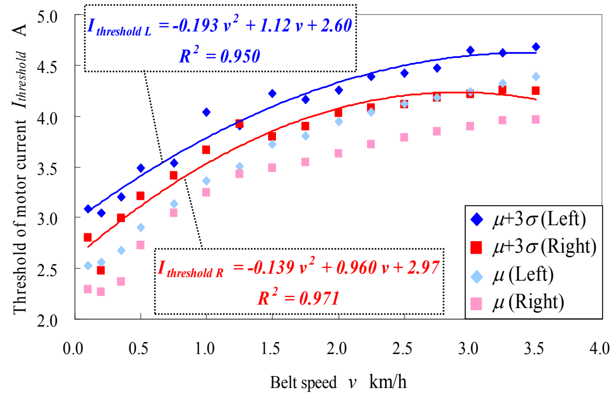


Fig. 6: Threshold of motor current $I_{Threshold}$.

The process of this system is as follows. The system rotates the treadmill belt with no load at velocity v_i (0.1 to 3.5 (km/h), 16 conditions), and measurements of motor current value are taken for 5 (s) at each condition. The system then calculates the average of motor current μ_i and the standard deviation σ_i for each velocity, and creates the data of the motor current as in (3).

$$I_i = \mu_i + 3\sigma_i\ , (i\text{=}0, 1, \ldots 15). \qquad (3)$$

Assuming the current value of each velocity follows normal distribution, Ii is upper limit of 99% confidence interval, and that is to say Ii exceeds 99% of motor current value in each velocity. We have anticipated that the offset has been adjusted just enough of noise distribution of motor current value. The system approximates $I_{Threshold}(v)$ with I_i and v_i as in (4). From the observed characteristic of I_i, second-order least squares method is chosen.

$$I_{Threshold}(v) = \sum_{k=0}^{2} a_k v^k\ . \qquad (4)$$

In regards, coefficient vector $\vec{a}$ is calculated as (5).

$$\vec{a} = \begin{bmatrix} a_0 \\ a_1 \\ a_2 \end{bmatrix},\ V = \begin{bmatrix} 1 & v_0 & v_0^{\ 2} \\ \vdots & \vdots & \vdots \\ 1 & v_{15} & v_{15}^{\ 2} \end{bmatrix},\ \vec{I} = \begin{bmatrix} I_0 \\ \vdots \\ I_{15} \end{bmatrix}.$$

$$\vec{a} = (V^T V)^{-1} V^T \vec{I}\ . \qquad (5)$$

As an example, Fig. 6 shows a constructed $I_{Threshold}(v)$. In Fig. 6, the determination coefficient of the left treadmill was 0.950 and that of right treadmill was 0.971. According to those determination coefficients, $I_{Threshold}(v)$ could be approximated in second-order formulate. The difference between the left and right of $I_{Threshold}(v)$ was due to the difference of the gearboxes and the treadmill mechanism.

III. Comparative Experiment of Measuring Method

A. Objective

As mentioned in Section I, this paper has two objectives. This experiment assumes the first objective, which is to evaluate accuracy of walk phase estimated by the algorithm, by comparing with conventional measuring methods.

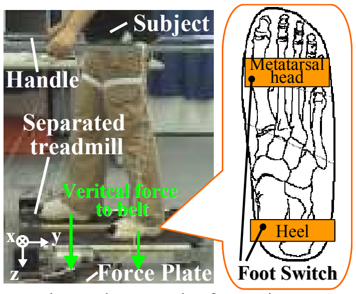


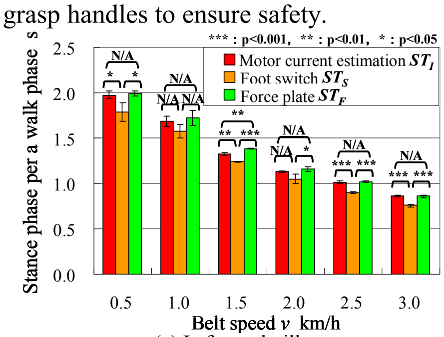
Fig. 7 Photograph of experiment.

TABLE I
EXPERIMENTAL CONDITONS OF BELT VELOCITY AND CADENCE

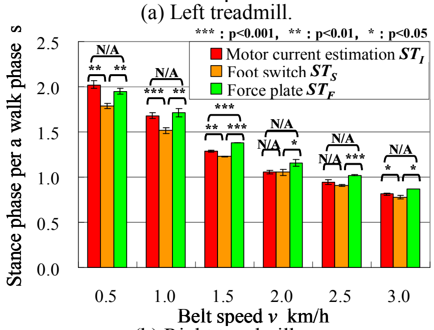
Belt velocity v km/h	Cadence step/min
0.5	43
1.0	53
1.5	62
2.0	72
2.5	82
3.0	91

B. Methodology

To verify the objective, we compared the original method using DC motor current value with the conventional methods which are foot switch and force plate. Experimental layout is shown in Fig. 7. The bilateral separated treadmill was placed on force plates (OR-6-7-200, AMTI). Subjects walked on the treadmill, wearing shoes with foot switches (151-BBW, Tapeswitch Japan), which were attached soles under metatarsal head and heel alike clinical methods [3]. Stance phase ST of walk phase was measured on each method. ST_I was estimated data with the proposed algorithm. ST_S was measured data using foot switch, and was time of one or more switches were stepped. ST_F was measured data using force plates, and was time of vertical force F_z exceeds 15 (N) (upper limit of 99% confidence interval in output F_z without load). This experiment was performed with 2 healthy subjects, subject A (body weight w=78(kg)) and B (w=54(kg)). The belt velocity v and cadence were set in range of use as shown in Table I, which were chosen based on the average Japanese gait [15]. Cadence was fixed with a metronome. At each v the measurement was conducted 3 times repeatedly. Each measurement was performed during 5 walk phases and ST was calculated as average. Before the experiments informed consent was obtained, and during the experiment subjects were let to grasp handles to ensure safety.



(a) Left treadmill.



(b) Right treadmill.
Fig. 8: Measured stance phase ST_I, ST_S and ST_F (subject A).

C. Results

The result of measured stance phase ST_I, ST_S and ST_F are shown in Fig. 8 and 9. Standard deviation indicated with error bar and significant difference analyzed with t-test is shown.

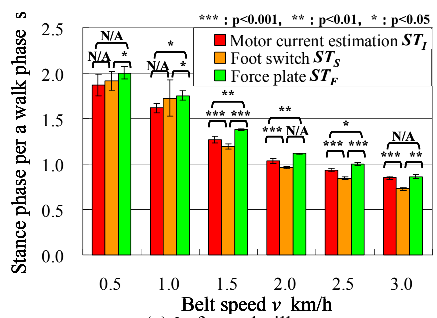
Between conventional methods, ST_S were significantly shorter than ST_F (22 in 24 conditions). The differences between ST_S and ST_F were in 0.1 to 0.3 (s). Between conventional methods and the proposed algorithm, individual differences were observed. In subject A, ST_S were mostly significantly shorter than ST_I (8 in 12 conditions), but ST_I and ST_F were not as different as ST_I and ST_S. In subject B, ST_I were mostly significantly shorter than ST_F (9 in 12 conditions), but ST_I and ST_S were not as different as ST_I and ST_F.

D. Discussion

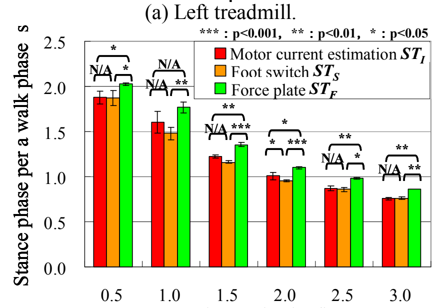
Firstly, we compared the foot switch with the force plate to measure walk phase. The force plates had higher accuracy than foot switch. This is because there is the gap between timing of foot contact and that of stepping on switch. Contrary, there is no gap using force plate between timing of foot contact and that of F_z rising. Therefore ST_F seemed higher in accuracy.

Secondly, we compared the proposed estimation algorithm with the conventional methods, that is, foot switch and force plate. The accuracy of the proposed algorithm depended on individual gait pattern. In the proposed estimation algorithm, rising motor current I caused by rising F_z was measured. Therefore, there is not timing gap such as the foot switch, because F_z was observed. However, there was another gap between the timing of I generated and that of I exceeding $I_{threshold}$, because of *offset* of $I_{threshold}$ as in (2). This gap could become large when subject gait had long time of small F_z.

The proposed estimation algorithm had lower accuracy than force plate. However, the result showed accuracy of the proposed algorithm was as the roughly same as that of foot switch.



(a) Left treadmill.



(b) Right treadmill.
Fig. 9: Measured stance phase ST_I, ST_S and ST_F (subject B).

IV. Experiment of Individual Walk Phase Estimation

A. Objective

This experiment assumes the second objective of this paper, which is to investigate effects of various individual gaits, and to determine whether this algorithm can be used to estimate the walk phase of hemiplegic patients.

The factors affecting the accuracy are forces from a leg to the belt while walking, F_z and F_y. Although there are individual differences, F_z depends primarily on body weight w [3]. And F_y mainly depends on body weight w and walk speed, which meets belt velocity v on the treadmill [3].

B. Methodology

To very the objective, this experiment was performed with shifting parameters body weight w and belt velocity v, in order to measure various gaits. To shift w, the experiment was performed with 5 healthy subjects. Table II shows body weight w of subjects. Belt velocity v and cadence were shifted as shown in Table I. Cadence was fixed with a metronome. Experimental layout is shown in Fig.10, which is the same as that of Section III except foot switches. In this experiment, walk phase measured with force plate was temporarily considered as the truth value, because the experiment of section III showed force plate had the highest accuracy in the compared other methods. Force to the treadmill F_z and F_y was measured using the force plates while subject was walking on the treadmill. Stance phase ST of walk phase was measured on each method. ST_I was estimated data with the proposed algorithm. ST_F was measured data with force plates, and was time of vertical force F_z exceeds 15 (N) (upper limit of 99% confidence interval in output F_z without load). At each v the measurement was conducted 3 times repeatedly. Each measurement was performed during 5 walk phases and ST was calculated as average. Before the experiments informed consent was obtained, and during the experiments subjects were let to grasp handles to ensure safety.

C. Results

We defined the time difference TD between estimated ST_I and measured ST_F as the accuracy index of estimated walk phase (6). The result of TD is shown in Fig. 11. The error bars indicate the standard deviation.

$$TD = ST_I - ST_F . \qquad (6)$$

In Fig. 11, TD were in -0.1 to 0.2 (s) for every belt velocity v except for subject #2. From this result, the error of estimation was within 0.2 (s) for about 4 subjects in 5.

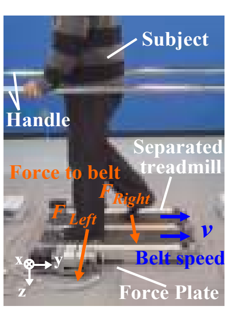


Fig. 10: Photograph of experiment.

TABLE II
BODY WEIGHT OF SUBJECTS

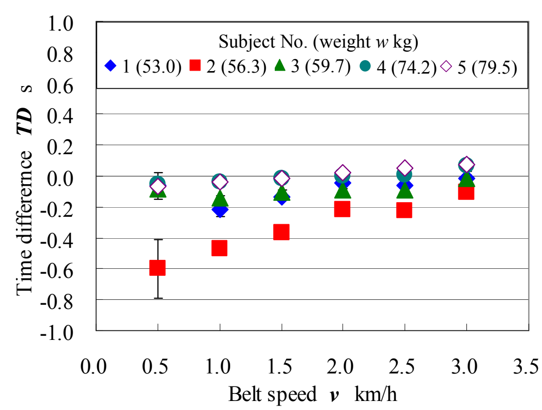
Subject No.	Body weight w kg
1	53.0
2	56.3
3	59.7
4	74.2
5	79.5

However, the TD of subject #2 was larger than the other subjects and showed especially small v on the left treadmill. This phenomenon was not observed in the TD of subject #1 or #3, both of whom had roughly the same body weight w as subject #2. This estimation failure was determined to have resulted from an individual gait difference.

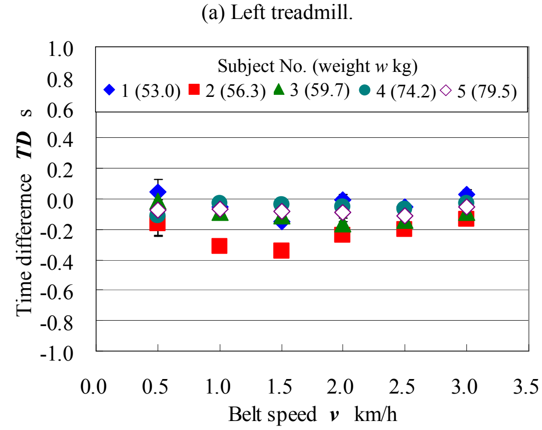
D. Discussion

To discover the characteristic gaits that affect the estimation, we compared F_y, F_z, ST_I and ST_F between subjects #1 and #2. In order to analyze large deference in each treadmill, we compared data on the belt velocity v that has the worst result of the estimating stance phase (v=0.5 (km/h) for left treadmill, and v=1.5 (km/h) for right treadmill).

First, the hypothesis that gait with long heel contact and long toe-off affects the estimation was raised from results of the left treadmill. Figure 12(a) shows F_y, F_z, ST_I and ST_F of subject #1, and (b) shows the same values of subject #2, on v=0.5 (km/h). In Fig. 12, the difference of F_z was observed between (a) and (b). Regarding the F_z of subject #2, it was found that rising and falling edges were longer than the edges of subject #1. This phenomenon indicates that the gait of subject #2 had long heel contact and long toe-off. In the time area of small F_z, I_f became small because friction force f was proportional to F_z. Therefore, it was believed that motor current I which was formulated in (1) became small and did not over $I_{threshold}(v)$ in this time area. ST_I and F_z of Fig. 12(b) support this hypothesis because the time area of F_z less than approximately 100 (N) was taken to be swing phase.

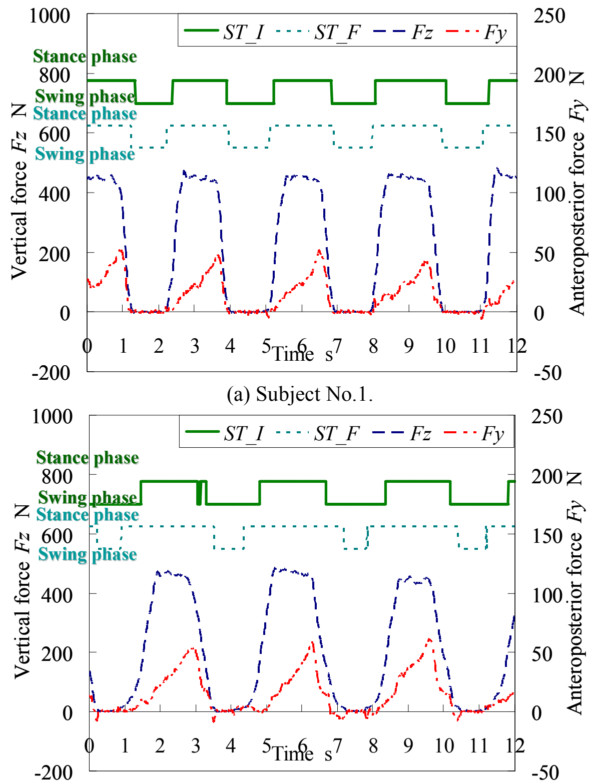


(a) Left treadmill.



(b) Right treadmill.

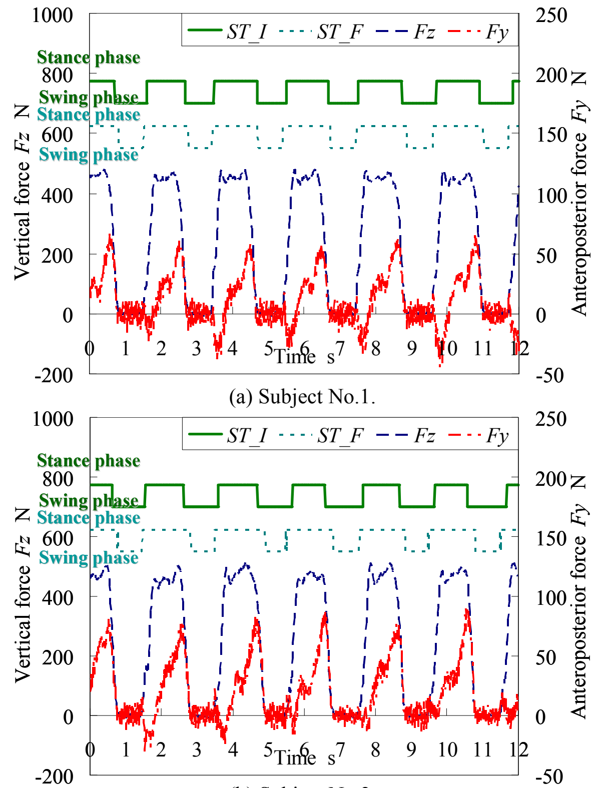
Fig. 11: Time difference between estimated and measured stance phase.



(a) Subject No.1.

(b) Subject No.2.

Fig. 12: Force to treadmill F_y, F_z and measured stance phase ST_I, (Left treadmill, v=0.5 km/h).



(a) Subject No.1.

(b) Subject No.2.

Fig. 14: Force to treadmill F_y, F_z and measured stance phase ST_I, (Right treadmill, v=1.5 km/h).

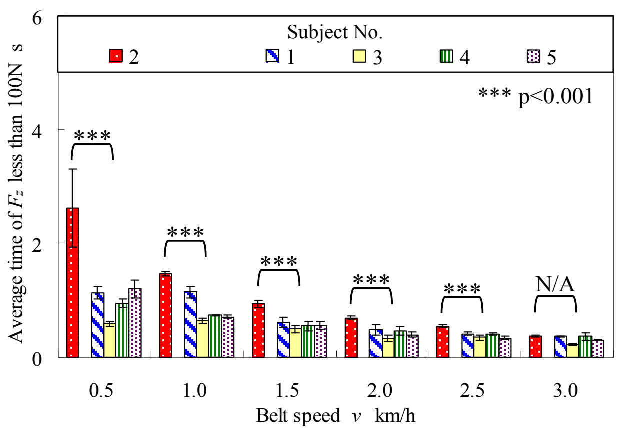


Fig. 13: Average time of F_z which was less than 100 N per walk phase, (Left treadmill).

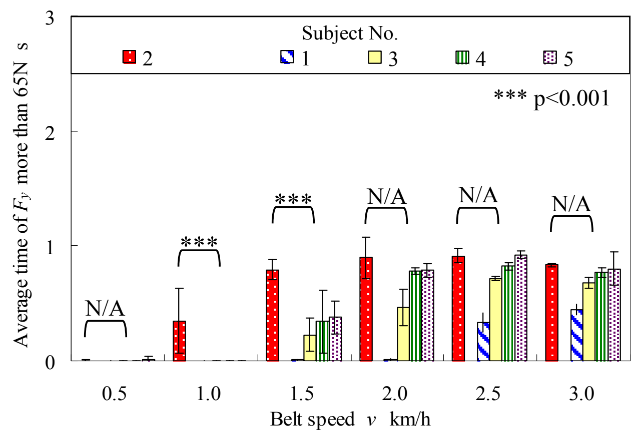


Fig. 15: Average time of F_y which was more than 65 N per walk phase, (Right treadmill).

This hypothesis was verified with results of the other subjects. Figure 13 shows the average time of F_z which was less than 100 (N) per walk phase about all subjects. In Fig. 13, the average time of subject #2 was larger than that of the other subjects, especially on low velocities. Ongoingly, we divided the subjects into two groups. First group had only subject #2 and second group had the other subjects. We found that there were significant differences in the average time for the F_z less than 100 (N) between the two groups on every velocity except 3.0 (km/h). Additionally, we found that as velocity v decreased, the difference between the two groups increased. This trend was also observed in the large TD of subject #2 as shown in Fig. 11(a). Therefore, it was concluded that estimations could be affected by the time area of small F_z, which was occurred due to the gait had long heel contact and long toe-off.

Next, the hypothesis that gait with strong kicking force on slow walk velocities affects the estimation was raised from results of the right treadmill. Figure 14(a) shows F_y, F_z, ST_I and ST_F of subject #1, and (b) shows the same values of subject #2, on v=1.5 (km/h). In Fig. 14, difference of F_y was observed between (a) and (b). Regarding the F_y of subject #2, the positive peaks were larger than that of subject #1. This phenomenon indicates that the gait of subject #2 had strong kicking force in the backward direction. In this time area, the nearby positive peaks of F_y, I_{Fy} became in negative on a grand scale because the load to the motor decreased. Therefore, it was believed that motor current I, which was formulated in (1) became small and did exceed $I_{threshold}(v)$ in this time area. The ST_I and F_z of Fig. 14(b) support this hypothesis because the time area of F_y more than approximately 65 (N) was taken to be swing phase.

This hypothesis was also verified with results of the other subjects. Fig. 15 shows the average time of F_y which was more than 65 (N) per walk phase about all subjects. In Fig. 15, the average time of subject #2 was larger than that of the other subjects, especially on middle velocity v. Ongoingly, we divided the subjects into two groups. First group had only subject #2 and second group had the other subjects. There were significant differences in the average time of F_y more than 65 (N) between the two groups on v=1.0, 1.5 (km/h) using the t-test. This trend was also observed in the large *TD* of subject #2 as shown in Fig. 11(b). Meanwhile, the average time of each subject for large velocity v was approximately as large as that of subject #2 on v=1.5 (km/h) in Fig. 15. However, this trend was not observed in the *TD* as shown in Fig. 11(b). This resulted because the positive peaks of F_y became impulsive at high velocities v, however the I was filtered by low pass (1st-order, cutoff frequency f_c=1.2(Hz)), which was set to remove impulsive current. Therefore, it was concluded that the estimation could be affected by the time area of large F_y especially in low velocity v, which was occurred due to the gait had strong kicking force in slow walk.

From above discussion, it was determined that gait with long heel contact and long toe-off or with strong kicking force on slow walk velocities could affect the accuracy of walk phase estimations. These affections were caused by the miss estimation when the algorithm compared I with $I_{threshold}$. For reducing these affections, it can be effective to optimize *offset* of $I_{threshold}$ by adjusting factor of standard deviation σ_i in (4) with design of experimental method, or to improve $I_{threshold}$ for changing according as timing of gait.

When hemiplegic walk phase is estimated with the proposed algorithm, the affection of small F_z can become problem. Since long heel contact and long toe-off are often observed in hemiplegic gait, which is caused by difficulty in loading and shifting body weight [16]. However, this algorithm could estimate time of loading body weight, because time of loading certain F_z was distinguished. Consequently, this estimate may be used as another evaluation value for the walk phase of hemiplegic patients. Meanwhile, affection of large F_y may be expected to have little effect, since F_y of hemiplegic gait tends to be smaller than that of a healthy gait [16] because of weak kicking ability.

V. Conclusion

We proposed a novel algorithm that can estimate walk phase of subjects by observing motor current value of a bilaterally separated treadmill. We found that estimation accuracy using proposed algorithm was as roughly same as that of foot switch. Also we found that the algorithm could estimate stance phase of walk phase with a maximum error 0.2 (s), compared to force plate, for about 4 out of 5 healthy subjects. Additionally, characteristic gait which has long heel contact and long toe-off or with strong kicking force on slow walk could affect accuracy of the algorithm. Because the former gait is often observed in hemiplegic gaits, provision will be need such as improving threshold of motor current $I_{threshold}$. On the other hand, the present algorithm could estimate the time of loading body weight on leg. Therefore, it is expected that the present algorithm become useful as another hemiplegic walk evaluation index.

In the future, we will develop a treadmill rehabilitation robot system that will set bilateral belt velocities automatically according to the ability of individual hemiplegic patients. The estimation algorithm developed in this paper will be used to measure ability of patient in the rehabilitation robot system.

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