31st Annual International Conference of the IEEE EMBS
Minneapolis, Minnesota, USA, September 2-6, 2009

Treadmill Motor Current Value Based Walk Phase Estimation

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

***Abstract*— We have developed a gait rehabilitation robot for hemiplegic patients using the treadmill. A walk phase, which includes time balance of stance and swing legs, is one of the most basic indexes to evaluate patients' gait. In addition, the walking phase is one of the indexes to control our robotic rehabilitation system. However, conventional methods to measure the walk phase require another system such as the foot switch and force plate. In this paper, an original algorithm to estimate the walk phase of a person on a treadmill using only the current value of DC motor to control the treadmill velocity is proposed. This algorithm was verified by experiments on five healthy subjects, and the walk phase of four subjects could be estimated in 0.2 (s) errors. However, the algorithm had erroneously identified a period of time in the stance phase as swing phase time when little body weight loaded on the subject's leg. Because a period of time with little body weight to affected leg is often observed in a hemiplegic walk, the proposed algorithm might fail to properly estimate the walk phase of hemiplegic patients. However, this algorithm could be used to estimate the time when body weight is loaded on patient legs, and thus could be used as a new quantitative evaluation index.**

I. Introduction

TREADMILL is often used for a gait rehabilitation of hemilegic patient. In addition, robotic gait rehabilitation systems using the treadmill have been studied [1]. The aim for hemiplegic patient to use the treadmill is to correct the asymmetric physical ability, because the physical workload can be modified with independent operation of the left and right treadmill [2][3]. Using a bilateral separated treadmill, we have been developing a rehabilitation robot capable of alleviating the asymmetry of body weight loading during the walk phase of hemiplegic patients [4]. In a previous study, we concluded that the asymmetry of body weight loading in simulated hemiplegia could be alleviated when the belt velocity on the affected side was reduced. In addition, the symmetry of a patient's walk phase was correlated to his or her ability of loading symmetric body weight to legs. As a result, we decided to develop a treadmill rehabilitation robot that could automatically adjust left- and right-side belt velocities depending on patient's walk phase. However, conventional methods to measure patient's walk phase, such as foot switches and force plate, require a significant amou-

Manuscript received April 7, 2009. This work was supported in part by Grant-in-Aid for Scientific Research (A) (20240058), Japan, and in part by the Global COE (Centers of Excellence) Program "Global Robot Academia", Waseda University, Tokyo, Japan.

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

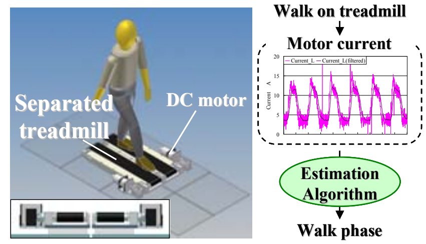


Fig. 1: Estimation of walk phase using the current value of a treadmill motor

nt of preparation and are cumbersome to use [5], which imposes a burden on both the patient and therapist.

The purpose of this paper is to propose an original method for measuring the walk phase only with the rehabilitation robot based on the separated treadmill. This proposal utilizes an original algorithm capable of estimating the walk phase of a patient by observing the current value of the treadmill motor (Fig. 1). This paper is organized as follows. Section II explains the proposed estimation algorithm. Section III proves an overview of the experimental methodology used during verification and shows results and discussion. Section IV provides a summary and future work.

II. Proposition of Walk Phase Estimation Algorithm

A. Qualitative relation of walk phase and motor current

A bilateral separated treadmill, which is a treadmill with completely separated left and right treadmill belts, was used. The DC motor was connected to a gearbox with a reduction ratio of 5 to 1, which provided power to each belt treadmill. The velocity of the DC motor 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 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. Measured walk phase and motor current of the left treadmill belt are shown in Fig. 2. Constant positive motor current value existed throughout the walk phase and that motor current increased during the stance phase.

B. Method to estimate walk phase from motor current

Figure 3 shows the treadmill mechanical model. 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.

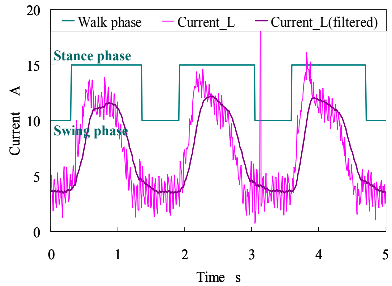


Fig. 2: Walk phase and motor current of left leg during subject walk

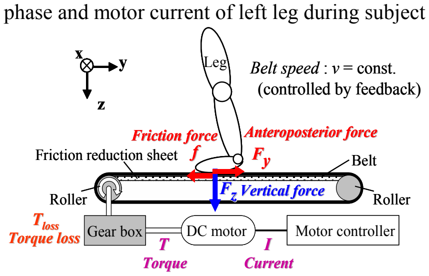


Fig. 3: Mechanical model of treadmill

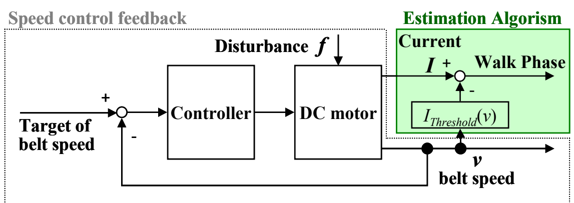


Fig. 4: Block diagram of treadmill with estimation algorithm

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 the walk phase shown in Fig. 2 is considered as I_{Tloss}. 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[6].

As the friction force f is proportional to the normal force, I_f arises only when vertical force from a leg F_z is applied to the belt, which is during the stance phase. Because the direction of f is forward constantly and always becomes against to the motor during the stance phase, motor current increase during the stance phase in Fig. 2 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 the stance phase. Attention has to be paid to the direction of the F_y, because the direction of F_y changes during a stance phase. In the earlier part of stance phase, F_y operates forward and the load on the motor increases and I_{Fy} has positive value. However, in the later part of stance phase F_y operates backward and the load on the motor decreases and I_{Fy} has negative value. When F_y operate backword strongly, I_{Fy} has too much negative value, and the value of I_f is denied by I_{Fy}. This disturbs appropriate estimation of the stance phase. F_y tends to be wider in the positive and negative directions with heavy subject and fast walking velocity [7].

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 $I_{Tloss}(v)$. $I_{Tloss}(v)$ depends on the belt condition such as temperature and humidity. Therefore, $I_{Tloss}(v)$ is measured before the usage of the treadmill each time. Second, the algorithm constructs a motor current threshold $I_{Threshold}(v)$ by adding offset to $I_{Tloss}(v)$ in order to reduce the affect of noise. Finally, the algorithm observes the motor current I and estimates the walk phase by determining whether I exceeds $I_{Threshold}(v)$ or not, as in (2). Figure 4 shows a block diagram of the bilateral separated treadmill within the estimation algorithm.

$$\begin{gathered} I_{Threshold}(v) = I_{Tloss}(v) + offset. \\ if \;\; I \geq I_{Threshold} \;\; then \;\; Stance\ Phase. \\ if \;\; I < I_{Threshold} \;\; then \;\; Swing\ Phase. \end{gathered} \tag{2}$$

In the proposed algorithm used to estimate walk phase, there is a problem that need to be solved in order to determine the motor current threshold $I_{Threshold}(v)$. The problem is the adjustment of the offset in (2). A function that can adjust the appropriate offset has been set in the system in order to determine $I_{Threshold}(v)$. $I_{Threshold}(v)$ was determined based on measurement of motor current value of treadmill belt with no load. $I_{Threshold}(v)$ was approximated by the second-order least squares method.

III. Estimation Experiment

A. Objective

The objectives are to verify accuracy of walk phase estimated by the algorithm for various gaits, to analyze characteristic gaits that affect the accuracy of the estimation, and to determine whether this algorithm can be used to estimate the hemiplegic walk phase.

B. Methodology

As shown in Fig. 5, the bilateral separated treadmill shown in Fig. 2 was placed on force plates (AMTI OR-6-7-200). The force to the treadmill F_z and F_y were measured using the force plates and the motor current value I of each treadmill was measured. Then, using the algorithm, the estimated time of the stance phase ST_I from I was measured and be compared with the stance phase ST_F that was measured from F_z.

The experiment was performed using five healthy subjects. The factors affecting accuracy are forces from the leg to the belt while walking, F_z and F_y. Although there are individual differences, F_z depends primarily on body weight, and F_y de-

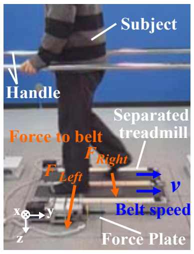


Fig. 5 Photograph of experiment

TABLE I
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

pends primarily on walk speed, which is the belt velocity and body weight. The weights of subjects #1, #2, #3, #4 and #5 are 53.0, 56.3, 59.7, 74.2 and 79.5 (kg). As shown in Table I, cadence was fixed with a metronome in order to simplify the experiment. Conditions of cadence were chosen based on the average Japanese gait. The measurement of the force and the motor current was performed three times repeatedly at each belt velocity v. Each measurement was performed during five walk phases. Before the experiments, informed consent was obtained from each subject, and each subject was allowed to grasp the treadmill handles to ensure safety.

C. Results

We defined the time difference TD between ST_I and ST_F as the index of accuracy for the estimated walk phase. The result of TD is shown in Fig. 6. Figure 6 (a) shows the TD of the left treadmill while (b) shows TD of the right treadmill.

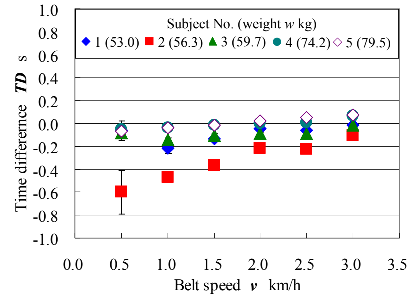
$$TD = ST_I - ST_F \tag{3}$$

As can be seen in Fig. 7, the values of TD were in -0.1 to 0.2 (s) for every belt velocity except for #2. The accuracy of estimation was within 0.2 (s) for about four subjects in five. However, the TD of #2 was smaller than the other subjects and showed especially small velocity on the left treadmill. This phenomenon was not observed in the TD of #1 or #3, both of whom had about the same weight as #2. This estimation failure was determined to have resulted from an individual gait difference.

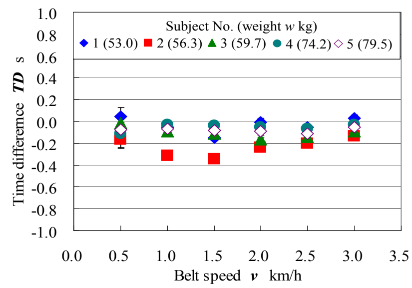
D. Discussion

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

First, we analyzed the gait characteristics in the left treadmill. Figure 7(a) shows F_y, F_z, ST_I and ST_F of #1 while (b) shows the same information for #2, on v=0.5 (km/h). In Fig. 7, the difference of F_z was observed between (a) and (b). Regarding the F_z of #2 shown in (b), it was found that the rising and falling edges were longer than the edges of #1 shown in (a). The gait of #2 resulted in long heel contact and long toe-off. In the time area of small F_z, I_f became small because friction force was proportional to F_z. Therefore, it was believed that motor current that was formulated in (1) became small and did not go over $I_{threshold}(v)$ in this time area. In Fig. 7(b), the time area of F_z less than approximately 100 (N) was estimated to be the swing phase. Figure 8 shows the



(a) Left treadmill

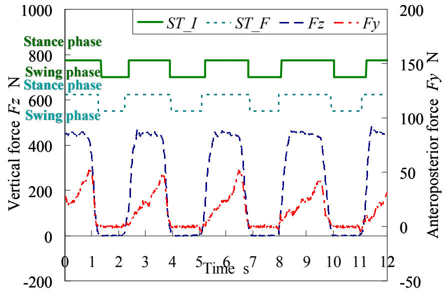


(b) Right treadmill

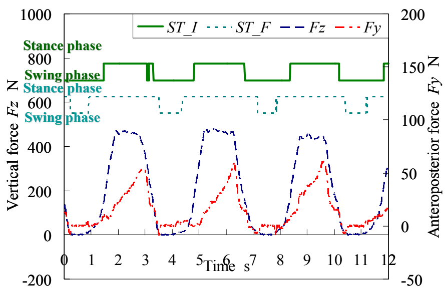
Fig. 6: Time difference between estimated and measured stance phase Note that The error bars indicate the standard division.

average time that was F_z smaller than 100 (N) during the walk phase for all subjects. In Fig. 8, the average time of #2 was larger than the times for the other subjects, especially at low velocities. there were significant differences in the average time for the F_z smaller than 100 (N) between #2 and other subjects on every velocity except 3.0 (km/h). Additionally, as velocity decreased, the difference between the two groups increased. Therefore, it was concluded that estimations could be affected by the time area of small F_z, and that this occurred due to characteristic gaits that had long heel contact and long toe-off. Gaits with long heel contact and long toe-off are often observed in the walk of hemiplegic patients who often have difficulty in loading and shifting body weight [3]. Therefore, it was determined that the gait characteristics of hemiplegic patients could affect to the accuracy of walk phase estimations. However, this algorithm can estimate times of loading body weight by means of the motor current I.

Next, we analyzed the affective gait characteristics that were observed on the right treadmill. Figure 7(a) shows F_y, F_z, ST_I and ST_F of #1 and (b) shows the same values for #2, on v=1.5 (km/h). Figure 8 shows that F_y differs between (a) and (b). Regarding the F_y of #2 shown in (b), the positive peaks were larger than the ones observed for #1 shown in (a). This indicates that the gait of #2 had strong kicking force in the backward direction. In the time area, nearby positive peaks of F_y, I_{Fy} became in negative on a grand scale because the load to the motor decreased. Therefore, motor current I, which was formulated in (1), became small and did exceed $I_{threshold}(v)$ in this time area. In Fig. 7(b), the time area of F_y larger than approximately65 (N) was taken to be the swing phase. Figure 10 shows that the average time of F_y larger than 65 (N) in the walk phase. In Fig. 10, the average time of #2 was larger than ones for the other subjects on low velocity. There were significant differences in the average time of F_y larger than 65



(a) Subject #1



(b) Subject #2

Fig. 7: Force to treadmill and stance phase(Left treadmill, v=0.5 km/h)

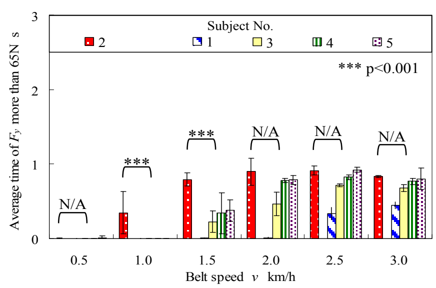
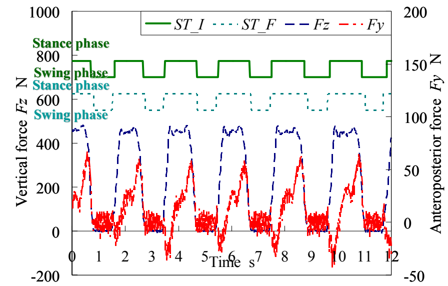
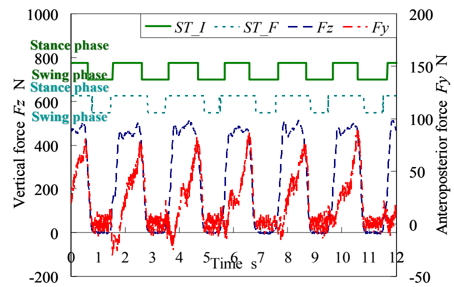


Fig. 8: Average time of F_z which is more than 65N (Left treadmill)



(a) Subject #1



(b) Subject #2

Fig. 9: Force to treadmill and stance phase (Right treadmill, v=1.5 km/h)

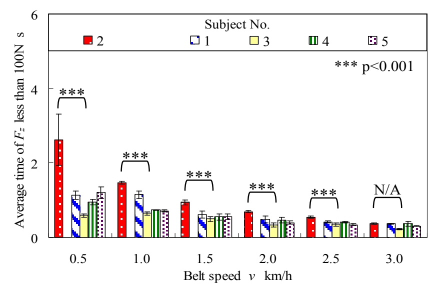


Fig. 10: Average time of F_y which is less than 100N (Right treadmill)

(N) between #2 and other subjects on v=0.5~2.5. This trend was also observed in the accuracy difference of the left treadmill shown in Fig. 6(b). This resulted because the positive peaks of F_y became impulsive at high velocities, however the I passed low pass filter. Therefore, the effects on the estimation could be caused by the time area of the large F_y in low velocity, and that this occurred as a result of the characteristic gaits that had strong kicking force. In the gait of a hemiplegic patient, the F_y tends to be smaller than that of a patient with a healthy gait [9]. Therefore, F_y is not expected to have a significant effect on efforts to estimate the walk phase of hemiplegic patients.

IV. Conclusion

We proposed an original algorithm to estimate the walk phase using the motor current value of a bilaterally separated motor treadmill. Stance phase of the walk phase could be estimated within 0.2 (s) error for about four out of five healthy subjects. Additionally, characteristic gaits with long heel contact and long toe off or with strong kicking force at slow walk velocities could affect to the accuracy of estimations. The former gait is often observed in patients with hemiplegic gaits. However, the algorithm would estimate the body weight load time. This estimation might possibly be useful as another evaluation index.

In the future, we will develop a rehabilitation robot system with bilateral belt treadmill which will use the estimation algorithm to control the belt velocity.

References

[1] Neckel, N.; Wisman, W.; Hidler, J., Limb Alignment and Kinematics Inside a Lokomat Robotic Orthosis, IEEE EMBS 06, 2006, 2698 – 2701.

[2] 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.

[3] T. Tani A. Ouchi, M. Fujie, "Development and Evaluation of Walk Training Robot System", J. Soc. Biome., 22(4), 1998, 169-173.

[4] E. Ohki, T. Ando, M. Nihei, M. G. Fujie, "Walking support robot "Tread Walk" for alleviating asymmetricity of hemiplegic walk -Effect of walk speed difference with separated treadmill-," Robomec 2008, 1P1-D07, 2008(in Japanese).

[5] Y. Ehara, S. Yamamoto, Introduction to Clinical Gait, Ishiyaku Publishers Inc., 2008, (in Japanese).

[6] T. Yada, "Relation of Frictional Loose of Gear to Speed and Torque," Trans. Jap. soc. mech. eng., 38(313), 1972, 2396-2402.

[7] S. Hesse, C. Bertelt, M. T. Jahnke, et al., "Treadmill Training With Partial Body Weight Support Compared With Physiotherapy in Nonambulatory Hemiparetic Patients," Stroke, 26, 1995, 976-981.

[8] N. Sekiya, H. Nagasaki, H. Ito, T. Furuna, "Correlation between Walk Speed, Cadence and Stride in Normal Walk", J. Jap. Phys. Ther. Association, 21(2), 1994, 416, (in Japanese).

[9] A. Tamai, M. Takami, T. Matsuyama, et al.,, "Pattern Classification of Floor Reaction Force Corresponding to Walking Pattern in Hemiplegic Gait," J. Jap. Phy. Ther. Asso.,13(5), 1986, 357-365 (in Japanese).