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Gait-controlled Mobility-aid Robot: Treadmill Motor Current Based Anteroposterior Force Estimation Using Frictional Model Reflects Characteristics of Ground Reaction Force

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***Abstract*— We have been developing a new mobility-aid robotic vehicle, "Tread-Walk 2 (TW-2)", which supports walking for the elderly. TW-2 is controlled by the natural walking movement, especially using ground reaction force during gait. In our previous work, we tried to estimate the user's anteroposterior force from the motor current value without a force sensor. But, a user of this vehicle experienced some discomfort both when he started walking and when he stopped walking. This problem is caused by inaccurate estimation of the user's anteroposterior force at the heel contact and the toe off. The estimation of the user's anteroposterior force is greatly related to inaccurate estimation of the vertical component of the ground reaction force, which is approximated by the square waveform in the stance phase. In this paper, we proposed the new approximation that reflects the characteristics of the ground reaction force during the human gait. This paper describes the novel method to approximate the waveforms of the vertical forces as isosceles trapezoidal waves. By comparing the estimated anteroposterior force using the new method with the measured value using the force plate, for two young subjects whose physical characteristics were different, the waveform pattern of the estimated force was found to be similar to that of the measured force. This showed that the proposed method could possibly be useful for accurate estimation of anteroposterior force. The technology to estimate the user's anteroposterior force accurately is a key to construct a control algorithm to improve the operability of TW-2.**

I. INTRODUCTION

ADVANCED countries have entered an era of predominantly elderly societies in recent years. The growing population of the elderly will require many kinds of mobility-aid devices. Many new devices have already been studied for supporting the mobility of the elderly [1], [2]. From the viewpoint of both preventive care and mobility aid, we have been developing a new robotic vehicle called

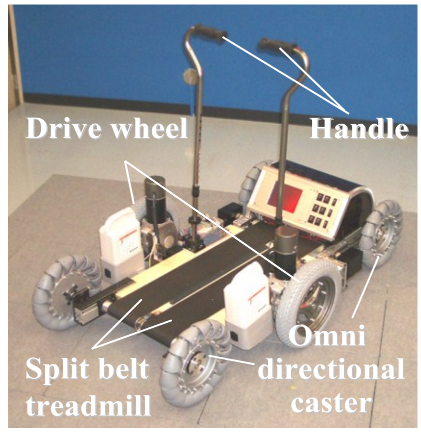

Fig. 1 Tread-Walk 2

"Tread-Walk 2 (TW-2)", which is controlled by the walking movement (Fig. 1) [3], [4].

TW-2 amplifies the walking velocity on the treadmill to the driving wheel to expand the user's range of travel. TW-2 estimates the user's desired walking velocity from the user's anteroposterior force and drives the treadmill belt. TW-2 uses active treadmill velocity control, which allows the user to change treadmill and mobility velocity by changing kicking force loading to treadmill, similar to the way velocity is changed in natural walking.

Real-time estimation of the user's anteroposterior force accurately (Fig. 2) is important In order to control treadmill belt , we used the method which usedthe current value of the DC motor that drives the treadmill belt without a force sensor [6].This concept contributeto reduce cost and size of TW-2 because any other sensor is needed.

The major technical problem to realize accurate anteroposterior force estimation is friction. We reported a mechanical model that considered the frictional forces influencing the current value of the DC motor in our previous study. We removed these modeled frictional losses from the current value of the DC motor and the vertical forces estimation with *** waveform model in order to accurately estimate the user's anteroposterior force. On conducting usability test, the user felt a sense of discomfort both when he/she started and stopped walking. This problem is caused by inaccurate estimation of the user's anteroposterior force at the heel contact and the toe off (Fig. 2). The inaccurate estimation of the user's anteroposterior force is greatly related to the frictional model. This model is using the approximation of the

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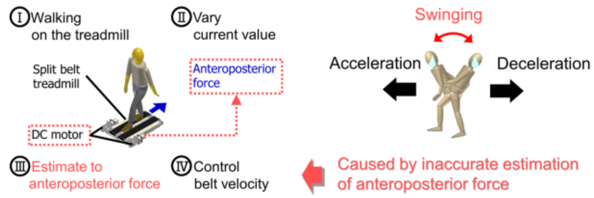


Fig. 2 Control treadmill system and Problem of operability

ground reaction force in the vertical direction. The comfort in starting and stopping walk movement is key technology for the TW-2 to be used by the elder.

The motivation of this study is to develop a novel algorithm to estimate the anteroposterior force more accurately. Specifically, we propose the method using novel frictional model that approximates the vertical forces by the isosceles trapezoidal waveform. This is quite different from the previous method which comprised square waveform approximation of the vertical forces. This method is proposed because the waveform of the vertical force generated during walking by elderly people, who form the target users of TW-2, is generally trapezoidal [7]. The objectives of the present article is to propose and evaluate the method. We conducted the result of the experiment with both previous and novel method to compare these results.

II. Mechanical Modeling to Estimate Anteroposterior Force

A. Mechanical model considers frictional force of treadmill

Fig. 3 shows the model and explains the variation in current values that occurs when the user walks on the treadmill. The DC motor connected to the treadmill detects the acceleration and deceleration forces applied to the surface by the user through the load current values of the DC motor. The belt velocity is increased when the current value of the DC motor decreases; conversely, the belt velocity is decreased when the current value increases. The DC motor actuates the treadmill belt and simultaneously acts as a sensor device. When the forces applied to the treadmill belt vary, the generating torque T is controlled automatically and the current value I varies correspondingly. This is because the DC motor is controlled by a feedback velocity control system. According to a mechanical model, the torque losses T_{loss} that occur in the gearbox are primarily the factor for raising I. Other factors for the existing current value I are the forces applied to the belt. These forces are classified into the frictional forces f under the treadmill belt, and the anteroposterior forces F_y which are the kicking and braking forces exerted by the user's leg(s) during the walking movement. By considering the factors shown above, the current value I is formulated as in (1).

B. Model of frictional loss in the gearbox

I_{gear} is the current value caused by torque loss T_{loss} in the gearbox. T_{loss} is related to non-reproducible factors such as gear attrition, grease temperature, and treadmill belt tension.

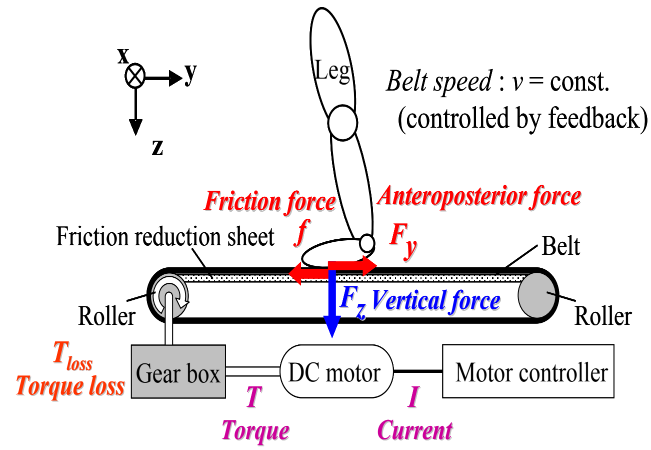


Fig. 3 Mechanical model of treadmill

Considering that T_{loss} mainly occurs in the gearbox, I_{gear} can be formulated in terms of the belt velocity v, because the torque loss of the gear is mainly concerned with rotation velocity [8].

In the previous study, I_{gear} was approximated with the current value I, and the belt velocity v by the experiment that moved the treadmill belt with no load at velocity v_i [5]. From the observed characteristic of I in this experimental result, I_{gear} could be approximated in second-order formulate as (2).

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

C. Model of frictional loss under the treadmill belt

I_{belt} is the current value caused by the frictional force f that occurs between the treadmill belt and the walking board. The direction of f is constantly forward (negative direction on the y-axis in Fig. 4), and is always opposed to the rotation of the DC motor. f is computed by multiplying the dynamic frictional coefficient μ' between the treadmill belt and the walking board by F_z, the floor reaction force in the vertical direction. Therefore, with the torque coefficient K_t, axis shaft diameter r, and the slowdown ratio of the gearbox n, the influence of frictional loss I_{belt} under the treadmill belt on the current value is formulated as (3).

$$I_{belt} = \mu' \times F_z / K_t \times n / r \tag{3}$$

III. Proposed Method to Estimate Anteroposterior Force

The qualitative relation between the anteroposterior force F_y and the current value I_{Fy} caused by only the anteroposterior force is formulated as (4), with the torque coefficient K_t, the axis shaft diameter r, and the slowdown ratio of the gearbox n.

$$F_y = K_t \times I_{Fy} \times n / r \tag{4}$$

We estimated the anteroposterior force from the current

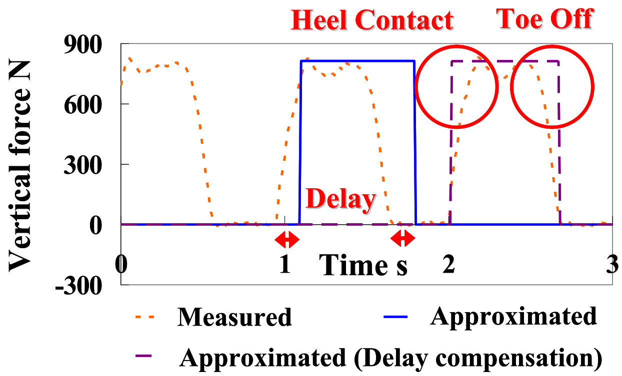


Fig. 4 Vertical forces in the previous method

value of the DC motor by substituting (2) and (3) that modeled in Section II into (4). K_t, r and n are known constant values; however, μ' and F_z are unknown values. In this study, μ' is replaced with the measured value of 0.12 from our previous study [5]. Therefore, if the vertical force is also known, it is possible to estimate the anteroposterior force from the current value of the DC motor.

However, the vertical force is unknown because TW-2 is not equipped with a force sensor. So, it is necessary to develop a method to estimate the vertical force.

In our previous study, we developed the method to estimate the vertical force without a force sensor [6]. The method is to approximate the waveform of the vertical force F_z as a square wave by adding the user's weight as a parameter to the stance phase estimated from the current value of the DC motor. Fig. 3 shows the waveform of the vertical force that is approximated in the previous study. The approximated vertical force has the delay that occurred in the estimation of the stance phase with the DC motor current value. So, we shifted the vertical force in the experimental result in order to compensate for the delay.

However, it was found that there was the error at the Heel contact and Toe off in the previous study, because the waveform of the vertical force measured by actually using the force plate in the experiment is nearer the trapezoidal wave than the square wave. The waveform of the floor reaction force in the vertical direction varies with the walking velocity [9]. As the subject's walking velocity increases, this waveform approaches the square wave [9]. Conversely, as the subject's walking velocity decreases, this waveform approaches the trapezoidal wave [9]. Therefore, it is necessary for the new method to approximate the vertical force that corresponds to the walking velocity.

We have proposed a new method in which the approximation considers the waveform of the vertical force changed corresponding to the walking velocity on the treadmill in order to accurately estimate the user's anteroposterior force from the DC motor current value. As the first step of our study, we have developed the method to approximate the waveform of the vertical force as the isosceles trapezoidal wave from the two viewpoints. We showed in the next paragraph about the two viewpoints.

First, the target users of TW-2 are elderly people. Especially, the waveforms of the floor reaction force in the vertical direction during the walk of an elderly person do not have bimodal peaks, and are nearly trapezoidal in shape [7]. Secondly, the user typically walks at a velocity between 1.0 km/h and 2.0 km/h on the treadmill in TW-2.

However, in this new proposed method, there are two problems that need to be solved. The first problem is to determine the absolute value of the vertical force. In this study, the absolute value of the vertical force uses the subject's body weight Wg because the vertical force depends primarily on body weight [10]. The second problem is the estimation of the walk phase. There are various methods to estimate the walk phase, such as using a wired foot switch or a force plate. However, these methods have their drawbacks. For example, the method using the wired foot switch may restrain the user's walking movements. Also, the method using the force plate may be too large. Therefore, in this study we used the method to estimate the walk phase by the current value of the DC motor, which has the advantage of not constraining the user's walking movements on the treadmill and of the size being small [11].

Fig. 5 shows a block diagram of the walk phase estimation algorithm used in this study. We have used an algorithm for estimating the walk phase of a subject as follows: The algorithm constructs a motor current threshold $I_{Threshold}$ by adding an offset to I_{gear} that is modeled in Section II in order to reduce the effects of noise. The algorithm observes the motor current I and estimates the walk phase by determining whether I exceeds $I_{Threshold}$ or not, as in (5).

$$\begin{aligned} &I_{Threshold} = I_{gear} + offset\,. \\ &if\ \ I \geq I_{Threshold}\ \ then\ \ \textbf{\textit{Stance Phase}}. \\ &if\ \ I < I_{Threshold}\ \ then\ \ \textbf{\textit{Swing Phase}}. \end{aligned} \qquad (5)$$

The waveform of the vertical force is nearly trapezoidal in shape but the slope of the waveform changes corresponding to the walking velocity. So, as a first step, the proposed new method is to approximate the waveform of the vertical force F_z as the isosceles trapezoidal wave which is generally present in the vertical force by adding the user's weight as a parameter to the time of the stance phase T_{sp} estimated from the current value of the DC motor. In this study, F_z is formulated as (6) (Fig. 6). We estimate the anteroposterior force from the current value of the DC motor by substituting (2) and (3) modeled in Section II, and (6) modeled in this section, into (4).

IV. Experiment

A. Objective

The objective is to verify the accuracy of the anteroposterior forces estimated by this proposed method from the current value of the DC motor.

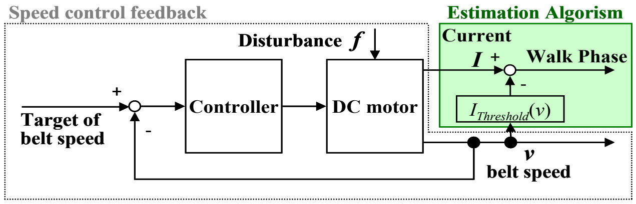


Fig. 5 Block diagram of walk phase estimation algorithm

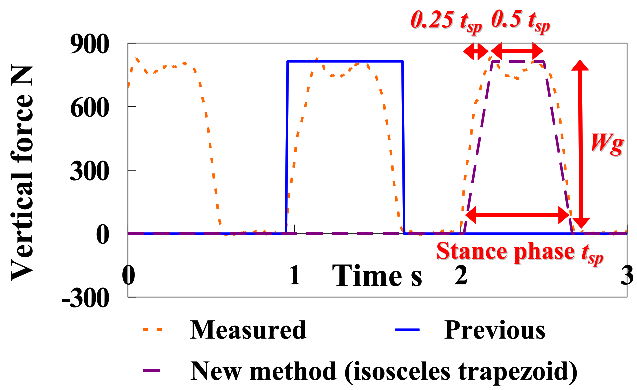


Fig. 6 Vertical forces in the new proposed method

B. Methodology

We compared the estimated anteroposterior force indicated by the new method that approximates the vertical force as trapezoidal waves with the force actually measured by the force plate.

We used a split belt treadmill, which is a treadmill with completely separated left and right treadmill belts (as shown in Fig. 7). We positioned the split belt treadmill at the center of the force plate (AMTI OR6-7-2000); this allowed us to measure the anteroposterior force while the subject walked on this treadmill. The anteroposterior force and the current value were measured while the subject walked on the treadmill at a constant velocity. The subject walked at 1.0, 1.5, 2.0 km/h of three conditions that decreased in steps of 0.5 from 2.0 km/h, which is the average walking velocity of elderly people. The measurement of the anteroposterior force and the current value of the DC motor were performed three times repeatedly at each belt velocity *v*. Each measurement was performed during ten walk cycles.

The experiment was performed with two healthy subjects whose physical characteristics were different. These are shown in Table I as subjects A and B. In contrast to the walking of the young subjects in this experiment, it is known that the extension angles of the hip joint, and the plantar and the dorsal flexion angles of the ankle, decrease in the walking of the elderly who are the target of this study.

Table I Physical Characteristics

Subject	A	B
Age, Sex	22, male	22, male
Height (cm)	166	175
Weight (kg)	61	83

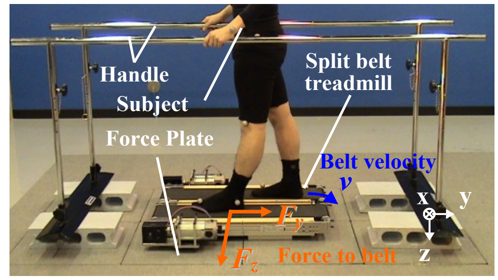


Fig. 7 Experimental set-up used for validation of the proposed algorithm

However, from the viewpoint of motion dynamics, the waveform pattern of the floor reaction force which is greatly related to verifying the effectiveness of this method during walking by elderly people is similar to that of the floor reaction force during walking by young people. In this paper, the experiment was performed with two young subjects as preliminary steps for our study. In the future, the experiment will be performed with elderly subjects. Before the experiment, informed consent was obtained from each subject, and each subject was allowed to grasp the treadmill handles to ensure safety. The experimental layout is shown in Fig. 7.

C. Results

We show the experimental result that compares the estimated anteroposterior force from the new method with that measured by the force plate at each belt velocity (1.0, 1.5, 2.0 km/h). Because the experimental results show similar trends for both the subjects A and B for each condition (velocity = 1.0, 1.5, 2.0 km/h), we show only the results for Subject A's walking at 2.0 km/h in Fig. 8. Fig. 8(a) shows the experimental result that compares the estimated anteroposterior force from the previous method with that measured by the force plate, and Fig. 8(b) shows the experimental result that compares the estimated anteroposterior force from the new proposed method with that measured by the force plate.

The experimental result in Fig. 8(a) shows that the error in the estimated anteroposterior force was large at the heel contact and the toe off. The maximum error was found to be 1.0×10^2 N and the RMS error was 20 N. However, the experimental result in Fig. 8(b) shows that the error in the estimated anteroposterior force was small at the heel contact and the toe off. The maximum error in this case was found to be 87 N and the RMS error was 16 N. So, the waveform of the estimated anteroposterior force from the new method is similar to that of the force measured by the force plate at each belt velocity in the results of both the subjects A and B.

The RMS error in estimated anteroposterior force at each belt velocity (1.0, 1.5, 2.0 km/h) for Subject A is shown in Fig. 8, and that for Subject B is shown in Fig. 9. It was found, in the results of both the subjects A and B, that the RMS error of estimated anteroposterior force from the new proposed method using the approximation of the vertical force as the trapezoidal wave was smaller than that of the estimated force from the previous method using the approximation of the vertical force as the square wave, at each belt velocity Additionally, it was found that the RMS error in the estimated anteroposterior force decreases, and the accuracy of the estimated anteroposterior force improves, as the subject's walking velocity decreases.

D. Discussion

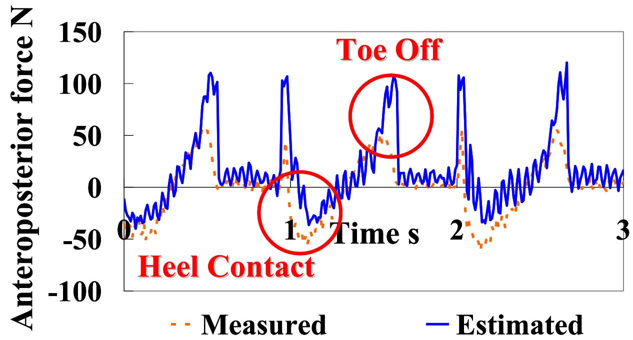
The waveform of the error in the estimated anteroposterior force, and those of the approximated vertical force and the force measured by the force plate are shown in Fig. 11. Fig. 11(a) shows the experimental result using the previous method, and Fig. 11(b) shows the experimental result using the new proposed method.

By comparing the approximated vertical force as the square wave in Fig. 11(a) with the approximated vertical force as the trapezoidal wave in Fig. 11(b), it was found that the waveform of the approximated vertical force as the trapezoidal wave in Fig. 11(b) nears that of the vertical force measured by the force plate. Because the error that occurred in the approximation of vertical force decreases, we considered that the error in the estimated anteroposterior force at the heel contact and the toe off decreases and the accuracy of the estimated anteroposterior force improves.

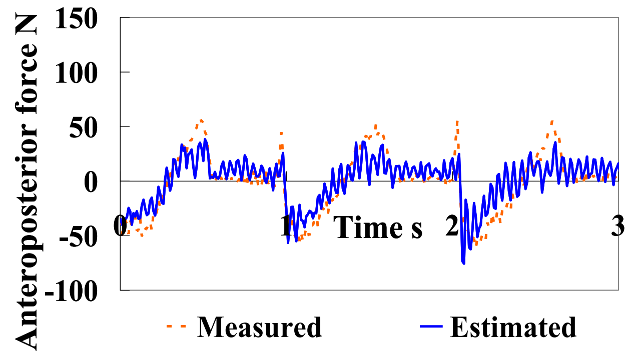
From the observed results in Fig. 9 and Fig. 10, it was found that the RMS error in the estimated anteroposterior force using the approximated vertical force by the square wave was smaller than that in the estimated anteroposterior force using the approximated vertical force as the trapezoidal wave, at each belt velocity (1.0, 1.5, 2.0 km/h), for both the subjects A and B. Thus, this experimental result showed that the proposed method is more accurate than the previous method in estimating the anteroposterior force.

However, it was also found that the accuracy of the estimated anteroposterior force does not improve as the subject's walking velocity increases. This is caused by change in the vertical force corresponding to change in the walking velocity. The waveform of the floor reaction force in the vertical direction varies with the walking velocity [9]. As the subject's walking velocity increases, this waveform approaches the square wave [9]. Conversely, as the subject's walking velocity decreases, this waveform approaches the trapezoidal wave [9]. So, compared with the experimental result at the walking velocity of 2.0 km/h, the vertical force is actually nearer the approximation of the square wave using the previous method than the approximation of the trapezoidal wave using the proposed method, in the experimental result at the walking velocity of 2.0 km/h.

Therefore, it was found that it is important to change the approximation of the vertical force corresponding to the walking velocity in order to accurately estimate the anteroposterior force. If the approximation of the vertical force changes corresponding to the walking velocity used, the error in the estimated anteroposterior force will be small, and this proposed method could possibly be useful for accurate estimation of the user's anteroposterior force on the treadmill. This proposed method might form important core technology as a first step of construction of a control algorithm that makes it possible for the user to always walk at



(a) Estimated anteroposterior forces in the previous method



(b) Estimated anteroposterior forces in the proposed method

Fig. 8 Measured and Estimated Anteroposterior forces

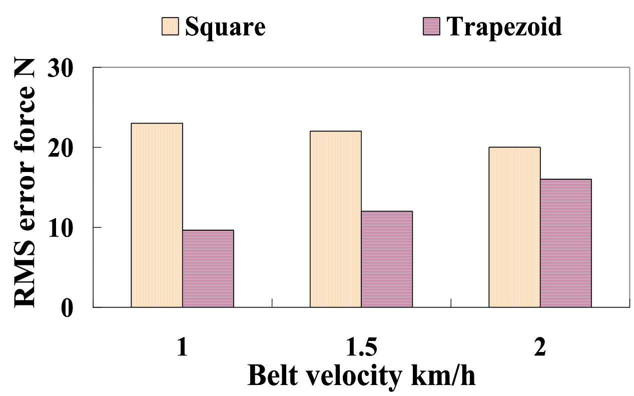


Fig. 9 Experimental result for Subject A

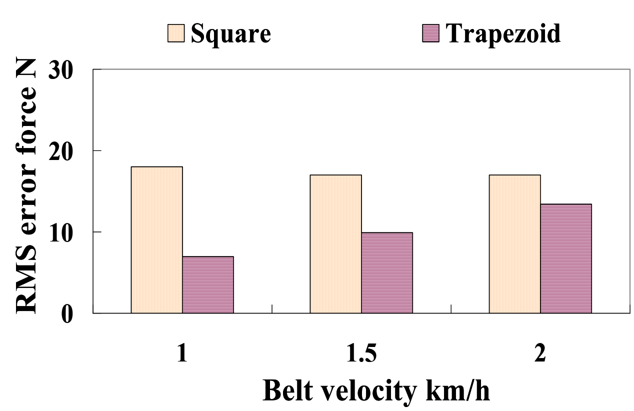
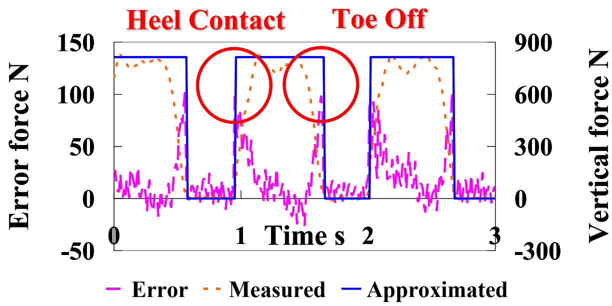
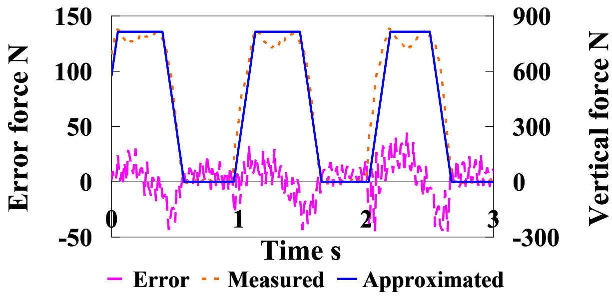


Fig. 10 Experimental result for Subject B



(a) Approximated vertical forces in the previous method



(b) Approximated vertical forces in the proposed method
Fig. 11 Error and the approximated vertical forces

his desired walking velocity on the treadmill both when starting and stopping walking. Thus, this is an application that makes the walking movement during the operation of TW-2 dramatically more comfortable. This technology makes the operability of TW-2 more intuitive than ever before.

V. Conclusion

We proposed a new method to approximate the waveform of the vertical force in order to estimate the anteroposterior force from the current value of the DC motor. To verify the accuracy of the anteroposterior force estimated with this proposed method, we compared the estimated anteroposterior force with the force measured by the force plate. From the results observed with two young subjects whose physical characteristics were different, the estimated waveform pattern was found to be similar to the measured one. Additionally, it was found in the experimental results of the subjects A and B that the RMS error in the estimated anteroposterior force decreased and the accuracy of the estimated anteroposterior force improved as the subjects' walking velocity decreased. However, it was also found that the accuracy of the estimated anteroposterior force did not improve as the subjects' walking velocity increased. This is attributed to change in the vertical force corresponding to change in the walking velocity. Therefore, it was found important to change the approximation of the vertical force corresponding to the walking velocity in order to accurately estimate the anteroposterior force. In the future, we will develop an algorithm to automatically approximate the change in force depending the gait cycle and, finally, improve the operability of TW- 2.

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