Estimation of Floor Reaction Force During Walking Using Frequency Analysis

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***Abstract* - This research group aims to develop a gait analysis system that can be easily used by individuals with little user burden. This research group has proposed a method to estimate the floor reaction force, which is an important parameter in analyzing walking, from the force balance equation using the acceleration measured by wearable sensors, thus obtaining its usefulness. Previously, we reduced the number of sensors from 15 for the whole body part to 5 selected for user burden, but the estimation accuracy decreased. Therefore, we consider how to reduce the number of sensors without sacrificing accuracy. In the proposed method, the acceleration of each part is discrete Fourier transformed and expressed in the frequency domain, a function representing the relationship of acceleration between each part is derived. This function is called the motion mode function, which is used to derive the acceleration in the frequency domain from the measured part to the unmeasured part. The inverse discrete Fourier transform was used to derive the acceleration in the time domain. As a result, the acceleration was estimated with high accuracy. In addition, in order to make the system more versatile, the acceleration of the unmeasured part is estimated using the average motion mode function, which is the average of the motion mode functions obtained from the acceleration of multiple trials by multiple persons and averaged for each order. The estimated acceleration was then used to estimate the composite floor reaction force in two directions. Correlation coefficients were used to check the accuracy of the estimates and demonstrate the usefulness of the proposed method.**

Keywords- Discrete Fourier transform; Gait analysis; Average motion mode function; Floor reaction force.

I. INTRODUCTION

Walking is one of the most familiar activities performed by most people. Gait analysis data are important information in clinical medicine and sports for evaluating the effectiveness of rehabilitation therapy and for teaching athletes to improve their normal function. Gait analysis is performed using the widely known optical motion capture (MC) and installed force plate [2]. These systems are capable of deriving floor reaction forces and joint moments, which allow for detailed and accurate analysis of walking [3]. However, these systems are very expensive and have a limited measurement range, so they are limited to use in facilities such as research institutes and hospitals. In addition, electromyography can be used to determine muscle tension, but surface electromyography can only measure a limited number of muscles, while needle electrodes can only be used in specialized institutions. One method for estimating exertional capacity during walking is to perform simulation analysis using a musculoskeletal model [4]. Joint moment and muscle tension can be calculated as the load on the intervertebral discs from the balance of forces, but it involves many assumptions and constraints due to the setting of unclear individual parameters. To address these problems, gait analysis systems using wearable sensors have been proposed. Lei Wang et al. [5] attached four wearable sensors to the lower limbs to estimate step length and left-right asymmetry. Shaghayegh Zihajehzadeh et al. [6] used a wearable sensor attached to the wrist to estimate the walking speed in the walking direction. However, these systems only estimate the most common walking parameters for gait analysis. Also, our research group has developed a wearable motion capture system that combines a sandal-type force plate, which is a wearable version of an installed force plate and embedded with a 3-axis force sensor with motion sensors in six parts throughout the body [7]. However, this system was still expensive, burdensome to the user, causing problems with walking. Moreover, to develop an inexpensive and user-friendly gait analysis system that can be used easily by individuals, this research group proposed a method that can easily estimate the left and right composite floor reaction forces (ground reaction forces), which are important parameters for analyzing walking, using a wearable inertial sensor. This method has no limitations on the measurement range and is inexpensive to use [8]. This method uses the balanced relationship between three kinds of forces acting on a human while walking: inertia, gravity, and floor reaction force. The inertial force of the whole body was derived from the acceleration of the 15 parts and the mass of each part, based on the idea that the whole body can be treated as 15 rigid bodies since the human body can be composed of 14 rigid bodies and the trunk can be divided into the upper and lower trunk at the lower end of the ribs, as described by Ae et al. [9]. The estimated floor reaction force derived from the proposed method showed waveforms comparable to those measured by the force plates. Therefore, to accurately estimate the floor reaction force, each inertial force can be derived from the acceleration measured from 15 parts of the body. In the previous paper [10], we thought that by reducing the number of sensors used, it would be possible to realize a system with less user burden. we tried to derive the floor reaction force by substituting the acceleration of the unmeasured part with the acceleration of the other measured part, but the estimation accuracy decreased with the reduction of the number of sensors. Therefore, we consider how to reduce the number of sensors without sacrificing accuracy. As one method, we can reduce the number of sensors and the resulting loss of accuracy by selecting measured parts with a limited number of sensors and estimating the acceleration of unmeasured parts from the measured parts. In order to estimate with less accuracy degradation, the relationship between the motion of the measured and unmeasured parts must be understood. However, the quantitative description of the motion relationship in terms of time waveforms is not easy. Because of this, we considered the frequency domain because the waveform can be broken down into its components and analyzed quantitatively [11]. By considering walking as a periodic motion and expressing the acceleration in the frequency domain through a discrete Fourier transform of the acceleration of each part, the characteristics of each component were extracted when the acceleration was decomposed into its constituent components. From the results, the characteristics of the motion of each part were clarified. For the left and right limbs such as the thigh and upper arm, we used the average of the left and right accelerations to align the walking frequencies and make it easier to grasp the characteristics. Based on the results of the discrete Fourier transform, a function was derived by dividing the unmeasured part by the measured part for each order component in order to quantitatively express the relationship between the two parts. This function is called the motion mode function in this study, which quantitatively shows the relationship of acceleration between two parts of the target. By using the obtained motion mode function, the acceleration of the unmeasured part could be estimated from the measured part in the frequency domain. The acceleration in the frequency domain can be derived by multiplying the motion mode function by the acceleration in the frequency domain of the measured part. The acceleration data in the time domain can be obtained by inverse discrete Fourier transforming the acceleration expressed in the frequency domain. In the previous paper [1], we compared the estimated acceleration of the unmeasured part with the measured value of MC, using the correlation coefficient as a measure of the estimation accuracy to show the usefulness of the method in two directions.

In the aforementioned previous report, the motion mode function, which represents the relationship between the accelerations of two parts, was created for each subject. In the method using the motion mode function of one subject, if the walking does not change significantly from trial to trial, the deviation of the motion mode function becomes small, allowing the acceleration of the unmeasured part to be estimated accurately. In this paper, we propose an average motion mode function to enhance the versatility of the system. The average motion mode function is a function that is derived from the motion mode function for each trial of multiple subjects and averaged for each order. The purpose of this paper is to accurately estimate the composite floor reaction force in two directions with a minimum number of measured parts by using this average motion mode function. First, the accelerations at each site for multiple trials of multiple subjects are obtained from the MC to derive the average motion mode function. These are discrete Fourier transformed to extract the features that are important for acceleration estimation. The average motion mode function is derived using the result of the discrete Fourier transform. The next step is to estimate the acceleration of the unmeasured part using the average motion mode function. To estimate the acceleration of the unmeasured part, the acceleration of the unmeasured part in the frequency domain is obtained by multiplying the average motion mode function by the acceleration of the measured part in the frequency domain of the trial that was not used in deriving the average motion mode function. The acceleration in the time domain is estimated by inverse discrete Fourier transforming the acceleration of the unmeasured part in the frequency domain. The accuracy of acceleration estimation for unmeasured parts depends largely on where the measured part is selected. Therefore, the selection of the measured part is discussed. There are two methods for selecting the optimal part. The first method is to derive the correlation coefficient between the actual measured value of the unmeasured part and the value estimated using the average motion mode function to determine the optimal part based on the estimation accuracy. The second selection criterion is the ability to accurately estimate the acceleration of a large mass part, which is important for accurately estimating the floor reaction force. Taking these two points into consideration, we select the optimal site for the acceleration estimation method of the unmeasured part. Finally, the composite floor reaction force in two directions is derived using the acceleration of the unmeasured part estimated using the average motion mode function. Correlation coefficients are again used to examine the accuracy of acceleration and floor reaction force estimates.

This research group aims to develop a system using wearable sensors that can be easily used in daily life. However, compared to MC, current wearable inertial sensors are susceptible to noise and it is difficult to measure motion accurately. Therefore, in this paper, the acceleration data used in the analysis will be obtained using MC to examine the validity of the methodology. In this paper, the vertical direction and the walking direction are focused on because the floor reaction force in the left-right direction is small. The floor reaction force is derived as a composite of the floor reaction forces applied to the left and right feet, because it is derived from the balance between the inertia of the whole body and the sum of gravity.

II. METHOD

A. *Method for obtaining acceleration data used in Fourier analysis*

The acceleration data used in the analysis is for two steps of one walking cycle, which is a steady-state walk from the fifth and sixth steps after the beginning of the walk, and the walk is accompanied by a metronome in order to unify the steps. Subjects practice walking to the rhythm of a metronome sufficiently before taking measurements.

The recursive marker is attached at the center of gravity of each segment of the whole body, which is divided into 15 parts as shown in Figure 1.

Figure 1. Attaching position of recursive marker for MC

The division of the whole body is based on the idea that the human body can be composed of 14 rigid bodies and that the trunk can be divided into two parts at the lower end of the ribs, the upper trunk and the lower trunk, so that the whole body can be treated as 15 rigid bodies, as described by Ae et al. [9]. In this study, we use the average of the left and right accelerations for the left and right limbs, such as the thigh and the upper arm, referring to them as the average thigh and the average upper arm, respectively. As a result, 9 parts will be considered instead of 15 parts in the whole body. The acceleration data used in the analysis need to have equal values at both ends to improve the accuracy of the Discrete Fourier Transform results. This time, we will measure 15 trials for each subject and obtain acceleration data for each part.

B. *Estimation of acceleration at the unmeasured part*

The method of estimating the acceleration of an unmeasured part using the acceleration information of a measured part is described.

First, by performing a discrete Fourier transform of the acceleration data at each part and obtaining the magnitude and phase for each order, the magnitude of the acceleration at each part can be quantitatively evaluated for each order, and the dominant frequency and its band can be determined when estimating the acceleration. In this paper, we focus on the magnitude at which the motion characteristics of each part become prominent in the discrete Fourier transform results of each part. Next, the motion mode function used to estimate the acceleration of the unmeasured part from the measured part is obtained. The motion mode function can be derived by dividing the unmeasured part by the measured part for each order using the result of the discrete Fourier transform of each part. For example, when the upper trunk is the measured part and the foot is the unmeasured part, by dividing the foot by the upper trunk, the motion mode function that can estimate the acceleration of the foot from the upper trunk can be derived. In the previous report, we acquired walking data for 15 trials for each subject and used the motion mode function that averaged the motion mode functions of the data for 14 trials. The reason why one trial is omitted is that it is to be used for verification to evaluate the usefulness of the motion mode function. In this paper, we derive the motion mode function for each trial of multiple subjects and average them for each order to derive the average motion mode function. The number of motion mode functions is 8 for the remaining parts for each measured part and 9 for the possible measured parts, so a total of 72 functions per direction are derived for each subject. The average motion mode function, which is the average of the motion mode functions for each of these parts for each order, is derived as a function that reflects the acceleration relationship between the parts for all subjects. The acceleration of the unmeasured part is estimated in the frequency domain by multiplying the obtained average motion mode function by the result of the discrete Fourier transform of the measured part for the remaining one trial of the subject. For example, when the upper trunk is the measured part and the foot is the unmeasured part, the acceleration in the frequency domain of the foot can be derived by multiplying the average motion mode function of the upper trunk and foot by the acceleration in the frequency domain of the upper trunk. Finally, the acceleration in the time domain is obtained by inverse discrete Fourier transforming the acceleration of the unmeasured part in the frequency domain derived using the average motion mode function.

In this study, we used the average of the left and right acceleration values for the left and right limbs, such as the thighs and upper arms, so that the walking frequency of each part of the body can be aligned. Therefore, it is easier to understand the dominant frequencies and their bands when estimating the acceleration of unmeasured parts as well as the relationship between the head, trunk, and limb movements.

III. EXPERIMENTAL DESIGN

In the experiment, MC with 12 cameras (manufactured by Motion Analysis Co., Ltd., MAC 3D System), force plate 3 units (manufactured by Tec Gihan Co., Ltd., TF-6090-C 1 unit, TF-4060-D 2 units), and a metronome were used. The force plates are a strain gauge type transducer that can measure force, moment, pressure center, etc. The MC and the force plate are synchronized to accurately measure the ground contact timing in walking.

Five healthy subjects (male: age 21±2, height 1.70±0.10 [m], weight 65±5 [kg]) were explained the purpose and contents of this study and gave oral and written consent. In addition, this study was approved by the Ethical Review Committee.

10 steps are measured from the beginning of walking, and the force plates are placed at the 5th to 7th step from the beginning of walking. To measure the ground contact timing required by the discrete Fourier transform, three force plates are set up so that one foot contacts one force plate. The sampling frequency is set to 100 Hz, and low-pass processing with a cutoff frequency of 9 Hz is used for smoothing.

IV. RESULT&DISCUSSION

A. Estimate the acceleration of the unmeasured part from the measured part

In this section, we showed whether the acceleration of the non-measured part can be estimated accurately from the measured part using the average motion mode function derived from the five subjects who participated in this experiment.

The number of motion mode functions is 8 for the remaining parts for each measured part and 9 for the possible measured parts, so a total of 72 functions per direction are derived for each subject. However, due to the limitation of space, the acceleration estimation results for the case where the measured part is the upper trunk and the unmeasured parts are the average thigh and the average upper arm are shown as an example. In this paper, the subjects walked to a metronome at a pace of 100 BPM (one walking cycle is 0.6 seconds), and the acceleration of each part was taken out for two steps (1.2 seconds).

First, the measured accelerations in the time domain of the upper trunk, average thigh, and average upper arm of one subject obtained from the experiment were shown. Figure 2 shows the acceleration of the upper trunk, the average thigh, and the average upper arm in the vertical direction, and Figure 3 shows the acceleration of the upper trunk, the average thigh, and the average upper arm in the walking direction. Figure 2 shows that in the vertical direction, the upper trunk and average upper arm move similarly, but the average thigh moves with more high-frequency waves than the upper trunk and average upper arm. Figure 3 shows that even in the walking direction, the average thigh moves with more high-frequency waves than the upper trunk and average upper arm. The acceleration waveforms in the time domain shown in Figure 2 and Figure 3 indicated that it is difficult to quantitatively show the relationship of motion between each part of the body in both the vertical and walking directions and to estimate the acceleration of the average thigh and the average upper arm from the upper trunk simply by making corrections such as the constant correction. In addition, we found that it was difficult to estimate the acceleration of other parts of the body from one part, not only the upper trunk.

Next, the acceleration data is discrete Fourier transformed and decomposed into frequency components, and the magnitude and phase are determined for each direction. From the results, the dominant frequencies and their bands in estimating the acceleration of the unmeasured part were identified. The important components are the walking frequency (The walking frequency is 1.667 Hz, and the order of the graph is an integer.) and it's integer multiples while 0.5 times the walking frequency (The frequency of the 0. 5 times component is 0.833 Hz, and the order of the graph is the value of the half component of the integer.) and its integer multiples are not focused on in this report because they are indicators of the degree of left-right balance. Figure 4 shows the magnitude and phase in the vertical direction of the upper trunk, Figure 5 shows the average thigh, and Figure 6 shows the average upper arm. Figure 7 shows the magnitude and phase in the walking direction of the upper trunk, Figure 8 shows the average thigh, and Figure 9 shows the average upper arm. The magnitude and phase were shown as the average and standard deviation calculated from the walking data of five subjects of 15 trials each (for a total of 75 trials). The result of the discrete Fourier transform has a magnitude in the components up to 9 Hz because of the low-pass processing with a cutoff frequency of 9 Hz.

From Figure 4 to Figure 9, the amplitudes of the walking frequency and its integer multiple components (The order of the graph is an integer value) were large in both the vertical and traveling directions. Since the walking was controlled to some extent, except for the higher-order components of the average thigh, it can be seen that the movements of each part were similar from trial to trial and the standard deviation was small. In the vertical direction, the motion of the upper trunk was highly dependent on the walking frequency component, while the average thigh and the average upper arm contain not only the walking frequency component but also many higher-order components. For the average thigh shown in Figure 5, the reason for the large standard deviation in the third order or higher of the walking frequency component was that the walking of one subject differed greatly from trial to trial. In the direction of travel, the upper trunk and average upper arm were highly dependent on the walking frequency component, while the average thigh not only has a relatively large walking frequency component but also moves with many higher-order components. Therefore, the walking frequency and its integer multiple components are important for the acceleration estimation of the unmeasured part, and higher-order components are also necessary.

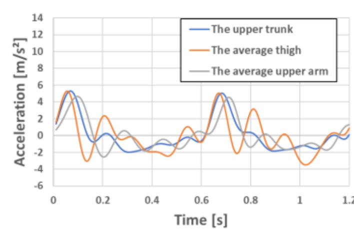

Figure 2. Acceleration data of upper trunk, average thigh, and average upper arm in the vertical direction. The blue line is the upper trunk acceleration, the orange line is the average thigh acceleration, and the gray line is the average upper arm acceleration. Acceleration data were measured by MC.

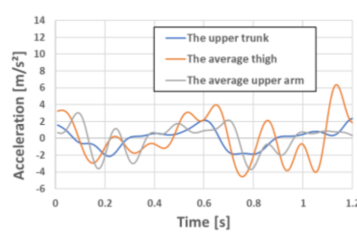

Figure 3. Acceleration data of upper trunk, average thigh, and average upper arm in the walking direction. The blue line is the upper trunk acceleration, the orange line is the average thigh acceleration, and the gray line is the average upper arm acceleration. Acceleration data was measured by MC.

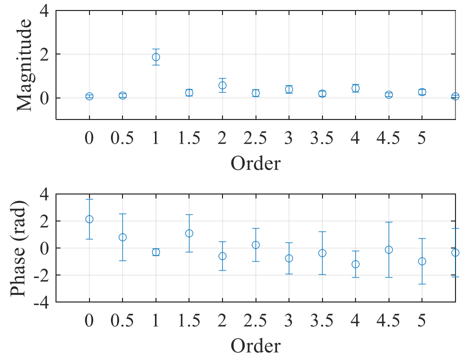

Figure 4. Result of the discrete Fourier transform of the upper trunk acceleration in the vertical direction. The magnitude and phase are shown as the average and standard deviation calculated from 75 walking data(5 subjects, 15 trials each). The upper trunk acceleration in the vertical direction is highly dependent on the walking frequency component.

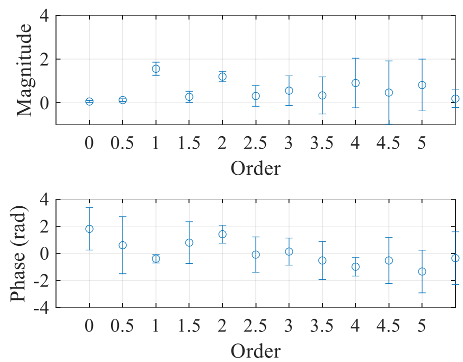

Figure 5. Result of the discrete Fourier transform of the average thigh acceleration in the vertical direction. The magnitude and phase are shown as the average and standard deviation calculated from 75 walking data(5 subjects, 15 trials each). The average thigh acceleration in the vertical direction contains not only the walking frequency component but also many higher-order components.

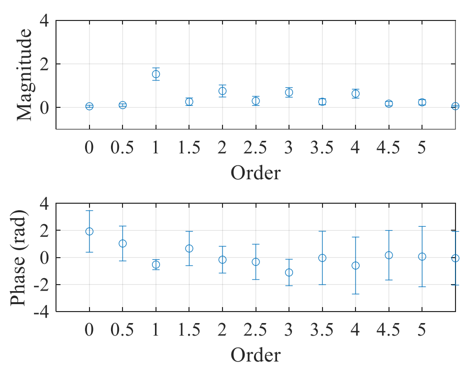

Figure 6. Result of the discrete Fourier transform of the average upper arm acceleration in the vertical direction. The magnitude and phase are shown as the average and standard deviation calculated from 75 walking data(5 subjects, 15 trials each). The average upper arm acceleration in the vertical direction contains not only the walking frequency component but also many higher-order components.

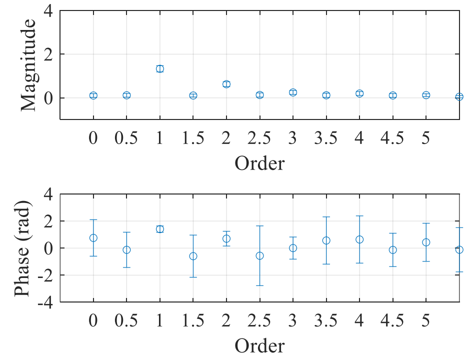

Figure 7. Result of the discrete Fourier transform of the upper trunk acceleration in the walking direction. The magnitude and phase are shown as the average and standard deviation calculated from 75 walking data(5 subjects, 15 trials each). The upper trunk acceleration in the walkingl direction is highly dependent on the walking frequency component.

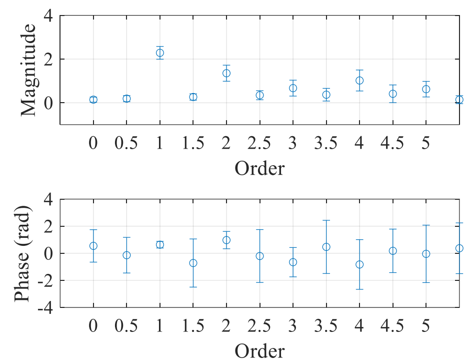

Figure 8. Result of the discrete Fourier transform of the average thigh acceleration in the walking direction. The magnitude and phase are shown as the average and standard deviation calculated from 75 walking data(5 subjects, 15 trials each). The average thigh acceleration in the walking direction contains not only the walking frequency component but also many higher-order components.

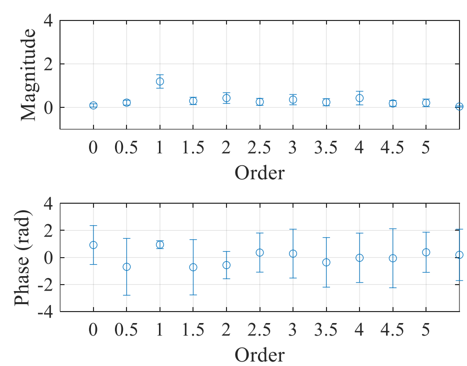

Figure 9. Result of the discrete Fourier transform of the average upper arm acceleration in the walking direction. The magnitude and phase are shown as the average and standard deviation calculated from 75 walking data(5 subjects, 15 trials each). The average upper arm acceleration in the walking direction is highly dependent on the walking frequency component.

Next, the average motion mode function is shown. As an example, the average motion mode function was derived with the upper trunk as the measured part and the average thigh and upper arm as the unmeasured parts. A motion mode function was derived for each of the 14 trials for each subject, and the average motion mode function was averaged for each order of the motion mode function for each trial for five subjects (Average of motion mode functions for 70 trials). Figure 10 shows the results for the average thigh with the upper trunk as the measured part, and Figure 11 shows the results for the average upper arm with the upper trunk as the measured part in the vertical direction. Figure 12 shows the results for the average thigh with the upper trunk as the measured part, and Figure 13 shows the results for the average upper arm with the upper trunk as the measured part in the walking direction. The gain and phase difference of the results were shown as the average and standard deviation calculated from the walking data of 70 trials.

In this paper, we focused on the walking frequency and its integer multiple components, which are important for estimating the acceleration of unmeasured parts. In the vertical direction, Figure 10 and Figure 11 show that the most important walking frequency component has a gain of near 1 and a phase difference near 0, indicating that the motion is similar in the vertical direction. For the integer multiple components larger than the quadratic component, the gain was larger than 1, and the average thigh and the average upper arm moved more than the upper trunk. In the walking direction, Figure 12 shows that the average thigh moves more than the upper trunk because the gain of the walking frequency component is near 2. Figure 13 shows that the upper trunk and average upper arm move similarly because the gain of the most important walking frequency component is near 1 and the phase difference is near 0. For the integer multiple components larger than the quadratic component, the gain was larger than 1 except for the quadratic components of the upper trunk and average upper arm, indicating that the average thigh and average upper arm moved more than the upper trunk. As for the quadratic components of the upper trunk and the average upper arm, Figure 9 shows that the gain of the average upper arm is smaller than 1 due to its smaller magnitude in the integer multiple components of the walking frequency component.

From Figure 10 to Figure 13, we consider that the reason for the large standard deviation in the integer components of the walking frequency is that the magnitude of each part is small in the integer components above the second-order component of the upper trunk. The large standard deviation in the integer multiple components of the walking frequency may affect the error between the measured and estimated values in high-frequency waves when the acceleration at the unmeasured part is estimated.

Next, the acceleration in the time domain of the unmeasured part is shown. As an example, the average thigh and upper arm accelerations were estimated from the upper trunk acceleration of trial 15 for one subject using the average motion mode function obtained from 70 trials of walking data shown in Figure 10 to Figure 13. The acceleration of the average thigh and the average upper arm was estimated in the frequency domain using a total of five points of the walking frequency and its integer multiple components of the average kinematic mode function and an inverse discrete Fourier transform is performed to obtain the acceleration data in the time domain.

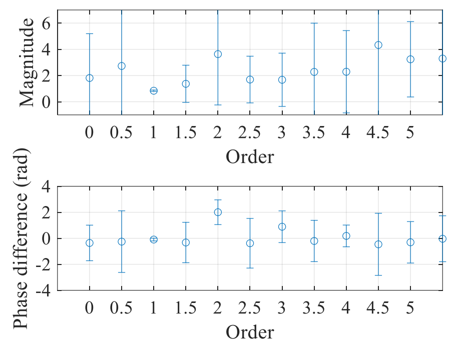

Figure 10. The average motion mode function in the vertical direction with the input as upper trunk and the output as average thigh. The magnitude and phase difference are shown as the average and standard deviation calculated from 70 walking data(5 subjects, 14 trials each).

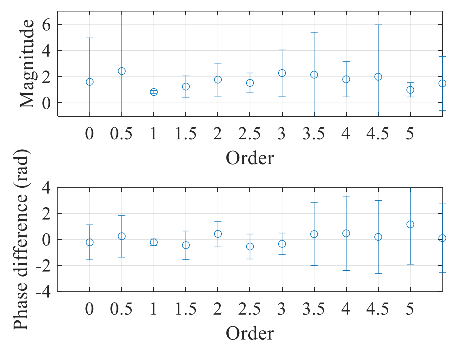

Figure 11. The average motion mode function in the vertical direction with the input as upper trunk and the output as average upper arm. The magnitude and phase difference are shown as the average and standard deviation calculated from 70 walking data(5 subjects, 14 trials each).

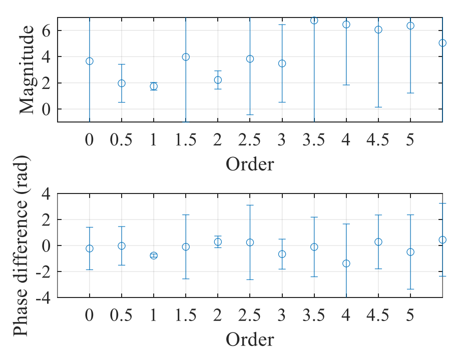

Figure 12. The average motion mode function in the walking direction with the input as upper trunk and the output as average thigh. The magnitude and phase difference are shown as the average and standard deviation calculated from 70 walking data(5 subjects, 14 trials each).

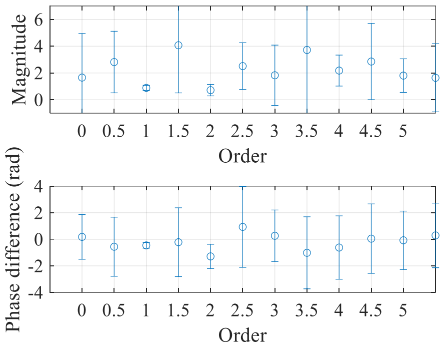

Figure 13. The average motion mode function in the walking direction with the input as upper trunk and the output as average upper arm. The magnitude and phase difference are shown as the average and standard deviation calculated from 70 walking data(5 subjects, 14 trials each).

The results of the comparison between the measured values of MC, the acceleration estimated using the average motion mode function, and the acceleration estimated from the motion mode function derived from 14 trials of one subject were shown in Figure 14 and Figure 15 for the vertical direction of the average thigh and the average upper arm, respectively, and in Figure 16 and Figure 17 for the walking direction of the average thigh and the average upper arm, respectively. In the method using the average motion mode function, the acceleration of the average thigh and the average upper arm can be estimated, using the measured part as the upper trunk, similar to the method using the motion mode function of one subject. The respective correlation coefficients between the measured values of MC and the acceleration estimated using the average motion mode function, and between the measured values of MC and the acceleration estimated from the motion mode function of one subject are shown in Table 1. Compare the correlation coefficients when using the average motion mode function and when using the motion mode function of one subject. In the vertical direction and the walking direction, the motion mode function of one subject is more accurately estimated in both cases where the unmeasured part is the average thigh and the average upper arm. This is because the deviation of the motion mode function becomes small in the method using the motion mode function of one subject, if the walking does not change significantly from trial to trial, and the acceleration can be estimated accurately. However, in the vertical direction, the correlation coefficients were high even when the average operating mode function was used, and accuracy was ensured. Figure 16 and Figure 17 show that in the walking direction, the average thigh captures the characteristics of the waveform, while the average upper arm cannot reproduce the higher-order components, but can estimate the phase and amplitude of the lower-order components.

The results on the accuracy of the acceleration estimation of the unmeasured part when the measured part is the other part are described in the next section.

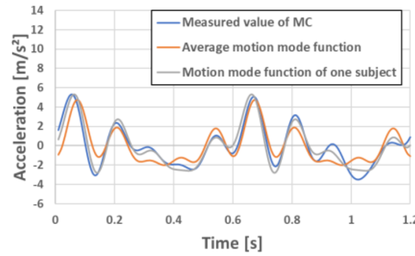

Figure 14. Comparison of acceleration estimated using measured value of MC and the average motion mode function, and acceleration estimated from the motion mode function derived from 14 trials of one subject, using the upper trunk as the measured part and the average thigh as the unmeasured part in the vertical direction.

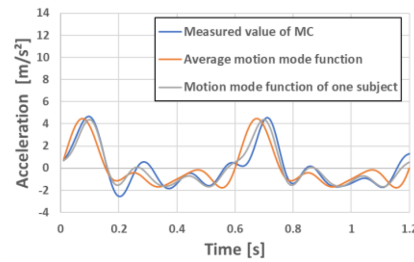

Figure 15. Comparison of acceleration estimated using measured value of MC and the average motion mode function, and acceleration estimated from the motion mode function derived from 14 trials of one subject, using the upper trunk as the measured part and the average upper arm as the unmeasured part in the vertical direction.

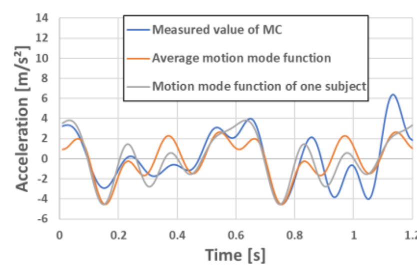

Figure 16. Comparison of acceleration estimated using measured value of MC and the average motion mode function, and acceleration estimated from the motion mode function derived from 14 trials of one subject, using the upper trunk as the measurement part and the average thigh as the unmeasured part in the walking direction.

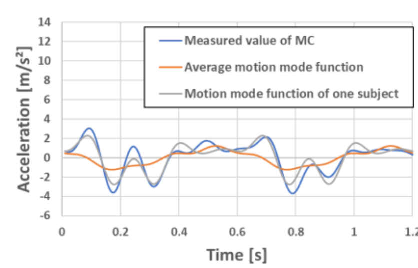

Figure 17. Comparison of acceleration estimated using measured value of MC and the average motion mode function, and acceleration estimated from the motion mode function derived from 14 trials of one subject, using the upper trunk as the measurement part and the average upper arm as the unmeasured part in the walking direction.

TABLE I. CORRELATION COEFFICIENTS OF THE ESTIMATED ACCELERATION OF THE UNMEASURED PART WITH MEASURED VALUES OF MC. CORRELATION COEFFICIENTS WITH MEASURED VALUE OF MC ARE SHOWN FOR THE CASE WHERE THE AVERAGE MOTION MODE FUNCTION OF FIVE PEOPLE IS USED AND THE CASE WHERE THE MOTION MODE FUNCTION OF ONE SUBJECT IS USED.

Vertical direction		
Function	Average thigh	Average upper arm
Average motion mode function	0.825	0.816
Motion mode function of one subject	0.962	0.959
Walking direction		
Function	Average thigh	Average upper arm
Average motion mode function	0.693	0.642
Motion mode function of one subject	0.848	0.926

B. Selection of the optimal part for estimation of acceleration

In this section, we consider which part of the body can be selected as the measured part to estimate the acceleration of the unmeasured part with good accuracy.

In the past, the research group has considered that accurate estimation of the acceleration at the part with large mass is important for accurate estimation of the floor reaction force because the higher the accuracy of the part with large mass, the higher the accuracy of the floor reaction force. Considering these factors, the optimal site was selected. First, the correlation coefficient between the measured values of MC and the values estimated from each measured part was derived. Next, the optimal part was selected from the derived correlation coefficients, focusing on the measured part that can accurately estimate the acceleration of the part with a large mass.

In this paper, the whole body was divided into three groups as a large block (A: head, upper trunk, lower trunk B: upper limbs C: lower limbs) and the part with the largest mass in each group is used as the measured part to estimate the acceleration of the unmeasured part. The mass of each body part was derived by multiplying the total body mass measured with a scale by the body part coefficients of Japanese athletes shown in Table 2, which were calculated by Ae et al. using the method of Yokoi et al. [12], who applied the model of Jensen et al. [13]. Body part coefficients for upper and lower limbs were shown as composite values for left and right.

The measurement sites were the upper trunk, the average upper arm, and the average thigh from each of the three groups, and the non-measurement sites were the upper trunk, the average thigh, the lower trunk, the average lower leg, and the head, which are the sites with large mass, respectively.

As an example, when the unmeasured part is the lower trunk, the estimated results from each measured part are shown in Figure 18 for the vertical direction and Figure 19 for the walking direction. In this paper, the average values of the correlation coefficients for each measured part were obtained for each of the two subjects, and are shown in Table 3 for the vertical direction and Table 4 for the walking direction.

TABLE II. BODY PART COEFFICIENTS OF JAPANESE ATHLETES(UPPER AND LOWER LIMBS ARE COMPOSITE VALUES OF LEFT AND RIGHT)

Group	Body part	Mass ration [%]
A	Head	6.9
	Upper tunk	30.2
	Lower trunk	18.7
B	Upper arm	5.4
	Forearm	3.2
	Hand	1.2
C	Thigh	22.0
	Lower thigh	10.2
	Foot	2.2

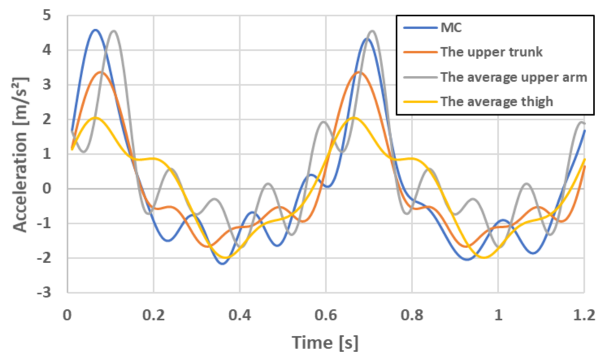

Figure 18. In the vertical direction, the unmeasured part is the lower trunk, and the measured value of MC are compared with the acceleration estimated with the upper trunk as the measured part, the acceleration estimated with the average upper arm as the measured part, and the acceleration estimated with the average thigh as the measured part.

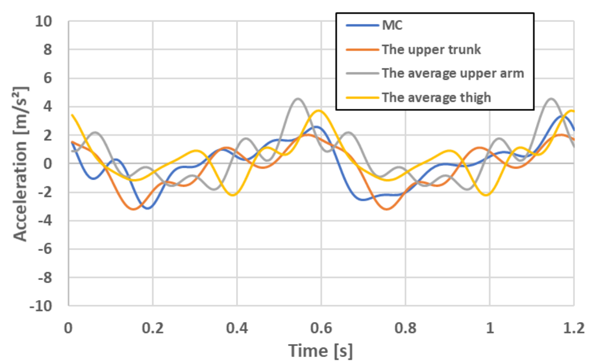

Figure 19. In the walking direction, the unmeasured part is the lower trunk, and the measured value of MC are compared with the acceleration estimated with the upper trunk as the measured part, the acceleration estimated with the average upper arm as the measured part, and the acceleration estimated with the average thigh as the measured part.

In Figure 18 and Figure 19, the characteristics of the measured waveforms in both the vertical and walking directions can be captured in terms of their magnitude and phase. To evaluate the results, the correlation coefficient was again used to quantitatively show the estimation accuracy. Table 3 shows that in the vertical direction, the average values of the correlation coefficients were the highest when the measured part was the upper trunk for subject A and the average thigh for subject B, and the values were comparable. Considering that the higher accuracy of the estimation of the large mass part is more accurate when estimating the floor reaction force, we decided to select the upper trunk as the measured part, which shows a relatively high correlation coefficient when the unmeasured parts are the upper trunk, average thigh, and lower trunk in the vertical direction.

Table 4 shows that in the walking direction, the average values of the correlation coefficients were the highest when the measured part was the average thigh for subject A and the upper trunk for subject B, and the values were comparable. In the walking direction, the correlation coefficients were similar when the measured parts were the upper trunk, average thigh, and average upper arm. However, for the upper and lower limbs, the average acceleration of the left and right limbs is used, requiring measurements at two parts each. Therefore, the upper trunk was selected as the measured part in order to reduce the number of sensors and to improve the estimation accuracy.

TABLE III. CORRELATION COEFFICIENT BETWEEN THE MEASURED VALUES OF MC AND THE ESTIMATED VALUES FROM EACH MEASURED PART IN THE VERTICAL DIRECTION. IN CASE THE MEASURED AND UNMEASURED PARTS ARE THE SAME, THE CORRELATION COEFFICIENT IS EXPRESSED AS 1.

Measured part \ Unmeasured part	Uppert trunk	Average thigh	Lower trunk	Average lower thigh	Head	Average
Subject A						
Upper trunk	1	0.825	0.929	0.843	0.927	0.905
Average thigh	0.853	1	0.824	0.901	0.846	0.885
Average upper arm	0.900	0.668	0.828	0.545	0.949	0.778
Subject B						
Upper trunk	1	0.661	0.812	0.723	0.867	0.813
Average thigh	0.954	1	0.792	0.651	0.833	0.846
Average upper arm	0.919	0.627	0.880	0.498	0.961	0.777

TABLE IV. CORRELATION COEFFICIENT BETWEEN THE MEASURED VALUES OF MC AND THE ESTIMATED VALUES FROM EACH MEASURED PART IN THE WALKING DIRECTION. IN CASE THE MEASURED AND UNMEASURED PARTS ARE THE SAME, THE CORRELATION COEFFICIENT IS EXPRESSED AS 1.

Measured part \ Unmeasured part	Uppert trunk	Average thigh	Lower trunk	Average lower thigh	Head	Average
Subject A						
Upper trunk	1	0.693	0.781	0.260	0.170	0.581
Average thigh	0.676	1	0.600	0.214	0.552	0.608
Average upper arm	0.721	0.648	0.538	0.173	0.308	0.478
Subject B						
Upper trunk	1	0.736	0.942	0.376	0.188	0.648
Average thigh	0.740	1	0.655	0.430	0.250	0.615
Average upper arm	0.733	0.780	0.707	0.0650	0.100	0.477

V. APPLICATION TO ESTIMATION OF FLOOR REACTION FORCE

A. *Floor reaction force estimation method using acceleration of each part*

We use the fact that the sum of the inertia force derived from the acceleration measured from 15 parts of the body and the mass of each part, and the gravity force is balanced by the floor reaction force [8]. Equation (1) is an equation for estimating floor reaction forces in the vertical direction. Where F_z is the vertical floor reaction force, m is the mass of each body part, a_z is the vertical acceleration of each body part, i is the number of body parts, M is the total body mass, and g is the acceleration of gravity.

$$F_z = \sum_i m_i a_{zi} + Mg. \tag{1}$$

Equation (2) is an equation for estimating the floor reaction force in the walking direction. The floor reaction force in the walking direction can be derived by removing the gravity term from (1) and using the acceleration in the walking direction for the acceleration at each site. Where F_y is the floor reaction force in the walking direction and a_y is the acceleration in the walking direction.

$$F_y = \sum_i m_i a_{yi}. \tag{2}$$

In this paper, the number of parts i is nine because the average of the acceleration of the left and right limbs, such as the thigh and upper arm, is used. The mass of each body part is derived by multiplying the total body mass measured by a scale by the body part coefficients in Table 2.

B. *Estimation Result*

The floor reaction force was derived using the acceleration derived from the acceleration estimation method for the unmeasured part shown in Section IV.A. The floor reaction force in two directions was derived by estimating the acceleration of the remaining eight parts, using the trunk as the measured part, which was selected as the optimal part in the acceleration estimation method for unmeasured parts described in Section IV.B. The correlation coefficient was used to judge the accuracy of the estimation.

The results of the comparison of the floor reaction force derived from the force plate measurements and the average motion mode function, the floor reaction force derived from the motion mode function of one subject, and the floor reaction force derived using nine measured parts were shown in Figure 20 for the vertical direction and Figure 21 for the walking direction. The measured values from the force plates were smoothed by low-pass processing with a cutoff frequency of 9 Hz as well as the acceleration obtained from the sensor. From Figure 20 and Figure 21, one cycle of walking was shown, but one foot was grounded outside of the force plate until time 0.13[s], so the measurement of the force plates indicated by the solid blue line did not measure the composite floor reaction force until time 0.13[s]. Table 5 showed the correlation coefficients between the measured values of the force plates and the floor reaction force derived using the average motion mode function, the floor reaction force derived using the motion mode function of one subject, and the floor reaction force derived using nine measured parts, respectively. Each correlation coefficient was derived using the data after 0.13[s].

TABLE V. CORRELATION COEFFICIENTS BETWEEN THE FLOOR REACTION FORCE MEASURED BY THE FORCE PLATE AND THE FLOOR REACTION FORCE DERIVED USING THE AVERAGE MOTION MODE FUNCTION, THE FLOOR REACTION FORCE DERIVED USING THE MOTION MODE FUNCTION OF ONE SUBJECT, AND THE FLOOR REACTION FORCE DERIVED USING NINE MEASUREMED PARTS IN THE VERTICAL DIRECTION AND THE WALKING DIRECTION.

Function	Vertical direction	Walking direction
Average motion mode function	0.934	0.887
Motion mode function of one subject	0.924	0.952
Nine measured parts	0.952	0.933

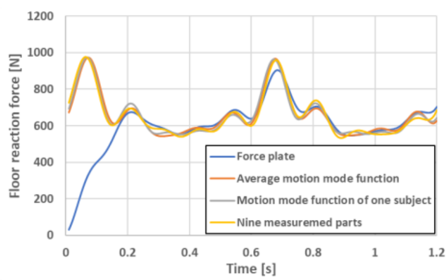

Figure 20. Comparison of the floor reaction force derived from the force plate measurements and the average motion mode function, the floor reaction force derived from the motion mode function of one subject, and the floor reaction force derived from nine measuremed parts in the vertical direction. The time up to 0.13 seconds is not included in the comparison because it is the point where the composite floor reaction force cannot be measured because one foot is grounded outside the force plate.

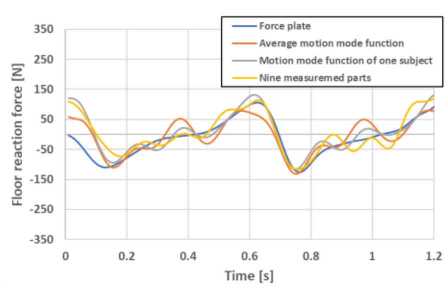

Figure 21. Comparison of the floor reaction force derived from the force plate measurements and the average motion mode function, the floor reaction force derived from the motion mode function of one subject, and the floor reaction force derived from nine measuremed parts in the walking direction. The time up to 0.13 seconds is not included in the comparison because it is the point where the composite floor reaction force cannot be measured because one foot is grounded outside the force plate.

Table 5 showed that in the vertical direction, the correlation coefficients of the method using the average motion mode function were almost equal to those of the method using the motion mode function of one subject and the method using nine measured parts, indicating that the method using the average motion mode function also ensured accuracy. Also, from Figure 20, the bimodal feature [14] that appears in the vertical floor reaction forces of one foot on each side in healthy walking appears as a trimodal feature in the composite vertical floor reaction forces presented. This result can be attributed to the fact that the acceleration of each unmeasured part was estimated accurately.

Table 5 showed that in the direction of travel, the correlation coefficients of the method using the average motion mode function and the method using nine measured parts were slightly lower than those of the method using the motion mode function of one subject. The method using the averaged motion mode function, shown by the solid orange line in Figure 21, included high-frequency waves, resulting in error from the target value measured by the force plates. This factor is thought to have been caused by a decrease in the estimation accuracy of the upper and lower limbs in the acceleration estimation of the unmeasured parts. However, the walking direction, which did not show good accuracy in the acceleration estimation, showed a much higher correlation coefficient and improved accuracy in the floor reaction force than in the acceleration. This can be attributed to the fact that the accuracy of acceleration estimation for the large mass part was high, which reduced the influence of the small mass part whose acceleration estimation accuracy was low. In this paper, the acceleration of the unmeasured part was estimated using a common order. However, since the magnitude of the higher-order components differs depending on the part, there is a limit to the accuracy of the estimation using a common order, but it is thought that the amplitude and phase are aligned by using a common order to the higher-order components.

VI. CONCLUSION AND FUTURE WORK

In this paper, in order to reduce the number of sensors used for estimating the floor reaction force, we proposed a method to estimate the acceleration of the unmeasured part from the measured part using the average motion mode function derived from the trials of multiple subjects, and to estimate the composite floor reaction force in two directions from these results. The average motion mode function can be derived by dividing the unmeasured part by the measured part for each order, and is a function that shows the relationship between the motion of the measured part and the unmeasured part. In this paper, we improved the versatility of the system by using the average motion mode function, which is the average of the motion mode functions of five subjects for each order.

In this paper, we first showed the correlation coefficients between the measured values of MC and the acceleration estimated using the average motion mode function and the acceleration estimated from the motion mode function of one subject, respectively, and indicated the usefulness of the method for estimating the acceleration of the unmeasured part shown in Section Ⅳ.A using the average motion mode function. Next, the correlation coefficients were used to examine which part of the body is suitable to be selected as the measured part when estimating the acceleration of the unmeasured part. In this paper, the measured and unmeasured parts were selected by focusing on the parts with a large mass, which is important in estimating the floor reaction force. The optimal part where the acceleration of the unmeasured part was estimated with high accuracy differed depending on the subject and direction. However, when the measured part was the upper trunk, the acceleration of the upper trunk, average thigh, and lower trunk, where the mass of the whole body is particularly large, was estimated accurately. To reduce the number of sensors, we thought it appropriate to select the upper trunk as the measured part. In the future, we will investigate whether it is possible to estimate the floor reaction force more accurately by changing the measured part according to the estimated direction. Finally, the floor reaction forces in the vertical and walking directions were derived by estimating the accelerations of the remaining 8 parts using the average motion mode function with the upper trunk as the measured part. The correlation coefficient between the floor reaction force measured by the force plates and the floor reaction force derived using the average motion mode function was 0.934 in the vertical direction and 0.887 in the walking direction. The walking direction, which did not show good accuracy in the acceleration estimation, showed a much higher correlation coefficient and improved accuracy in the floor reaction force. In addition, the bimodal characteristics that appear in the vertical floor reaction forces of one foot on each side in healthy walking were trimodal in the composite vertical floor reaction forces shown in this study, which well captured the characteristics of healthy walking. Therefore, the usefulness of the proposed method for estimating the floor reaction force using the average motion mode function was confirmed.

In the future, we will aim to further improve the versatility of the system using the average motion mode function, and propose a system that can accurately estimate the floor reaction force even for unknown subjects with different paces.

ACKNOWLEDGMENT

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