Estimation of Body Part Acceleration While Walking Using Frequency Analysis

Estimating head acceleration from movement of upper trunk

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Abstract— **This study group is developing a mobile system that can easily estimate the floor reaction force with a small number of wearable sensors for reduced user burden. In this study, we propose a method to measure accelerations by wearable inertial sensors and we estimate the floor reaction force based on these measurements. In previous works, the number of sensors was reduced from 15 body parts to 5 selected body parts, in order to decrease burden on a user. However, the estimation accuracy also decreased. Therefore, in this paper, we consider reducing the number of sensors without sacrificing accuracy. In the previous report, the relations between the acceleration of each part of the body were quantified by analyzing the acceleration of each part in the Fourier analysis and expressing it in the frequency domain. This paper quantifies the relations between the accelerations of the head and the upper trunk using previously reported methods, then estimates the head acceleration from the upper trunk acceleration. As a result, it was possible to capture the characteristics of the head acceleration in two directions while walking and it was also possible to estimate it accurately.**

Keywords- Fourier analysis; gait analysis; motion mode function.

I. INTRODUCTION

Walking is one of the most familiar activities performed by many people. Gait analysis data are important information in the fields of healthcare, clinical medicine and sports, so they are effective for health promotion, rehabilitation, improving athletic function in athletes, and so on. General gait analysis is performed by combining an optical Motion Capture (MC) system and force plates. The advantage of this measurement system is that it enables detailed analysis of gait, such as calculation of joint moments. However, this measurement system is very expensive, moreover there is a limit to the measurement range for using installation type equipment. To address this problem, our study group proposed a method for estimating floor reaction forces using wearable inertial sensors and succeeded in making it mobile [1]. This method divided the body into 15 parts, as reported by Ae et al. [2], then it was shown that the sum of the inertial forces and gravity obtained from each of these parts is balanced with the measured values of the force plate. To accurately estimate the floor reaction force, it is achievable to use each inertial force derived from the acceleration measured from 15 parts of the whole body. However, for a simple system to reduce the burden on the user, the idea is to reduce the number of sensors. Attempts to estimate using a small number of inertial sensors selected so far resulted in poor accuracy [1].

Therefore, in this paper, we consider reducing the number of sensors without sacrificing accuracy. One method is to estimate the acceleration at the unmeasured part from the measured part, which reduces the number of sensors and the loss of accuracy caused by the reduction in the number of sensors. For this purpose, it is necessary to understand the relation between the movement of the measured part and the unmeasured part. However, it is not easy to describe the relation quantitatively using the time history waveforms as they are. Hence, in the previous report [3], walking was regarded as a periodic motion and Fourier analysis was performed on the acceleration of each part. As a result, using the acceleration expressed in the frequency domain, it was possible to extract the characteristics of each part of the movement. Based on this, the unmeasured part was divided by the measured part for each frequency component. This is called “motion mode function” in this paper. This motion mode function enabled us to quantify the relation between the acceleration of the two target parts. In this paper, the acceleration at the unmeasured part is estimated in the frequency domain using the obtained motion mode function and is converted into an acceleration in the time domain by inverse Fourier analysis. The usefulness of the method is investigated by comparing the estimated acceleration with the measured acceleration.

This report shows the estimation results of the acceleration in the vertical direction and walking direction, with the head and the upper trunk as target parts, where the difference in movement is relatively large.

II. METHOD

A. Method for obtaining acceleration data for Fourier analysis

The development of a simple system using inertial sensors is our goal. However, in this paper, acceleration data

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for analysis are acquired using MC as a basic study to estimate acceleration at the unmeasured part.

In the experiment, MC (manufactured by Motion Analysis Co., Ltd.), force plate 3 units (manufactured by Tec Gihan Co., Ltd., TF-6090-C 1 unit, TF-4060-D 2 units), metronome were used. Force plates are strain-gauge transducer that can measure forces, moments, and centers of pressure. Acceleration is obtained by attaching recurrent markers to the head and upper trunk, as defined by Ae et al. [2], as shown in Figure 1.

Figure 1. Position of recursive markers.

Two healthy subjects (male: age 22±0, height 1.75±0.05 [m], weight 65±5 [kg]) are measured and 10 steps are taken from the start of walking. The acceleration data to be examined are for one gait cycle of two steps (1.2 seconds) of the fifth step and the sixth step, which are steady walking. The cadence is set to 100 BPM (0.6 seconds per walking cycle) with a metronome. Each subject measure 15 times of trial data, according to the rhythm of the metronome. The accuracy of the analysis is improved if both ends of the acceleration data of one gait cycle are the same. Therefore, measurements were taken after sufficient walking exercises by following per the metronome so that both ends of the data were aligned as much as possible. The acceleration data used for the analysis were obtained by sampling at a sampling frequency of 100 Hz and smoothed by low-pass processing with a cutoff frequency of 9 Hz.

B. Estimation of acceleration at the unmeasured part

First, the magnitude and phase of each frequency component are calculated by the Fourier analysis of the acceleration data of the head and the upper trunk, respectively. This result shows which frequency components are important in estimating the acceleration.

Next, the motion mode function for estimating the acceleration of the unmeasured part is obtained. In this paper, the head acceleration is estimated from the upper trunk acceleration, assuming the upper trunk as the measured part and the head as the unmeasured part. The motion mode function is derived by dividing the head by the upper trunk for each frequency component using the Fourier analysis results of the head and the upper trunk. Finally, the head acceleration is estimated by multiplying the obtained motion mode function by the Fourier analysis result of the upper trunk. In this paper, the motion mode function is obtained as an average motion mode function using 14 times of trial data, and the head acceleration is estimated from the remaining one trial data.

III. RESULT

Since similar results were obtained in two subjects, only the results for subject A are shown.

First, the acceleration data of the head and the upper trunk obtained from the experiment are shown. Figure 2 and Figure 3 show the vertical direction and the walking direction, respectively. The blue line is the head acceleration, and the orange line is the upper trunk acceleration. In the vertical direction shown in Figure 2, the acceleration waveform showed a similar trend. In the walking direction shown in Figure 3, the upper trunk acceleration was higher than the head acceleration. However, the relation between the movement of the head and the upper trunk in both the vertical and walking directions could not be quantified, and it is not possible to estimate the head acceleration from the upper trunk acceleration by correcting the constants.

Next, the acceleration data of the head and the upper trunk are decomposed into frequency components by a Fourier analysis, and the magnitude and phase are obtained for each direction. From the results, the important frequency bands for estimating the acceleration of the unmeasured part of the head were identified. The magnitude and phase in the vertical direction are shown in Figure 4 for the head and in Figure 5 for the upper trunk. The magnitude and phase in the walking direction are shown in Figure 6 for the head and in Figure 7 for the upper trunk. The magnitude and phase are shown as the mean and standard deviation calculated from 14 times of trial data.

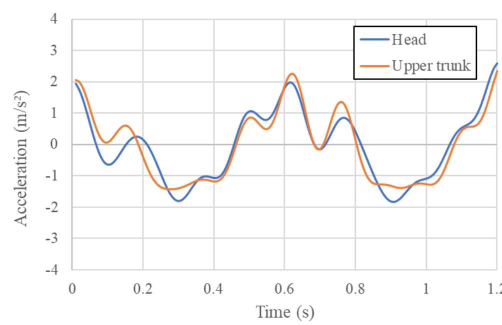

Figure 2. Acceleration data of the head and upper trunk in the vertical direction.

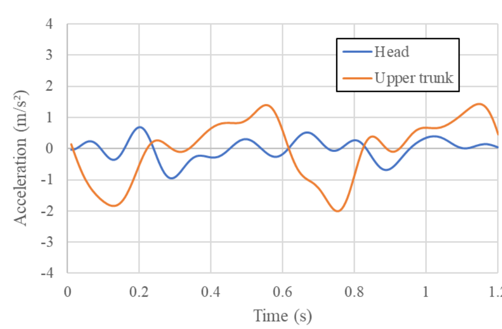

Figure 3. Acceleration data of the head and the upper trunk in the walking direction.

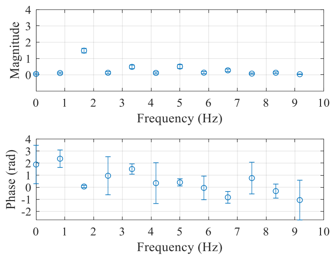

Figure 4. Result of frequency analysis of the head acceleration in the vertical direction.

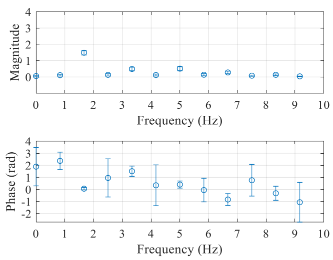

Figure 5. Result of frequency analysis of the upper trunk acceleration in the vertical direction.

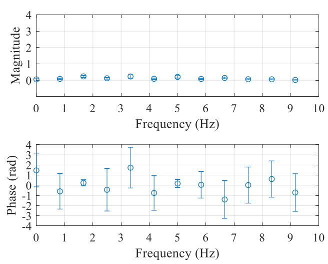

Figure 6. Result of frequency analysis of the head acceleration in the walking direction.

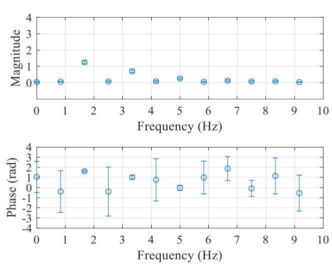

Figure 7. Result of frequency analysis of the upper trunk acceleration in the walking direction.

From Figure 4 to Figure 7, the gait frequency component (1.667Hz) and its integer multiple components have a magnitude in both the vertical direction and the walking direction. In particular, the magnitude of the gait frequency component is extremely large in the vertical direction. In the walking direction, there is a magnitude in the second-order component of the gait frequency in the upper trunk, and the magnitude in the head is small from the low-frequency bands to the high-frequency bands.

Hence, the movement in the vertical direction was found to be highly dependent on the gait frequency component. In the walking direction, the movement of the head was small, and the movement of the upper trunk was found to be significantly involved up to the second-order component of the gait frequency.

Next, the motion mode functions of the head and the upper trunk are obtained. The magnitude and phase difference of the calculated average motion mode function are shown in the vertical direction in Figure 8 and the walking direction in Figure 9. Gains and phase differences of the results are shown as means and standard deviations obtained from 14 times of trial data.

Next, the motion mode functions of the head and the upper trunk are obtained. In this paper, the motion mode function is defined as the average of the motion mode functions of 14 times of trial data. The magnitude and phase difference of the calculated motion mode function are shown in the vertical direction in Figure 8 and the walking direction in Figure 9. Gains and phase differences of the results are shown as means and standard deviations obtained from 14 times of trial data.

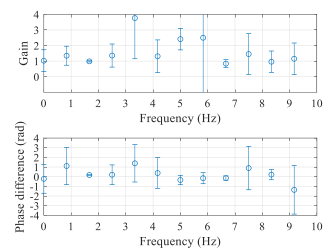

Figure 8. The average motion mode function in the vertical direction with the input as the upper trunk and the output as the head.

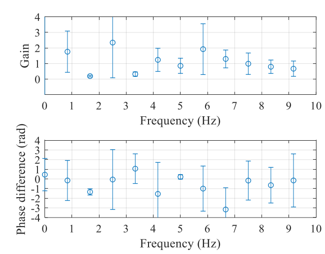

Figure 9. The average motion mode function in the walking direction with the input as the upper trunk and the output as the head.

In this paper, the relation between the head acceleration and the upper trunk acceleration, focusing on the gait frequency component and its second-order component indicated. Figure 8 shows that the second-order component has a large gain, while the most important gait frequency component has gain around 1 and phase difference around 0. From this, it can be seen that in the vertical direction, the head and the upper trunk have similar movements in the most important gait frequency component. From Figure 9, it was found that in the walking direction, the gain was small and the amplitude of the head was smaller than that of the upper trunk in the low order components, which are main components.

Finally, using the motion mode function obtained from the 14 times of trial data shown in Figure 8 and Figure 9, the head acceleration was estimated from the measured upper trunk acceleration for the remaining one trial data. From the results in Figure 4 to Figure 7, it was found that the gait frequency (1.667Hz) and the second-order component are greatly involved in the movement. However, in this paper, we also focus on high-frequency components that are less involved in motion but have a magnitude. Therefore, the inverse Fourier transform is performed using a total of 5 points of the gait frequency component and its integer multiple components to convert them into a waveform in the time domain. The measured and estimated head accelerations are shown in Figure 10 in the vertical direction and Figure 11 in the traveling direction. The red line shows the estimated acceleration and the blue line shows the measured acceleration.

Since similar results were obtained in two subjects, only the results for subject A are shown. The estimation accuracy of the estimated acceleration compared to the measured acceleration was considered using the correlation coefficient, and the results are shown in TABLE I.

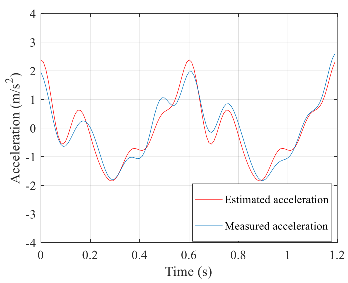

Figure 10. Comparison of head vertical acceleration estimated using the mean vertical motion mode function with measured values.

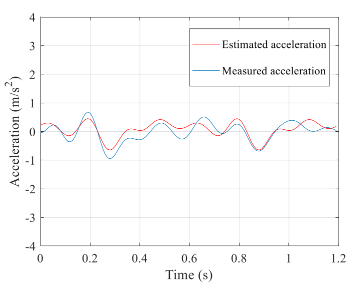

Figure 11. Comparison of head acceleration in the walking direction estimated using the mean vertical motion mode function with measured values.

TABLE I. CORRELATION COEFFICIENT BETWEEN ESTIMATED ACCELERATION AND MEASURED ACCELERATION

	Vertical direction	**Walking direction**
Correlation Coefficient	0.961	0.822

It can be seen from TABLE I that the correlation is strong in the vertical direction and that the estimation can be performed with high accuracy. In the walking direction, the correlation coefficient is smaller than that in the vertical direction, but the correlation is stronger. Consequently, it is found that the estimated the head acceleration derived from the average motion mode function shows the same tendency as the measured acceleration.

IV. CONCLUSION AND FUTURE WORK

In this paper, as a basic study to reduce the number of sensors used in estimating the floor reaction force, the acceleration at the unmeasured part was estimated using a motion mode function that describes the relation between the motion of each part. In this paper, the head acceleration is estimated from the upper trunk acceleration, with the upper trunk as the measured part and the head as the unmeasured part. As a result of Fourier analysis of the head acceleration and the upper trunk acceleration, it was found that the gait frequency and its integer multiple components have magnitude and are important for estimating the acceleration. Therefore, the inverse Fourier transform is performed using a total of 5 points of the gait frequency component and its integer multiple components to estimate the waveform of the head in the time domain. As a result, the correlation between the measured acceleration and the estimated acceleration at the head was high in both the vertical direction and the walking direction, and it was possible to make an accurate estimation.

In this paper, the acceleration of the head is estimated from the upper part of the trunk as an example to demonstrate the usefulness of the proposed method. In the future, by applying the proposed method to the entire body, the number of sensors used to accurately estimate the floor reaction force will be reduced. This will build a mobile system for healthcare wearable sensing.

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