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Relationship between joint motion and acceleration during single-leg standing in healthy male adults

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Abstract. [Purpose] The purpose of this study was to clarify the relationship between acceleration and joint movement by synchronizing accelerometers and a three-dimensional motion analysis system, and to show the utility of an accelerometer as a postural control assessment tool. [Subjects and Methods] Head, lumbar, shank accelerations and various joint angles during single-leg standing were measured of 20 healthy males. Root mean squares of acceleration and joint angle were calculated. Fast Fourier transform analysis was performed for head, lumbar, and shank accelerations, and the median frequencies were calculated. Then, principal component analysis was performed for the median frequency of each acceleration. Stepwise multiple regression analysis was also used to examine the relationship between joint angle and acceleration. [Results] The score of the first principal component was highest for shank acceleration, while that of the second principal component was highest for lumbar and head accelerations. In multiple regression analysis, hip flexion/extension and adduction/abduction were identified as variables associated with head acceleration. [Conclusion] We were able to confirm the aggregation of acceleration into two components, which we interpreted as postural control strategies using primarily the ankle and hip joints. Furthermore, though multiple regression analysis, we were able to clarify the joint movement indicated by acceleration of each segment.

Key words: Postural control, Acceleration, Ankle sprain

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INTRODUCTION

There are various risk factors of ankle sprain[1)], but postural control deficit is considered a major factor. Since Freeman's landmark work in 1965[2)], center of pressure (COP) assessment of postural control with respect to ankle sprain has been widely utilized and published[3–6)]. Many different force plate measurements have been presented in the literature; however, there is no consensus on the best measure to use, or on the relationship between ankle sprain and postural control deficit[7)]. COP is the result of gravity and acceleration of body segments[4)]. Therefore, even if the same COP is measured, the movement of each body segment may differ. Knapp et al.[6)] pointed out the limits of measuring COP alone, and suggested the need to verify the possibility that individuals with chronic ankle instability use a variety of compensatory mechanisms to maintain balance. A novel approach to investigating the relationship between ankle sprain and postural control is measurement of the movements of each body segment, and in our earlier studies, accelerometers were used to assess the motion of various body segments with the aim of developing a new perspective on this relationship.

Accelerometers can measure small movements of a target site. For example, if an accelerometer is placed on the shank, it can measure small movements in each plane of rotation, anteroposterior (AP) and mediolateral (ML) tilt, and translational motion of the shank produced by adjacent joint motion. Accelerometers have recently attracted attention as an instrument for measuring postural control[8, 9)]. Our results assessing postural control by measurement of COP and acceleration suggest that individuals with a history of ankle sprain have a higher shank-to-head acceleration ratio and different postural control characteristics than individuals without a history of ankle sprain, in spite of their being no significant difference in their path lengths of COP[10)]. In addition, in individuals with a history of ankle sprain, maximum acceleration of the shank reflects COP velocity and the amplitude of ML sway[11)]. In other words, although COP results can be identical, the degree of the effect of body movements on COP can differ between individuals with and without ankle sprain. We speculate that the reason for the inconsistencies regarding the best measure to use and the relationship between ankle sprain and postural control

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may be the influence of the postural control strategy used. In our previous study, we were able to detect compensatory body movements by measuring acceleration in addition to COP[10, 11]. Thus, accelerometric measurement of the movement of each body segment and its effect on COP will help enhance our understanding of postural control capacity and ankle sprain. However, it is currently unclear how the acceleration results obtained in this manner reflect body movement, and previous studies have not, therefore, been able to determine the actual postural control strategy of individuals with a history of ankle sprain.

Recently, frequency analysis has been used to assess change over time (e.g. COP and joint movement during single-leg standing), and various interpretations of the frequency data have been made[12–14]. In single-leg standing, a value of <0.10 Hz is considered to reflect the effect of visual control, a value of 0.10–0.39 Hz is considered to reflect the effect of vestibular sensation, a value of 0.39–1.56 Hz is considered to reflect the effect of cerebellar function, and a value of 1.56–6.25 Hz is considered to reflect the effect of spinal reflexes and muscle activity. In addition, with regard to postural control strategies, principal component analysis of joint movement frequencies has demonstrated high principal component scores for the ankle in ankle joint strategies, as well as high scores for the trunk and upper limbs in hip joint strategies[14]. Thus, frequency analysis can be used to assess task changes with time. Therefore, in this study, we measured head, lumbar, and foot acceleration of healthy male adults during single-leg standing, and tried to interpret the acceleration data using frequency analysis. The purpose of this study was to clarify the relationship between acceleration and joint movement by synchronizing accelerometer data with data captured by a three-dimensional motion analysis system.

SUBJECTS AND METHODS

Subjects

Twenty healthy active men from the university with no known lower-limb pathology, lower-limb injury, or central nervous system abnormality (age, 23.0 ± 3.5 years; height, 171.1 ± 4.7 cm; weight, 65.5 ± 6.4 kg) volunteered to participate in this study. Prior to assessment, the mean Cumberland Ankle Instability Tool[15] (CAIT) score was 29.4 ± 0.7 points (maximum, 30 points), and none of the subjects had symptoms of ankle instability. The dominant leg, which was defined as the limb used to kick a ball[16], was the right in 18 subjects and the left in 2. Ethical approval for this study was obtained from the Ethics Committee of Gunma University (approval code 14-2). This study was conducted in accordance with the Declaration of Helsinki. Written informed consent was obtained from all the participants, and the rights of all subjects were protected.

Methods

Each acceleration and joint movement during single-leg standing was measured. Subjects were instructed to stand barefoot as still as possible, with their arms folded across their chests, while standing on their dominant limb and holding the opposite limb with slight knee flexion[17]. They were also instructed to look straight ahead at a target point. Acceleration was measured using triaxial accelerometers (MVP-RF8-AC; MicroStone Corp., Nagano, Japan). Accelerometers were placed on the forehead, L3[8, 9], and above the lateral malleolus of the support leg to measure accelerations of the head, lumbar area, and shank, respectively[10, 11]. Accelerometers were fixed with double-sided sticking tape and velcro belts. The AP and ML components were measured at a sampling frequency of 50 Hz. For three-dimensional motion analysis during single-leg standing, infrared reflective markers were affixed to a total of 17 landmark positions: the bilateral acromia, lowermost ribs, anterior superior iliac spines (ASIS), posterior superior iliac spines (PSIS) and greater trochanters, the medial and lateral sides of the knee joint space, medial and lateral malleoli of the ankle, medial and lateral sides of the calcaneus, and the head of the second metatarsal bone of the supporting leg. The markers on the calcaneus were affixed as previously described by Simon et al[18]. The sampling frequency was 50 Hz, similar to the accelerometer measurement. Once the single-leg standing of the subject was stable, measurements were taken after synchronizing the acceleration data and three-dimensional data via a synchronizing switch (MVP-RFS-RC04, MicroStone Co., Ltd.). The single-leg standing time was 30 s, and data from the middle 20 s was used in the analysis. Prior to measurement, each subject practiced maintaining single-leg standing and then executed the single-leg standing task three times. During the measurements, the COP was concurrently measured using a force platform (AMTI Corp.), and the trial in which the root mean square (RMS) values of acceleration were the highest was selected for data analysis. If the elevated leg touched the floor or if the position of the supporting foot deviated during single-leg standing, the trial was excluded.

Acceleration data were filtered using a high-pass filter with a cutoff frequency of 0.5 Hz to eliminate convergent gravity components using vibration displacement analysis software (MVP-RF-S ver. 1.0.8; MicroStone Corp., Nagano, Japan)[19]. Using analytical software, KineAnalyzer (Kissei Comtec Co., Ltd.), the midpoint between the center of the joint and each marker was calculated for the hip, knee, and ankle; and the joint angles of the trunk, hip, and knee were computed three-dimensionally from the angle formed by the lines connecting markers and the long axis. The definitions of the joint angles are shown in Table 1 and Fig. 1. The RMS values of acceleration and joint angle were calculated using KineAnalyzer. Fast Fourier transform (FFT) analysis was performed on the head, lumbar, and shank acceleration values using KineAnalyzer, and the median frequencies of the AP and ML directions were calculated. FFT analysis was performed using 2048 FFT points and a Hamming window function.

Principal component analysis was performed on the median frequency of each acceleration, and we attempted to aggregate and interpret the data using the number of components and the principal component scores. With acceleration as the dependent variable, and the joint angle as the independent variable, stepwise multiple regression analysis was used to examine the relationship between the joint angle and acceleration. Acceleration and the joint angle in the same

Table 1. Definitions of joint angles

Item		Definition
Trunk	Forward/backward bend	Angle between line from center of both acromions to center of both ribs (lowest part) and line from center of both ribs (lowest part) to center of plane containing ASIS and PSIS
	Lateral bend	
	Rotation	Angle between line connecting both acromions and line connecting both ASISs
Hip	Flexion/extension	Angle between line perpendicular to plane of both ASISs and both PSISs and femur long axis (from hip joint center to knee joint center)
	Adduction/abduction	Angle between line connecting both ASISs and femur long axis
	Rotation	Angle between line from center of both ASISs to center of both PSISs and line perpendicular to plane of 3 femur markers
Ankle	Plantarflexion/dorsiflexion	Angle between tibia long axis (from knee joint center to ankle joint center) and line from ankle joint center to head of second metatarsal bone
	Inversion/eversion	Angle between intermalleolar axis (from medial malleolus to lateral malleolus) and line from medial to lateral points on calcaneus
	Shank rotation	Angle between intermalleolar axis and line from medial to lateral points on calcaneus

Each joint angle was calculated on frontal, sagittal, and horizontal planes.
ASIS: anterior superior iliac spine; PSIS: posterior superior iliac spine

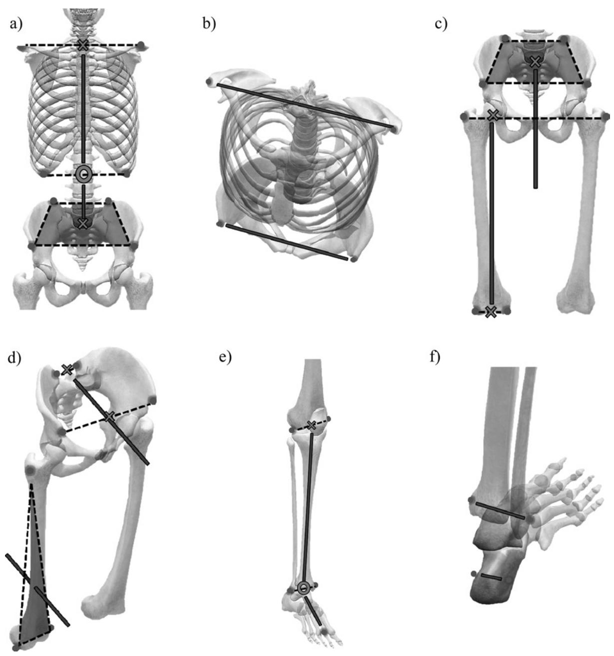

Fig. 1. Description of joint angles
a) Trunk forward/backward bend and lateral bend
b) Trunk rotation
c) Hip flexion/extension and adduction/abduction
d) Hip rotation
e) Ankle plantarflexion/dorsiflexion
f) Ankle inversion/eversion and shank rotation

plane were selected as the variables for the analysis. Statistical analysis was performed using IBM SPSS Statistics Ver. 21 for Windows, with a significance level of 5%.

RESULTS

The RMS values of acceleration and the joint angle values, and the median frequencies of the accelerations are shown in Table 2. Principal component analysis found the values were aggregated in two components (Table 3). The principal component scores of the first principal component showed the highest values for shank acceleration in the AP direction at 0.87, followed by shank acceleration in the ML direction; the scores were negative for head acceleration in both the AP and ML directions. The principal component scores of the second principal component were negative for shank acceleration in both the AP and ML directions, but highly positive for lumbar and head accelerations.

In the multiple regression analysis, hip flexion was determined as the sole variable associated with head acceleration in the AP direction, and hip adduction/abduction was the sole variable associated with head acceleration in the ML direction (Table 4). Their standard partial regression coefficients (β) were 0.485 ($p = 0.030$) and 0.606 ($p = 0.005$), respectively. Moreover, hip adduction/abduction was detected as the sole variable associated with lumbar acceleration in the ML direction, with a β of 0.588 ($p = 0.006$). Ankle eversion/inversion was detected as the sole variable associated with shank acceleration in the ML direction, with a β of 0.481 ($p = 0.032$). Lumbar and shank accelerations in the AP direction showed no significant relationships.

DISCUSSION

In this study, principal component analysis (PCA) was performed on the median frequency values of head, lumbar, and shank accelerations, and a comprehensive interpretation of acceleration data was attempted. PCA aggregated the acceleration data into two components: a first principal component and a second principal component. Looking at the principal component scores, the first principal component scores exhibited high positive values for shank acceleration in the AP and ML directions, whereas they exhibited negative values for head acceleration in the AP and ML directions. Based on this, we conclude that the first principal component reflects body movement derived from quick movement of

Table 2. Acceleration (m/s^2) and joint angle (degree) outcome measures and frequency analysis results for acceleration (Hz)

		Head	Lumbar	Shank
Acceleration	AP	0.07 ± 0.01	0.07 ± 0.01	0.19 ± 0.05
	ML	0.12 ± 0.02	0.09 ± 0.02	0.13 ± 0.04
Median frequency	AP	1.59 ± 0.57	3.98 ± 1.33	11.26 ± 3.06
	ML	2.11 ± 0.62	5.87 ± 1.72	18.60 ± 2.22
		Trunk	Hip	Ankle
Joint angle	Sagittal	0.49 ± 0.20	0.49 ± 0.29	0.46 ± 0.18
	Frontal	0.55 ± 0.23	0.50 ± 0.26	0.38 ± 0.13
	Horizontal	0.50 ± 0.16	0.83 ± 0.31	0.68 ± 0.33

All data are presented as mean ± SD. Root mean square values were calculated for acceleration and joint angle. Trunk and hip flexion/extension and ankle dorsiflexion were calculated in the sagittal plane. Trunk lateral bend, hip adduction/abduction, and ankle eversion/inversion were calculated in the frontal plane. Trunk, hip, and lower thigh rotation were calculated in the horizontal plane. Median frequency of acceleration was calculated by using the fast fourier transform method.
AP: anteroposterior; ML: mediolateral

Table 3. Principal component analysis results for median frequency of acceleration

		First main component	Second main component
Interpretation		Postural control strategy centered ankle joint	Postural control strategy centered hip joint
cumulative contribution ratio		37.0 %	61.2 %
Head acceleration	AP	−0.15	0.72
	ML	−0.68	0.26
Lumbar acceleration	AP	0.36	0.71
	ML	0.56	0.55
Foot acceleration	AP	0.87	−0.14
	ML	0.72	−0.21

AP: anteroposterior; ML: mediolateral

Table 4. Multiple regression analysis results

Dependent variable	Related variable	R^2	Intercept	Standard partial regression coefficient (β)
Head Ac				
AP	Hip flex/ext*	0.24	0.07	0.49
ML	Hip abd/add**	0.37	0.10	0.61
Lumbar Ac				
AP	–			
ML	Hip abd/add**	0.35	0.07	0.59
Shank Ac				
AP	–			
ML	Ankle inver/ever*	0.23	0.08	0.48

Multiple regression analyses were performed with each acceleration as dependent variables and each joint angle as explanatory variables. Analyses were performed using stepwise selection of explanatory variables. *$p < 0.05$. **$p < 0.01$.
Abd: abduction; Ac: acceleration; add: adduction; AP: anteroposterior; ever: eversion; ext: extension; flex: flexion; inver: inversion; ML: mediolateral

the ankle joint. The second principal component exhibited results that paired with the first principal component, and exhibited high positive values for head and lumbar accelerations in the AP and ML directions. In short, the second principal component reflects body movement derived from quick movement of the lumbar region and head. The postural control strategy in the single-leg standing can thus be divided into two main strategies: a strategy that primarily uses the ankle joint, and a strategy that mainly uses the hip joint. The former encompasses a variety of methods, such as ankle strategy, moving the COP, the inverted pendulum model, and an in-phase pattern; while the latter encompasses the hip strategy, segment acceleration, counter rotation, and a counter-phase pattern[19, 20]. The multi-joint coordination patterns in the frontal plane were examined using frequency domain principal component analysis of 14 joints' angular motion time series[14]. In that study, in the case of ankle strategy, the principal component score of ankle motion was high, and the frequency was mainly 0.8–1.2 Hz. On the other hand, in the case of hip strategy, the principal component score of the trunk and upper limbs was high, and the frequency was mainly 0.3–1.0 Hz, which is lower than that of ankle motion[14]. In our present study, because frequency analysis of acceleration was used, the frequency values were different, but the principal component scores had many similar features. Thus, a comprehensive interpretation is possible if the first principal component reflects a strategy that primarily uses the ankle joint, and the second principal component reflects a strategy that primarily uses the hip joint. There is a possibility that shank acceleration can be used to detect body movement that occurs due to the ankle strategy, while head and lumbar acceleration can be used to detect body movement that occurs due to the hip strategy. Therefore, accelerometer measurements of the head, lumbar region, and ankle appear to enable simple assessment of postural control strategies.

Looking further at the results of the multiple regression analysis, significant relationships between head and lumbar accelerations and hip joint movement, and between shank acceleration and ankle joint movement were observed. The R^2 values obtained by multiple regression analysis are low, and there may be limits to the interpretation of these results. However, these results do explain the results of the frequency and principal component analyses. With regard to joint movement and acceleration in the frontal plane, particularly in the ML direction, the joint movements that had significant relationships with the head, lumbar, and shank accelerations were identified. Previous studies have demonstrated that joint movements in the frontal plane are more unstable during single-leg standing, appearing as greater movements[14]. Furthermore, the proportion of individuals using a postural control strategy that primarily uses the hip joint is reportedly greater than that using the ankle joint strategy[20], and the segments above the hip joint tend to move more greatly in the ML direction. We speculate that these reasons are responsible for the significant relationships observed between acceleration and the joint movement at all three locations (head, lumbar region, and shank) in the ML direction. In a previous study of ankle sprain and postural control, Knapp et al.[6] found ML force plate measurements were significant for differentiating between those with and without CAI. Ross et al.[21] and Wikstrom et al.[5] reported mixed AP and ML results, but demonstrated the significance of ML force plate measurements. Although only the frontal plane was investigated, Tropp et al.[22] reported COP displacement and hip and ankle joint motion measurements significantly differentiated between individuals with and without ankle instability. We have also reported that individuals with a history of ankle sprain showed a higher shank-to-head acceleration ratio[10], and that the maximum acceleration of the foot reflects COP velocity and the amplitude of ML sway[11]. Thus, with regard to ankle sprain and postural control, frontal plane differences appear to be important. For the assessment of postural control, we consider that accelerometers are extremely useful novel tools that are especially adept at identifying joint movements in the ML direction which better reflect the relationship between ankle sprains and postural control.

Recent studies on the relationship between ankle sprains and postural control have focused on postural control strategies while analyzing movement from a kinematic viewpoint. Dherty et al.[23] tracked changes in COP over time using the fractal dimension method and also measured the joint angle using a three-dimensional movement analyzer. They reported seeing a difference in the hip joint angle between healthy subjects and those with a history of ankle sprains. Martinez-Ramirez et al.[9] conducted body movement, wavelet, and acceleration analyses using tri-axial inertial/magnetic sensors and the dynamic Star Excursion Balance Test (SEBT) rather than single-leg standing. Despite there being no difference in the SEBT reach distance between healthy volunteers and subjects with a history of ankle sprains, differences in the frequency characteristics and body movements were seen, indicating the importance of paying attention to the nature of the movements. These previous studies demonstrate two main points: ankle sprain patients tend to employ a strategy that uses primarily the hip joint, and the essential relationship between ankle sprains and postural control cannot be discovered by conventional COP measurement methods (total trace length, rectangular area, movement velocity, amplitude, etc.) or simple reach distance data. In the present study, head and lumbar accelerations were found to be associated with hip joint movement, while ankle acceleration was found to be associated with ankle movement. In particular, we verified that frontal plane changes can be detected, demonstrating that postural control assessment using an accelerometer can easily elucidate postural control strategies and can be applied in clinical settings, in sports, and in other fields.

By performing principal component analysis on the frequency of head, lumbar, and shank accelerations, we were able to confirm the aggregation of acceleration into two components, which appear to reflect postural control strategies using primarily the ankle joint and the hip joint, respectively. Furthermore, by examining the relationship between the joint angle and acceleration by multiple regression analysis, we were able to clarify the joint movement indicated by the acceleration of each segment. Based on these findings, we consider that postural control assessment using an accelerometer can be utilized in sports medicine and other fields, and to promote this, we have pioneered analytical methods in this study with the aim of creating indices that enable quicker and easier assessment. Topics for future research include designing a longitudinal study of postural control assessment using accelerometers, clarifying the causal relationship between ankle sprains and postural control strategies, and implementing ankle sprain prevention activities by creating a simple assessment index that can be used in clinical settings and in sports.

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