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Original Article

Trunk lean gait decreases multi-segmental coordination in the vertical direction

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Abstract. [Purpose] The strategy of trunk lean gait to reduce external knee adduction moment (KAM) may affect multi-segmental synergy control of center of mass (COM) displacement. Uncontrolled manifold (UCM) analysis is an evaluation index to understand motor variability. The purpose of this study was to investigate how motor variability is affected by using UCM analysis on adjustment of the trunk lean angle. [Subjects and Methods] Fifteen healthy young adults walked at their preferred speed under two conditions: normal and trunk lean gait. UCM analysis was performed with respect to the COM displacement during the stance phase. The KAM data were analyzed at the points of the first KAM peak during the stance phase. [Results] The KAM during trunk lean gait was smaller than during normal gait. Despite a greater segmental configuration variance with respect to mediolateral COM displacement during trunk lean gait, the synergy index was not significantly different between the two conditions. The synergy index with respect to vertical COM displacement during trunk lean gait was smaller than that during normal gait. [Conclusion] These results suggest that trunk lean gait is effective in reducing KAM; however, it may decrease multi-segmental movement coordination of COM control in the vertical direction.

Key words: Uncontrolled manifold analysis, Trunk lean gait, Center of mass

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INTRODUCTION

There is motor variability in all human motions, and human behavior has redundant degrees of freedom (DOF)[1)]. Uncontrolled manifold (UCM) analysis is a quantitative analysis to examine motor redundancy[2)]. With regard to the UCM analysis, motor variability is defined by all segmental configurations that contribute to a particular motor task, which can be divided into two variance components. One component represents the variance within UCM (V_{UCM}), which does not affect the performance variable ("good variance"). The other component represents the variance that is orthogonal to UCM (V_{ORT}), which affects the performance variable ("bad variance"). If the value of the two variance components is V_{UCM}>V_{ORT}, it can

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be concluded that the performance variable is stabilized by synergy[3]. The strength of synergy is reflected by the synergy index (ΔV), which is computed as the normalized difference between V_{UCM} and V_{ORT}[4].

Walking is one of the most common motor tasks during daily activities. The UCM analysis can be used to explain the particular organization of gait variability and can help to further understand the functional purposes in which gait variability plays a role during various task conditions[5]. Previous studies have evaluated center of mass (COM) variability using UCM analysis during the stance phase of walking[6–8]. Black et al. showed that V_{UCM} variance at heel strike of children with Down syndrome was larger than that of healthy children[6]. Qu et al. demonstrated the effects of load carriage and fatigue with respect to gait variability, and both factors were associated with motor synergy in stabilizing COM in the frontal plane[7]. Papi et al. investigated COM control variability in stroke patients during gait with and without orthoses[8]. The UCM analysis for the investigation of COM variability may become a useful index to assess the coordination of multi-segmental movements during the stance phase of gait. In addition, kinematic synergy may change in joint disorders with disability during walking and/or when various tasks are performed during walking.

Knee osteoarthritis (OA) is one of the most common musculoskeletal disorders that cause knee pain and disability during the stance phase of walking[9, 10] and mainly affects the medial compartment of the knee[11]. External knee adduction moment (KAM) during the stance phase reflects the compressive forces that act on the medial compartment[12]. Excessive mechanical stress of the medial compartment increases the risk of initiation and progression of knee OA[11]. Recent studies have reported that the gait modification of trunk lean in the frontal plane decreases KAM during the stance phase[13–15]. However, despite evidence demonstrating beneficial biomechanical effects of trunk lean gait on the knee joint, the mechanisms by which kinematic strategies are used in order to walk with a trunk lean angle remain unclear.

KAM is largely determined by the mediolateral movement of the trunk, which is related to COM movement[13, 16]. Therefore, it is thought that COM displacement in the frontal plane by leaning the trunk toward the stance limb is associated with decreased KAM during the stance phase. However, the strategy of trunk lean gait to reduce the intensity of the knee OA symptoms may affect the multi-segmental synergy to control COM displacement in the frontal plane.

Other studies have suggested that trunk lean gait may potentially worsen adverse effects (e.g., symptomatic effects, biomechanical effects on other joints, the lateral compartment of the knee, and balance)[13, 14]. Moreover, Simic et al. reported that trunk lean gait may cause difficulty in coordinating body movements during the stance phase for knee OA patients[14]. However, it is not clear how multi-segmental movements are coordinated to stabilize whole-body COM movement when the trunk lean angle is adjusted during the stance phase.

A previous study showed that trunk lean gait increased the energy cost[17]. Human walking is performed by the exchange of gravitational potential and kinetic energies[18], and it is performed by controlling periodic vertical displacement of whole-body COM during each stride[19]. Therefore, trunk lean gait may affect the synergy of COM movement in the vertical direction because it is thought that gait modification gives priority to the control of COM movement in the mediolateral direction.

The main purpose of this study was to quantify how the kinematic synergy to COM variability is affected when the trunk lean angle is adjusted during the stance phase. We hypothesized that the synergy index stabilizing COM in the mediolateral direction increases significantly during trunk lean gait compared to normal gait. Furthermore, we hypothesized that the synergy stabilizing COM in the vertical direction to control COM in the frontal plane significantly decreases during trunk lean gait compared to normal gait.

SUBJECTS AND METHODS

Fifteen healthy young adults [eight females and seven males; age, 22.5 ± 1.5 years; height, 165.4 ± 9.5 cm; and mass, 58.5 ± 10.1 kg] participated in this study. This study was approved by the Ethics Committee of the Division of Physical Therapy and Occupational Therapy Sciences, Graduate School of Health Sciences, Hiroshima University (Approval No. 1414). All subjects provided written informed consent prior to participation. None of the subjects had previously been treated for any clinical lower back and/or lower extremity conditions or had any activity-restricting medical and/or musculoskeletal conditions.

The subjects walked five times across a 10-m laboratory walkway at their preferred gait speed under two different conditions: normal and trunk lean gaits. With regard to trunk lean gait, the subjects were instructed to lean the trunk toward the study limb during the ipsilateral stance phase and to reach maximum trunk lean to the target angle after initial contact of the study limb. Usual trunk motion was encouraged during the contralateral stance phase[14, 15, 17]. Using a real-time visual feedback system, subjects were instructed to increase their trunk lean angle to the target angle equal to 10° greater than that observed during normal gait[15, 17]. This target trunk lean angle was then set on the real-time visual feedback projector screen, which was positioned directly in front of the subjects. Trunk marker positional data were streamed from the Vicon Nexus v1.8.5 software (ViconMX, Oxford, UK) to Matlab software R2014a (MathWorks, Natick, MA, USA) in real time. The Matlab software calculated and displayed the trunk lean angle animation. This real-time visual feedback system has been previously shown to be feasible in the training of gait modification[14, 15, 17]. Prior to data collection, the subjects practiced for approximately 10 min to achieve the trunk lean angle of the target. After that, data collection of gait modification conditions commenced. For each condition, data collection required five trials to ensure appropriate gait modification. If the subjects could not achieve the trunk lean angle of the target, they were provided with additional verbal feedback and encouraged to continue trying to reach the target. The error range

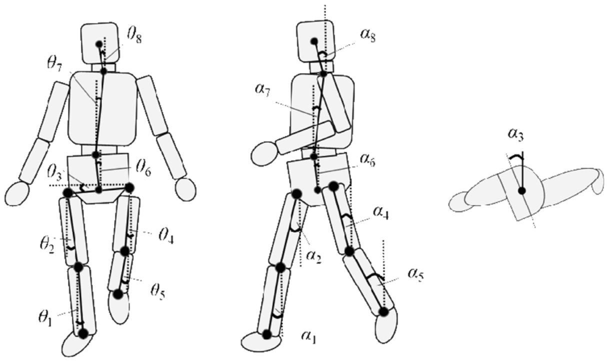

Fig. 1. A geometric model was used to extract an analytical expression for each elemental variable matrix. The left, middle, and right illustration represents a view of the frontal, sagittal, and transverse planes, respectively.

of the trunk lean angle corresponded to ± 2° during the stance phase. Because of difficulty in obtaining the exact trunk lean target angle, the five trials closest to the target angle were included in the analysis.

Infrared-reflecting markers were attached to 40 anatomical landmarks[20)]. Kinematic data during gait were collected using Vicon MX, a three-dimensional motion analysis system (ViconMX, Oxford, UK) with six infrared cameras. Kinetic data were collected using eight force plates (Tec Gihan, Uji, Japan) to measure ground reaction forces under each individual foot. These three-dimensional coordinates were collected by the three-dimensional motion analysis system at a sampling rate of 100 frame/s, and the three-dimensional ground reaction forces were collected by the force plates at a sampling frequency of 1,000 Hz. The stance phase was defined as when the vertical vector of the ground reaction was above 10 N[21)]. Marker and force plate data were low-pass filtered using a 4th-order Butterworth filter (6 Hz and 20 Hz, respectively). The range of the analysis set the stance phase (from heel strike to toe off) of the right lower extremity. The trunk lean angle was calculated as the angle between the trunk vector (a line joining the center between the line connecting the midpoint across both posterior superior iliac spines and the line connecting the midpoint across both acromia) and global vertical axis. To calculate the magnitude of trunk lean angle, the trunk lean angle was selected as the maximum value for each trial and was averaged with each number of trials. KAM was calculated by using a tibial coordinate system with the origin in the knee joint center[22)]. KAM was then normalized to each subject's body mass (N·m/kg). The KAM data were analyzed at the points of the first KAM peak during the stance phase.

Previous studies have suggested that whole-body COM could be the preferentially controlled variable to achieve stability during walking[6, 7)]. Therefore, we applied the UCM analysis to characterize the control of COM during the stance phase. Prior to UCM analysis, all kinetic data were time normalized (0–100%) from the heel strike to toe off (stance phase). The performance variables were COM displacements (we set COM in each direction: COM in the mediolateral direction and vertical direction). In our experiment, the UCM analysis generated a geometric model of the performance variable. The geometric model for COM displacement was composed of the following eight segments: 1) stance-limb shank, 2) stance-limb thigh, 3) pelvis, 4) swing-limb thigh, 5) swing-limb shank, 6) trunk, 7) thorax, and 8) head. The UCM analysis was separately performed in the mediolateral and vertical directions (Fig. 1). With regard to COM in the mediolateral direction, segments i=1–5 had motions outside in the frontal plane as defined by angles α_1, α_2, α_3, α_4, and α_5, respectively. The shank and thigh movements in the sagittal plane during gait were larger than those in the frontal plane. In addition, the pelvic movement in the transverse plane during gait was larger than that in the frontal plane; therefore, theses angles were included to account for the change in the effective length of the segments when projected onto the frontal plane[4)]. The angles α_1 and α_5 represent the projection of the vector connecting the ankle and knee joint centers of the limb to the frontal plane, effectively incorporating knee and ankle movements in the sagittal plane (α_1: stance limb, α_5: swing limb). The angles α_2 and α_4 represent the projection of the vector connecting the knee and hip joint centers of the limb to the frontal plane, effectively incorporating hip and knee movements in the sagittal plane (α_2: stance limb, α_4: swing limb). The angle α_3 represents the projection of the vector connecting both hip joint centers of the limb to the frontal plane, effectively incorporating both hip movements in the transverse plane. The geometrical models for COM delimited to the mediolateral direction in the frontal plane and to the vertical direction in the sagittal plane are described by Eqs (1) and (2), respectively.

COM in the mediolateral direction

$$=x_0 + C_1 \times M_1 \times L_1 \times \cos\alpha_1 \sin\theta_1 + C_2 \times M_2 \times L_2 \times \cos\alpha_2 \sin\theta_2 + C_3 \times M_3 \times L_3 \times \cos\alpha_3 \cos\theta_3 + C_4 \times M_4 \times L_4 \times \cos\alpha_4 \sin\theta_4 + C5 \times M5 \times L5 \times \cos\alpha_5 \sin\theta_5 + C_6 \times M_6 \times L_6 \times \sin\theta_6 + C_7 \times M_7 \times L_7 \times \sin\theta_7 + C_8 \times M_8 \times L_8 \times \sin\theta_8 \quad (1)$$

COM in the vertical direction

$=z_0 + C_1 \times M_1 \times L_1 \times \cos\alpha_1 + C_2 \times M_2 \times L_2 \times \cos\alpha_2 + C_3 \times M_3 \times L_3 \times \cos\alpha_3 \sin\theta_3 + C_4 \times M_4 \times L_4 \times \cos\alpha_4 + C_5 \times M_5 \times L_5 \times \cos\alpha_5 + C_6 \times M_6 \times L_6 \times \cos\alpha_6 + C_7 \times M_7 \times L_7 \times \cos\alpha_7 + C_8 \times M_8 \times L_8 \times \cos\alpha_8$ (2)

where x_0 and z_0 are the segmental positions of the absolute coordinate system in the mediolateral and superior-inferior directions, respectively; C_i is the estimated position of COM_i on the segment; M_i is the proportion of the total body mass of each segment; L_i is the length of the segment; and θ_i and $\alpha_{i\ (i=1,\ 2,\ 4,\ 5,\ 6,\ 7,\ \text{and}\ 8)}$ are the segment angles relative to the frontal and the sagittal planes, respectively; and α_3 is the segment angle relative to the transverse plane[23].

A linearization approximation of the geometric model of the performance variable in each plane (sagittal or frontal) was obtained at the mean segmental configuration during each stance phase across all repetitions using the Jacobian system (J). J is the matrix of the partial derivatives that correspond to changes in the performance variables with respect to each of the segmental angles (the elemental variables)[3]. The null space of J, ε, was calculated to provide basis vectors spanning the linearized UCM. The null space has $n-d$ vectors that span UCM ($\varepsilon_1, \varepsilon_2, \ldots, \varepsilon_{n-d}$), where n represents the number of dimensions in the segmental configuration space and d represents the number of dimensions of the performance variable. For the analysis regarding the control of COM in the mediolateral direction, n=13 and for the analysis regarding the control of COM in the vertical direction, n=9. In each direction, d=1. Every percentage of each stance ($\theta - \overline{\theta}$) was projected onto the null space:

$$\Theta_{\mathrm{UCM}} = \sum_{i=1}^{n-d} \left(\theta - \overline{\theta}\right) \cdot \varepsilon_i$$

and onto a component orthogonal to this subspace:

$$\Theta_{\mathrm{ORT}} = \left(\theta - \overline{\theta}\right) - \Theta_{\mathrm{UCM}}$$

N is the number of repetitions. The variance in Θ, which did not affect good variance, was calculated as the average squared length of Θ_{UCM} per DOF over all N steps:

$$\mathrm{V}_{\mathrm{UCM}} = \sqrt{\left(n-d\right)^{-1} N^{-1} \Sigma \left(\Theta_{\mathrm{UCM}}\right)^2}$$

The variance that affected bad variance was calculated as follows:

$$\mathrm{V}_{\mathrm{ORT}} = \sqrt{d^{-1} N^{-1} \Sigma \left(\Theta_{\mathrm{ORT}}\right)^2}$$

The UCM analysis was calculated using the whole-body COM in the frontal (COM in the mediolateral direction) and sagittal (COM in the vertical direction) planes, separately. The average total variance in the segmental configuration space per total DOFs was calculated using V_{TOT}.

$$\mathrm{V}_{\mathrm{TOT}} = \left(\frac{1}{n+d}\right)\left(d\mathrm{V}_{\mathrm{ORT}} + \left(n-d\right)\mathrm{V}_{\mathrm{UCM}}\right)$$

V_{TOT} was calculated as $(12V_{UCM}+V_{ORT})/14$ with respect to COM in the mediolateral direction and as $(8V_{UCM}+V_{ORT})/10$ with respect to COM in the vertical direction. Statistically, if $V_{UCM}>V_{ORT}$ then synergy exists to stabilize the whole-body COM during the stance phase. The synergy index was calculated as follows[4]:

$$\Delta\mathrm{V} = \frac{\mathrm{V}_{\mathrm{UCM}} - \mathrm{V}_{\mathrm{ORT}}}{\mathrm{V}_{\mathrm{TOT}}}$$

The more positive ΔV is, the stronger the synergy. Non-positive values indicate the absence of synergy. When the performance variable is COM in the mediolateral direction, ΔV ranges from −14 (all variance is partitioned into V_{ORT}) to 14/12 (all variance is partitioned into V_{UCM}). When the performance variable is COM in the vertical direction, ΔV ranges from −10 (all variance is partitioned into V_{ORT}) to 10/8 (all variance is partitioned into V_{UCM}). The different components of variance (V_{TOT}, V_{UCM}, and V_{ORT}) are always positive and the index of synergy ΔV ranges from positive to negative values. Therefore, these variables do not follow a normal distribution. In order to solve this problem and to apply statistical analysis, the components of variances were log-transformed using mathematical transformations[24]. ΔV of COM in the mediolateral direction was transformed using Fisher's z-transformation:

$$\Delta Vz = \frac{1}{2}\log\left[\frac{14+\Delta V}{\left(\frac{14}{12}-\Delta V\right)}\right]$$

ΔV of COM in the vertical direction was transformed using Fisher's z-transformation:

$$\Delta Vz = \frac{1}{2}\log\left[\frac{10+\Delta V}{\left(\frac{10}{8}-\Delta V\right)}\right]$$

When the performance variable is COM in the mediolateral direction, $\Delta Vz<0.54$ represents the absence of synergy. When the performance variable is COM in the vertical direction, $\Delta Vz<0.45$ represents the absence of synergy. Prior to statistical analysis, V_{UCM}, V_{ORT}, V_{TOT}, and ΔVz were averaged across the first half (0–50%) and latter half (51–100%) of the stance phase. Whole-body COM variance in the mediolateral direction and COM variance in the vertical direction were calculated using the whole-body COM variance at each percentage of normalized movement time and were averaged across the first and latter halves of the stance phase.

To compare the difference between conditions, a t-test was performed for maximum trunk lean angle, KAM, and COM variance during the stance phase (COM variances in the mediolateral and vertical directions). To test the hypothesis that the synergy index existed, a mixed design ANOVA was performed, and it included a within-subject factor of the variance component (V_{UCM} and V_{ORT}) and between-subject factor of condition (normal and trunk lean gait). A significant main effect of the variance component ($V_{UCM}>V_{ORT}$) indicated the existence of synergy. This was followed by pre-planned, paired t-tests to compare ΔVz, V_{UCM}, and V_{ORT} between conditions. All statistical analyses were performed using IBM SPSS version 22.0 for Windows (IBM Japan, Tokyo, Japan) with significance set at $p<0.05$.

RESULTS

The maximum trunk lean angle during the stance phase of trunk lean gait was significantly larger than that of normal gait ($p<0.001$: normal gait=1.0 ± 1.5°, trunk lean gait=11.0 ± 1.0°). The KAM peak was significantly lower during trunk lean gait than during normal gait ($p<0.001$: normal gait=0.6 ± 0.1 N·m/kg, trunk lean gait=0.4 ± 0.1 N·m/kg). Tables 1 and 2 show the parameters of each variances (Table 1: mediolateral direction, Table 2: vertical direction). COM variances in the mediolateral direction and vertical direction did not differ significantly between both conditions. ANOVA revealed a significant effect of the variance component (the first half of COM in the mediolateral direction: $p<0.001$, the latter half of COM in the mediolateral direction: $p<0.01$, the first half of COM in the vertical direction: $p<0.001$, and the latter half of COM in the vertical direction: $p<0.001$) indicating the presence of synergy ($V_{UCM}>V_{ORT}$). ΔVz of COM in the mediolateral direction did not significantly differ between both conditions. ΔVz of the first half of COM in the vertical direction of trunk lean gait was significantly smaller than that of normal gait ($p<0.05$). V_{UCM} of COM in the mediolateral direction of trunk lean gait was significantly larger than that of normal gait (the first half of COM in the mediolateral direction: $p<0.001$ and the latter half of COM in the mediolateral direction: $p<0.01$). V_{ORT} of COM in the mediolateral direction did not differ significantly between both conditions. V_{UCM} of COM in the vertical direction did not differ significantly between both conditions. V_{ORT} of the first half of COM in the vertical direction was significantly larger during trunk lean gait than during normal gait ($p<0.01$). V_{TOT} of COM in the mediolateral direction was significantly larger during trunk lean gait than during normal gait (the first half of V_{TOT}: $p<0.001$ and the latter half of V_{TOT}: $p<0.01$). V_{TOT} of COM in the vertical direction did not differ significantly between both conditions.

DISCUSSION

The purpose of this study was to quantify how the kinematic synergy to COM is affected when the trunk lean angle is adjusted during the stance phase using real-time visual feedback. Although we hypothesized that the synergy stabilizing COM in the mediolateral direction increases when there is need for trunk lean angle adjustment, the results did not support this hypothesis. The findings showed that despite the fact that gait with the adjusted trunk lean angle using real-time visual feedback increased V_{UCM} during the first half of the stance phase, the synergy index to stabilize COM in the mediolateral

Table 1. The parameters of each variances in the mediolateral direction

	First half of stance phase		Latter half of stance phase	
	Normal gait	Trunk lean gait	Normal gait	Trunk lean gait
COM variance (mm^2)	41.6 ± 37.9	62.9 ± 53.7	47.6 ± 42.2	61.0 ± 46.7
ΔVz	1.7 ± 0.4	1.9 ± 0.5	1.4 ± 0.4	1.7 ± 0.5
V_{UCM} ($\times 10^{-4}$ rad^2)	2.6 ± 1.2	4.7 ± 1.6***	2.3 ± 1.1	4.5 ± 1.8**
V_{ORT} ($\times 10^{-4}$ rad^2)	1.2 ± 1.1	1.9 ± 1.6	2.0 ± 1.5	2.5 ± 2.0
V_{TOT} ($\times 10^{-4}$ rad^2)	2.3 ± 1.1	4.2 ± 1.4***	2.2 ± 1.0	4.0 ± 1.6**

Values are express as mean ± SD, **p<0.01, ***p<0.001

Table 2. The parameters of each variances in the vertical direction

	First half of stance phase		Latter half of stance phase	
	Normal gait	Trunk lean gait	Normal gait	Trunk lean gait
COM variance (mm^2)	61.2 ± 61.5	59.7 ± 27.7	58.5 ± 29.2	114.4 ± 162.4
ΔVz	1.9 ± 0.4	1.6 ± 0.3*	1.8 ± 0.4	1.5 ± 0.3
V_{UCM} ($\times 10^{-4}$ rad^2)	3.7 ± 2.1	4.2 ± 1.2	4.1 ± 2.2	5.0 ± 2.1
V_{ORT} ($\times 10^{-4}$ rad^2)	0.7 ± 0.4	1.8 ± 1.2**	1.0 ± 0.5	2.4 ± 3.1
V_{TOT} ($\times 10^{-4}$ rad^2)	3.0 ± 1.7	3.6 ± 1.0	3.3 ± 1.8	4.2 ± 1.8

Values are express as mean ± SD, *p<0.05, **p<0.01

direction displacement did not significantly differ between both conditions.

A previous study showed that V_{UCM} of children with Down syndrome at heel strike during gait was larger than that of children with typical development[6]. It is demonstrated that children with Down syndrome, who have inherently unstable mechanical systems, adopt this strategy to deal with motor dysfunction (e.g., high joint laxity, low muscle tone, and poor overall postural control)[6]. Our results showed that V_{UCM} and V_{TOT} with respect to COM in the mediolateral direction during the first and second halves of the stance phase were larger during trunk lean gait than during normal gait; however, the synergy index did not differ between both conditions. These results suggest that trunk lean gait increased total variance to maintain synergy during the entire stance phase; therefore, coordination was not increased by trunk lean gait. In a previous study[25] regarding postural control and COM displacement, it was observed that the greater the difficulty of the target task, the larger the variance for not affecting COM displacement. This study set the task of adjusting the trunk lean angle during the stance phase; therefore, it was expected that the task would change the coordination of multi-segmental movements to control COM displacement in the frontal plane. The results of this study suggest that V_{UCM} of COM in the mediolateral direction increased to reach the target trunk lean angle during the stance phase.

With respect to controlling COM in the vertical direction, we hypothesized that the synergy stabilizing COM in the vertical direction would decrease to control COM in the frontal plane. In support of our hypothesis, our results showed V_{ORT} was significantly greater during trunk lean gait than during normal gait, and ΔVz was significantly smaller during trunk lean gait than during normal gait in the first half of the stance phase. With respect to the second half of the stance phase, ΔVz and V_{ORT} did not differ significantly between both conditions. Because the magnitude of bad variance affecting the performance variable did not differ between both conditions, decrease in the synergy index as in the first half of the stance phase was not confirmed. These results indicate that trunk lean gait decreases the coordination of multi-segmental movements to control COM in the vertical direction during the first half of the stance phase. It has been reported that gait modification decreases KAM during the first half of the stance phase[13–15]. Our data were consistent with the results of previous studies with respect to the decrease of KAM during the stance phase. However, despite evidence demonstrating the decrease of knee mechanical stress of trunk lean gait, trunk lean gait may affect COM control in the vertical direction by adjusting the angle in the frontal plane.

A limitation of this study was the analysis of healthy young adults as the study subjects. A previous study showed that according to increase in the severity of OA, the trunk lean angle value was naturally greater during the stance phase[26]. In addition, there is a possibility of changing the motor variability associated with the severity of knee OA. Therefore, future research needs to examine the trunk lean gait of knee OA using similar methods. Despite the above-mentioned limitation, this study is the first to assess kinematic synergy by adjusting the trunk lean angle with real-time visual feedback during the stance phase using UCM analysis. Our findings provide important basic data to further understand the control mechanisms of multi-segmental coordination of the trunk lean gait during the stance phase.

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