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Title: Biomechanical mechanism of lateral trunk lean gait for knee osteoarthritis patients

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ABSTRACT

The biomechanical mechanism of lateral trunk lean gait employed to reduce external knee adduction moment (KAM) for knee osteoarthritis (OA) patients is not well known. This mechanism may relate to the center of mass (COM) motion. Moreover, lateral trunk lean gait may affect motor control of the COM displacement. Uncontrolled manifold (UCM) analysis is an evaluation index used to understand motor control and variability of the motor task. Here we aimed to clarify the biomechanical mechanism to reduce KAM during lateral trunk lean gait and how motor variability controls the COM displacement. Twenty knee OA patients walked under two conditions: normal and lateral trunk lean gait conditions. UCM analysis was performed with respect to the COM displacement in the frontal plane. We also determined how the variability is structured with regards to the COM displacement as a performance variable. The peak KAM under lateral trunk lean gait was lower than that under normal gait. The reduced peak KAM observed was accompanied by medially shifted knee joint center, shortened distance of the center of pressure to knee joint center, and shortened distance of the knee–ground reaction force lever arm during the stance phase. Knee OA patients with lateral trunk lean gait could maintain kinematic synergy by utilizing greater segmental configuration variance to the performance variable. However, the COM displacement variability of lateral trunk lean gait was larger than that of normal gait. Our findings may provide clinical insights to effectively evaluate and prescribe gait modification

training for knee OA patients.

1. Introduction

Knee osteoarthritis (OA) is a musculoskeletal disorder resulting in pained walking and even impaired walking ability. Excessive medial compartment loading of the knee joint during walking is a major risk factor for OA progression (Bennell et al., 2011; Miyazaki et al., 2002). The external knee adduction moment (KAM) during walking is generally regarded as a substitute evaluation for medial compartment loading of the knee joint (Andriacchi et al., 2000; Andriacchi et al., 2006). The first peak in KAM is related to the presence (Hurwitz et al., 2002) and progression (Miyazaki et al., 2002) of knee OA.

Gait modification increasing the lateral trunk lean during the stance phase of the symptomatic knee reduces peak KAM and KAM impulse (Simic et al., 2012). However, exactly how the lateral trunk lean gait reduces KAM is not well known. KAM is determined by the product of the ground reaction force (GRF) vector magnitude and the knee–GRF lever arm (KLA). The lateral trunk lean may influence KAM by causing the center of mass (COM) displacement in the frontal plane toward the stance limb during walking (Hunt et al., 2008); however, the movement strategy for how the GRF vector magnitude or KLA change to reduce KAM according to the lateral trunk lean gait is unknown. Therefore, elucidating the mechanism of underlying the reduction of KAM of the lateral trunk lean gait for knee OA may be useful for exercise instruction to promisingly reduce KAM.

Despite evidence demonstrating the benefits of lateral trunk lean gait KAM reduction, lateral trunk lean gait remained difficult for knee OA patients (Hunt et al., 2011). Simic et al. (2011) reported that lateral trunk lean gait may occur in conjunction with difficulty in coordinating body movements to achieve an adequate lateral trunk lean angle during the stance phase for knee OA patients. However, it has not been reported in terms of coordinating the body movements during lateral trunk lean gait.

Coordination of body segments during walking was investigated in terms of motor variability using uncontrolled manifold (UCM) analysis (Krishnan et al., 2013; Rosenblatt et al., 2014; Rosenblatt et al., 2015; Black et al., 2007; Qu et al., 2012; Papi et al., 2015). UCM analysis is a quantitative tool used to understand motor variability (Scholz et al., 1999; Stergiou et al., 2011). UCM analysis discerns variability of a selected functional task with many degrees of freedom (DOFs), and this analysis tests the assumption of all combinations with respect to motor elements (i.e., elemental variables) that lead to important variables produced by the motor system (i.e., performance variables). 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 projected onto UCM (V_{UCM}) that does not affect the performance variable (good variability); the other component represents the variance orthogonal to UCM (V_{ORT}) that affects the performance variable (bad variability). As good variability increases, more movement patterns are used to perform a task; bad variability increases performance variability and thus destabilizes performance

(Krishnan et al. 2013). If $V_{UCM} > V_{ORT}$, the selected performance variable is stabilized by synergy (Scholz et al., 1999).

Previous studies describing the coordination of walking using UCM analysis demonstrated that the performance variable sets the mediolateral trajectory of the swing foot (Krishnan et al., 2013; Rosenblatt et al., 2014; Rosenblatt et al., 2015) and the COM displacement during walking (Black et al., 2007; Qu et al., 2012; Papi et al., 2015). In these studies, the synergy index value was changed by the degree of task difficulty according to various conditions of the foot contact position (Rosenblatt et al., 2014; Rosenblatt et al., 2015). However, the change of motor variability with regards to controlling the COM displacement during lateral trunk lean gait is unknown. Lateral trunk lean gait for knee OA patients may change motor variability and thus increase task difficulty.

There are advantages of gait modifications with respect to KAM reduction; however, the lateral trunk lean gait may affect the control of the COM displacement owing to the movement strategy of the leaning trunk toward the stance side. Utilizing the UCM approach in this study, one can investigate whether the lateral trunk lean gait for knee OA patients affects the control of the COM displacement, and it may clarify the availability of the lateral trunk lean gait in terms of motor coordination related to task difficulty during walking.

This study had two purposes. The first purpose was to clarify the biomechanical mechanism to reduce KAM of lateral trunk lean gait for knee OA patients. We hypothesized that KAM of lateral trunk lean gait would decrease because the KLA would be shortened via shifting COM in the frontal plane toward the stance limb. The secondary purpose was to quantify how motor variability that controls the COM displacement during the stance phase of gait is affected when the lateral trunk lean angle was adjusted. We hypothesized that V_{UCM} would significantly increase to maintain the COM displacement in the frontal plane when knee OA patients walked to adjust the lateral trunk lean angle compared with those with normal gait.

2. Methods

2.1. Subjects

Subjects with knee pain were recruited from orthopedic clinics and through advertisements to the local residents. After recruitment, 20 subjects with radiographic medial OA were diagnosed by an experienced orthopedic surgeon. If patients had bilateral knee OA, the limb comprising the more symptomatic knee was selected for this study. Subjects were included if they reported knee pain on most days of the previous month and had tenderness in combination with osteoarthritic signs according to the Kellgren/Lawrence (K/L) classification of Grade 1 or higher over the medial tibiofemoral compartment of the knee. Subjects were observed by an orthopedist for the improvement of knee pain and the prevention of the progress of the knee OA. Exclusion criteria were as follows: patients diagnosed with any neurological disease, rheumatoid arthritis, and lower limb artificial joint replacement and those using a cane daily or who had difficulty walking without assistance. Knee OA severity was assessed according to the Kellgren/Lawrence (K/L) grading scale (Kellgren et al., 1957). The Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC) was used to assess knee pain, stiffness, and impairment in physical function (Bellamy et al., 1988). Patient characteristics are presented in Table 1. This study was approved by the Ethics Committee of Division of Physical Therapy and Occupational Therapy Sciences, Graduate School of Biomedical and Health Sciences, Hiroshima University (Approval no. 1414), and all patients provided informed consent prior to participation.

2.2. Experimental set-up and procedures

Patients walked across a 10-m laboratory walkway at a comfortable walking speed under two conditions: normal and lateral trunk lean gaits. Walking speed was assessed using two photoelectric timers (TM-02; Tamagawa Shop, Hiroshima, Japan). With regard to lateral trunk lean gait, patients were instructed to lean their trunk toward the study limb during the ipsilateral stance phase and to reach their maximum lateral trunk lean to the target angle after initial contact of the study limb. Using a real-time visual feedback system, patients were instructed to shift trunk lean displayed in real time to match a target angle of 10° (Fig. 1). If the patients could not achieve the target trunk lean angle, they were provided additional verbal feedback and encouraged to continue to try to reach the target. 10° was chosen as the target lateral trunk lean angle as previous studies have shown significant changes in KAM at this target angle, while maintaining a feasible amount of lateral trunk lean (Clark et al., 2013; Takacs et al., 2014). Trunk marker positional data were streamed from the Vicon Nexus version 2.1.1 software (VICON MX; Vicon Motion Systems, Oxford, UK) to MATLAB R2014a software (MathWorks, Natick, MA) in real time. MATLAB calculated and displayed the lateral trunk lean angle animation. Before data collection, patients practiced for approximately 10 min to achieve the target trunk lean angle. The error range for the trunk lean angle corresponded to ± 2° during the stance phase, and the practice was completed after the patients successfully achieved the target angle within the error range. For each condition, data collection required a minimum of 10 trials to ensure appropriate gait modification, and subjects took one step per trial. The 10 trials within the error range (± 2°) for the target angle were included in the analysis.

2.3. Kinematics and kinetics measurements

Infrared-reflecting markers were attached to 40 anatomical landmarks (Anan et al., 2015). Kinematic data during gait were collected using a 3D motion analysis system with six infrared cameras (VICON MX; Vicon Motion Systems, Oxford, UK). Kinetic data were collected using eight force plates (Tec Gihan, Uji, Japan) to measure GRF under each individual foot. These 3D coordinates were collected by the motion analysis system at a sampling rate of 100 frame/s, and the 3D GRF data were collected by the force plates at a sampling frequency of 1000 Hz. The stance phase was defined as when the vertical vector of GRF was >10 N (O'Connor et al., 2007). Kinematic and kinetic data were low-pass filtered using 4^{th}-order Butterworth filters (6 Hz and 20 Hz, respectively). The lateral trunk lean angle was calculated as the angle between the trunk line (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. Here the mean of the maximum value of lateral trunk lean angle of 10 trials was used. KAM was calculated using a tibial coordinate system with the origin in the knee joint center (Kito et al., 2010). Peak KAM was calculated at the points of the first KAM peak during the stance phase. KAM impulse was calculated as the timed integral of KAM for stance duration (Kito et al., 2010). Collected marker coordinates were used to define the respective local coordinate systems of a nine rigid-body-link model consisting of the head (both temple and nuchal), thorax (both acromia and superior edge of the iliac crests), pelvis (both anterior superior iliac spines and posterior superior iliac spines), both thighs (the superior aspect of the greater trochanter and medial and lateral epicondyles of the femur), both shanks (the medial tibial condyle, lateral tibial condyle, medial malleolus, and lateral malleolus), and both feet (the posterior distal aspect of the calcaneus and the head of the first and fifth metatarsals). The whole-body COM displacement was calculated using coefficients of each body segment's inertia obtained from the work of Okada et al. (1996). The knee joint center was located at the midpoint between the lateral and medial femoral epicondyles (Shull et al., 2013). The knee joint center position was defined as the distance in the frontal plane from the center between the lateral and medial malleoli (ankle center), and the medial position to the ankle center was considered as positive value. The COM and center of pressure (COP) displacements in this study were defined as the distance from the knee joint center in the frontal plane. Step width was defined as the distance between both ankle joint centers during initial double support (Favre et al., 2015). GRF was calculated as the resultant force vector of the vertical and mediolateral components (Hunt et al., 2006). KLA was calculated as the perpendicular distance between the line of action of the GRF vector and the knee joint center in the frontal plane of the shank reference frame. KAM and GRF were then normalized to each patient's body mass. The COP and COM displacements, knee joint center, KLA, and GRF vector were analyzed at the first peak KAM during the stance phase.

2.4. UCM analysis

UCM analysis was used to characterize the control of the COM displacement during the stance phase. For UCM analysis, data were time-normalized (0%–100%) from the initial contact to toe off. For analysis with regards to controlling the COM displacement in the frontal plane of the lateral trunk lean gait, the performance variable was selected as the COM displacement in the mediolateral direction. UCM analysis generated a geometric model of the performance variable. The geometric model comprised the following eight segments: stance-limb shank, stance-limb thigh, pelvis, swing-limb thigh, swing-limb shank, trunk, thorax, and head (Fig. 2). The segments i (i = 1–5) had motions outside the frontal plane as defined by angles α_1, α_2, α_3, α_4, and α_5, respectively. α_1 and α_5 represent the projection of the line 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 and α_5: swing limb). α_2 and α_4 represent the projection of the line 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 and α_4: swing limb). α_3 represents the projection of the line connecting both hip joint centers of the limb to the frontal plane, effectively incorporating both hip movements in the transverse plane. Due to the large amount of motion outside the frontal plane, the movements of the lower limbs in the sagittal plane and the pelvis in the transverse plane were included to account for changes in the effective length of the segments when projected onto the frontal plane (Krishnan et al., 2013). The geometrical model for COM delimited to the mediolateral direction in the frontal plane is described as follows:

$$\mathrm{COM} = x_0 + C_1 \times M_1 \times L_1 \times \cos\alpha_1 \sin\theta_1 + C_2 \times M_2 \times L_2 \times \cos\alpha_2 \sin\theta_2$$

$$+ C_3 \times M_3 \times L_3 \times \cos\alpha_3 \cos\theta_3 + C_4 \times M_4 \times L_4 \times \cos\alpha_4 \sin\theta_4$$

$$+ C_5 \times M_5 \times L_5 \times \cos\alpha_5 \sin\theta_5 + C_6 \times M_6 \times L_6 \times \sin\theta_6$$

$$+ C_7 \times M_7 \times L_7 \times \sin\theta_7 + C_8 \times M_8 \times L_8 \times \sin\theta_8$$

where x_0 is the segmental position of the absolute coordinate system in the mediolateral direction; $C_{\mathrm{i}\ (\mathrm{i} = 1\text{–}8)}$ is the estimated position of $\mathrm{COM_i}$ on the segment; M_i is the proportion of the total body mass of each segment; L_i is the length of the segment; θ_i are the segment angles relative to the frontal plane; and α_3 is the segment angle relative to the transverse plane.

A linearization approximation of the geometric model of the performance variable 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 corresponding to changes in the performance variable with respect to each of the segmental angles (the elemental variables) (Scholz et al., 1999). ε, the null space of **J**, 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 $d = 1$. Every percentage of each stance ($\theta - \bar{\theta}$) was projected onto the null space:

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

and onto a component orthogonal to this subspace:

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

Consider 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:

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

The variance that affected bad variance was calculated as follows:

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

The UCM analysis was calculated using the whole-body COM in the frontal planes. The average total variance in the segmental configuration space per total DOFs was calculated using V_{TOT}:

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

The strength of synergy is reflected by the synergy index (ΔV), and ΔV was calculated as follows (Krishnan et al., 2013):

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

The more positive ΔV is, the stronger the synergy. Non-positive values indicate the absence of synergy. ΔV ranges from −14 (all variance is partitioned into V_{ORT}) to 14/12 (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. These variables do not follow a normal distribution. To address this and to apply statistical analysis, the ΔV was log-transformed using Fisher's z-transformation (Robert et al., 2009):

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

Prior to statistical analysis, V_{UCM}, V_{ORT}, V_{TOT}, ΔVz, and COM variability were averaged across the first half (0–50%) and latter half (51%–100%) of the stance phase.

2.5. Statistical analysis

The normality of the data distributions was assessed using the Shapiro–Wilk test. To compare the difference between conditions, a t-test was performed for KAM-related biomechanical parameters. The peak KAM was analyzed using analysis of covariance (ANCOVA), with walking speed as a covariate because it may affect the peak KAM (Zeni et al., 2009; Gerbrands et al., 2017). Regarding the synergy index, a mixed design ANOVA was performed and included a within-subject factor of the variance component (V_{UCM} and V_{ORT}) and between-subject factor of condition (normal and lateral trunk lean gait). A significant main effect of the variance component ($V_{UCM} > V_{ORT}$) indicated the existence of synergy. Further, paired t-tests were used to compare ΔVz, V_{UCM}, and V_{ORT} under the two conditions. All statistical analyses were performed using IBM SPSS version 22.0 for Windows (IBM Japan, Tokyo, Japan), with significance set at $p < 0.05$.

3. Results

The lateral trunk lean angle was 3.0 ± 2.0° under the normal gait condition and 11.1 ± 1.9° under the lateral trunk lean gait condition. The gait parameters are shown in Table 2. Walking speed and stance time under the lateral trunk lean gait condition were significantly distinct compared with the normal gait condition ($p < 0.001$ and $p < 0.001$, respectively). The step width under the lateral trunk lean gait condition was significantly larger than that under the normal gait condition ($p < 0.001$). The peak KAM and KAM impulse under the lateral trunk lean gait condition significantly decreased compared to the normal gait condition ($p < 0.001$ and $p < 0.001$, respectively) (Table 3). After adjusting for walking speed, the peak KAM under trunk lean gait condition significantly decreased compared with the normal gait condition ($p < 0.05$). The data related to KAM are shown in Table 4. The KLA under the lateral trunk lean gait condition was significantly shorter than that under the normal gait condition ($p < 0.01$). The GRF vector magnitude did not significantly differ between conditions. The knee joint center position the lateral trunk lean gait condition significantly shifted medially than that under the normal gait condition ($p < 0.001$). The COP displacement under the lateral trunk lean gait condition was significantly shorter than that under the normal gait condition ($p < 0.001$). The COM displacement did not significantly differ between the conditions.

COM variability during the first half of the stance phase under the lateral trunk lean gait condition was significantly larger than that under the normal gait condition ($p < 0.05$) (Fig. 3).

Our ANOVA analysis showed a significant main effect of variance component ($V_{UCM} > V_{ORT}$), indicating the presence of kinematic synergy ($p < 0.01$). ΔVz did not significantly differ between conditions (Fig. 4). V_{UCM} under the lateral trunk lean gait condition was significantly larger than that under the normal gait condition (first half: $p < 0.05$ and latter half: $p < 0.05$). V_{ORT} did not significantly differ between conditions (Fig. 5). V_{TOT} of the lateral trunk lean gait condition was significantly larger than that of the normal gait condition (first half: $p < 0.01$ and latter half: $p < 0.01$) (Fig. 6).

4. Discussion

We aimed to determine the biomechanical mechanism of KAM reduction during the lateral trunk lean gait in knee OA patients. Contrary to our hypothesis, the COM displacement did not significantly differ between the conditions. We speculated that the COM displacement relative to the knee joint center did not significantly differ according to the movement strategy of shifting the knee joint position to the opposite direction of leaning trunk toward the stance side. Therefore, the medial shift of the knee joint center may shorten the distance of the COP displacement and KLA with respect to the knee joint center, and the motion may have led to the KAM reduction. Favre et al. (2016) reported that the instruction to modify trunk sway affected the step width. In this study, the step width under the lateral trunk lean gait condition was significantly larger than that under the normal gait condition.

Thus, the increase in the step width may be related to the medial shift of the knee joint center. Our finding will help clinicians to further understand the mechanisms that reduce the KAM in the lateral trunk lean gait for knee OA patients.

There was no significant difference in kinematic synergy between the conditions, despite increased V_{UCM} and V_{TOT}. Rosenblatt et al. (2015) reported that an increased synergy index was shown with a decreased V_{ORT}. Other studies that investigated upper extremity of older adults demonstrated that V_{ORT} of older adults was larger than that of younger adults, resulting in a reduced synergy index (Verrel et al., 2012; Kapur et al., 2010). The change in V_{ORT} may influence the synergy index magnitude. V_{ORT} in this study did not significantly differ between conditions, thus possibly leading to a similar synergy index.

Walking in daily living is repeatedly performed; therefore, it is necessary to assess whether knee OA patients can stably perform lateral trunk lean gait across repetitions. UCM analysis is described as a method of analysis through which hypotheses regarding controlled and uncontrolled degrees of freedom of movements can be tested (Scholz et al., 1999). In this analysis, the trail-to-trial variability can be used to assess stability and control the means to stabilization so that the lack of control implies reduced stability. Hsu et al. (2013) reported that older adults changed their joint coordination pattern to control the COM during balance recovery and had a lower synergy index with increased V_{ORT}, suggesting that UCM analysis can be used to detect poor balance coordination in the elderly. In this study, our results showed that knee OA patients can perform lateral trunk lean gait by increasing V_{UCM} (without increasing V_{ORT}), while synergy index did not change. Rosenblatt et al. (2014) reported that increasing good variability may indicate improving the stability of gait patterns. Our findings indicate that lateral trunk lean gait was possible without changing the coordination of each joint movement to control COM displacement by utilizing good variability to accomplish stable task, although the COM variability increased. Thus, our results suggested that knee OA patients stably perform lateral trunk lean gait across repetitions.

This study had a limitation related to differing gait speeds. Considering the findings that slower walking leads to greater gait variability (Kang et al., 2008), slower gait speed under lateral trunk lean gait may affect gait variability. If so, it is expected that both V_{UCM} and V_{ORT} increases under lateral trunk lean gait. In agreement with this view, Domkin et al. (2002) reported that an improvement in movement speeds after practicing a bimanual pointing task was accompanied by decreases in both V_{UCM} and V_{ORT}. However, in our study, knee OA patients adapted movement strategy to achieve task by increasing V_{UCM} in terms of trunk lean gait because V_{ORT} did not significantly differ between the conditions. Therefore, we believe that UCM analysis in this study could assess the characteristics of synergy in terms of trunk lean gait, although a possibility that slower gait speed modifies the synergy cannot be excluded. Our results showed only a temporary change in the kinematic and variability data of trunk lean gait using a real-time visual feedback system. Thus, future research should investigate thorough longitudinal studies to determine the change via motor learning.

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Conflict of Interest Statement

The authors declare no conflicts of interest.

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Figure Legends

Fig. 1. The real-time movement visual feedback system used for training and gait with altered lateral trunk lean angle is shown. In order to reach the target angle, subjects walked toward a projection screen, which displayed the lateral trunk lean angle toward the stance limb in real time. On the screen, the thick arrow represents the target angle and the thin arrow represents the lateral trunk lean angle in real time.

Fig. 2. A geometrical model was used to extract an analytical expression for each elemental variable matrix. The left, middle, and right illustrations represent views in the frontal, sagittal, and transverse planes, respectively.

Fig. 3. COM variability during the stance phase. **A** represents the first half during the stance phase. **B** represents the latter half during the stance phase. Data represent the means and standard deviations of both conditions. COM variability under the trunk lean gait condition was larger than that under the normal gait condition (**A**, $p < 0.05$, Normal gait = 49.9 ± 25.3 mm, Lateral trunk lean gait = 69.9 ± 32.7 mm; **B**, $p = 0.19$, Normal gait = 57.7 ± 28.6 mm, Lateral trunk lean gait = 77.2 ± 54.3 mm).

Fig. 4. ΔVz during the stance phase. **A** represents the first half during the stance phase. **B** represents the latter half during the stance phase. Data represent the means and standard deviations of both conditions. ΔVz did not significantly differ between the conditions (**A**, $p = 0.66$, Normal gait = 1.6 ± 0.2, Lateral trunk lean gait = 1.6 ± 0.3; **B**, $p = 0.11$, Normal gait = 1.5 ± 0.3, Lateral trunk lean gait = 1.6 ± 0.2).

Fig. 5. V_{UCM} and V_{ORT} during the stance phase. **A** represents the first half during the stance phase. **B** represents the latter half during the stance phase. Data represent the means and standard deviations of both conditions. V_{UCM} under the lateral trunk lean gait condition during stance phase was larger than that under the normal gait condition (**A**, V_{UCM}: $p < 0.05$, Normal gait = $4.1 \times 10^{-4} \pm 2.6 \times 10^{-4}$ rad^2, Lateral trunk lean gait = $6.4 \times 10^{-4} \pm 4.7 \times 10^{-4}$ rad^2; V_{ORT}: $p = 0.06$, Normal gait = $2.0 \times 10^{-4} \pm 1.1 \times 10^{-4}$ rad^2, Lateral trunk lean gait = $2.7 \times 10^{-4} \pm 1.5 \times 10^{-4}$ rad^2; **B**, V_{UCM}: $p < 0.05$, Normal gait = $4.0 \times 10^{-4} \pm 2.3 \times 10^{-4}$ rad^2, Lateral trunk lean gait = $6.6 \times 10^{-4} \pm 4.9 \times 10^{-4}$ rad^2; V_{ORT}: $p = 0.26$, Normal gait = $2.6 \times 10^{-4} \pm 1.3 \times 10^{-4}$ rad^2, Lateral trunk lean gait = $3.3 \times 10^{-4} \pm 2.8 \times 10^{-4}$ rad^2).

Fig. 6. V_{TOT} during the stance phase. **A** represents the first half during the stance phase. **B** represents the latter half during the stance phase. Data represent the means and standard deviations of both conditions. V_{TOT} under the lateral trunk lean gait condition was larger than that under the normal gait condition (**A**, $p < 0.05$, Normal gait = $3.7 \times 10^{-4} \pm 2.2 \times 10^{-4}$ rad^2, Lateral trunk lean gait = $5.6 \times 10^{-4} \pm 4.1 \times 10^{-4}$ rad^2; **B**, $p < 0.05$, Normal gait = $3.6 \times 10^{-4} \pm 2.0 \times 10^{-4}$ rad^2, Lateral trunk lean gait = $5.9 \times 10^{-4} \pm 4.4 \times 10^{-4}$ rad^2).

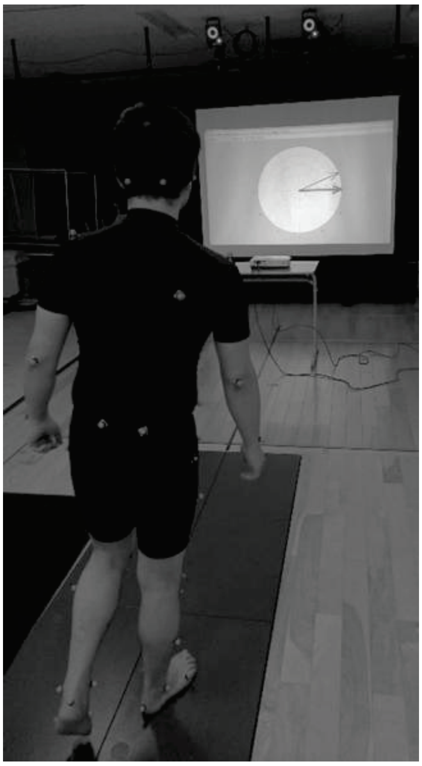
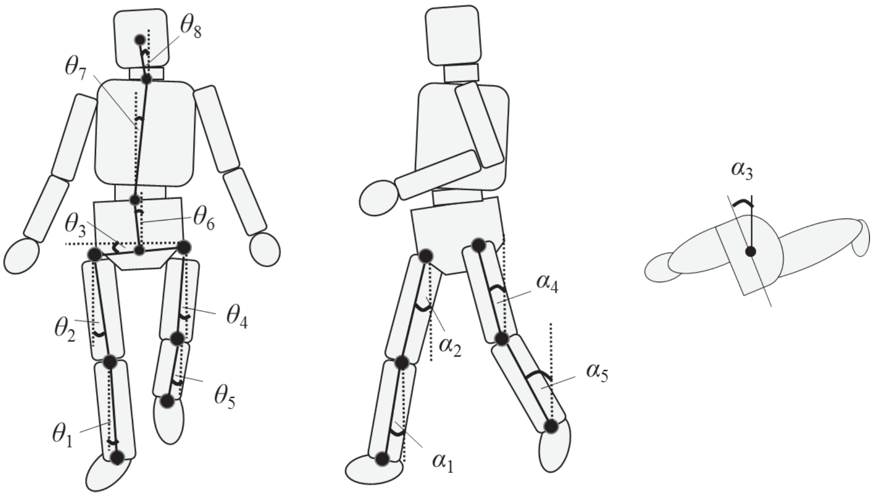


Fig. 1



θ_8
θ_7
θ_6
θ_3
θ_2
θ_4
θ_5
θ_1
α_4
α_2
α_5
α_1
α_3

Fig. 2

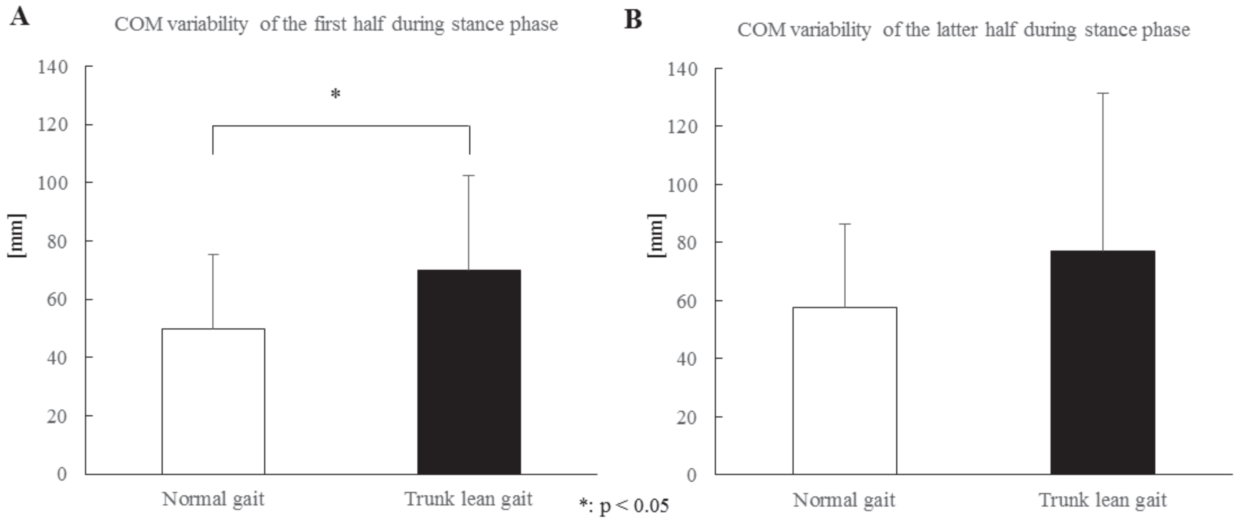

A
COM variability of the first half during stance phase
140
120
100
80
60
40
20
0
[mm]
*
Normal gait
Trunk lean gait
*: p < 0.05
B
COM variability of the latter half during stance phase
140
120
100
80
60
40
20
0
[mm]
Normal gait
Trunk lean gait

Fig. 3

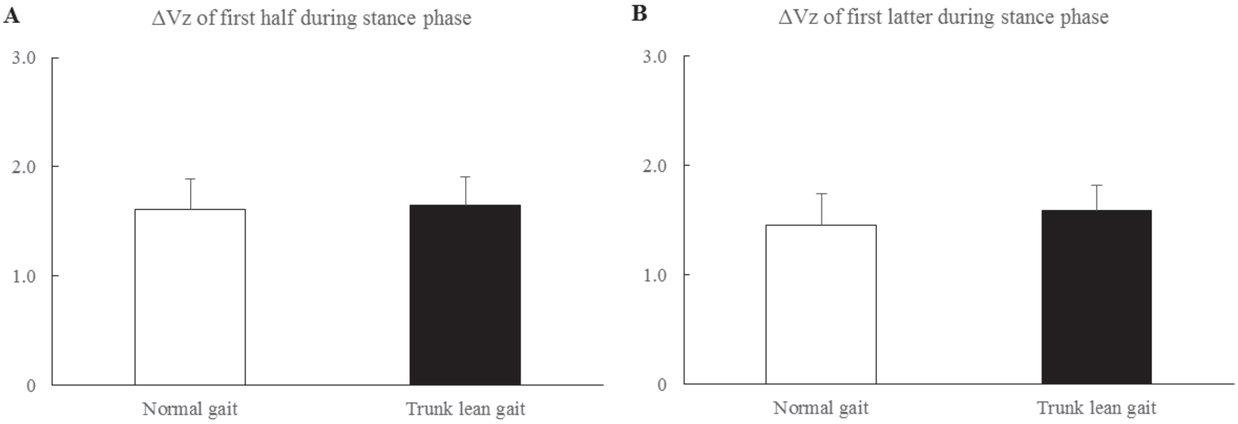
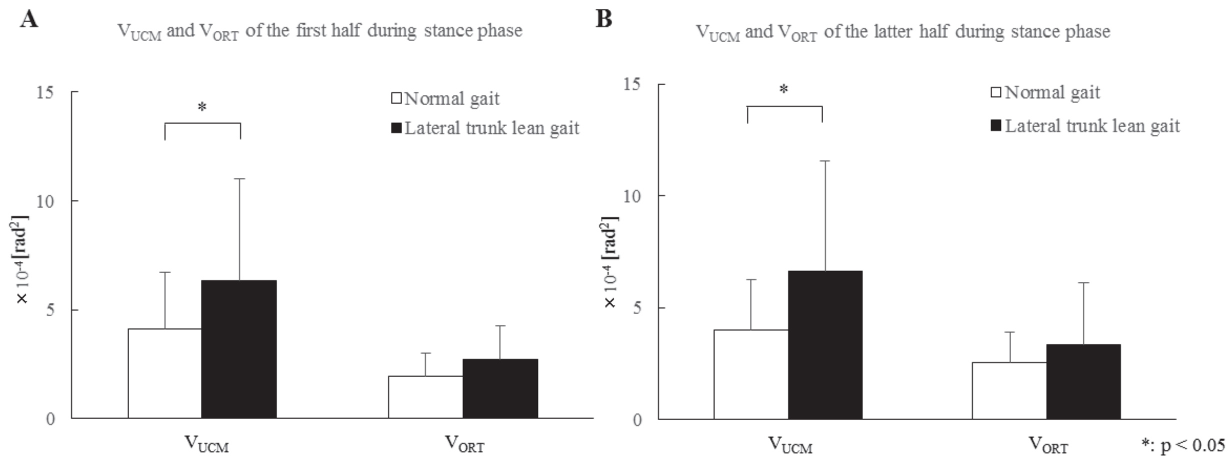

A
ΔVz of first half during stance phase
3.0
2.0
1.0
0
Normal gait
Trunk lean gait
B
ΔVz of first latter during stance phase
3.0
2.0
1.0
0
Normal gait
Trunk lean gait

Fig. 4



A
VUCM and VORT of the first half during stance phase
15
10
5
0
×10-4[rad2]
*
Normal gait
Lateral trunk lean gait
VUCM
VORT
B
VUCM and VORT of the latter half during stance phase
15
10
5
0
×10-4[rad2]
*
Normal gait
Lateral trunk lean gait
VUCM
VORT
*: p < 0.05

Fig. 5

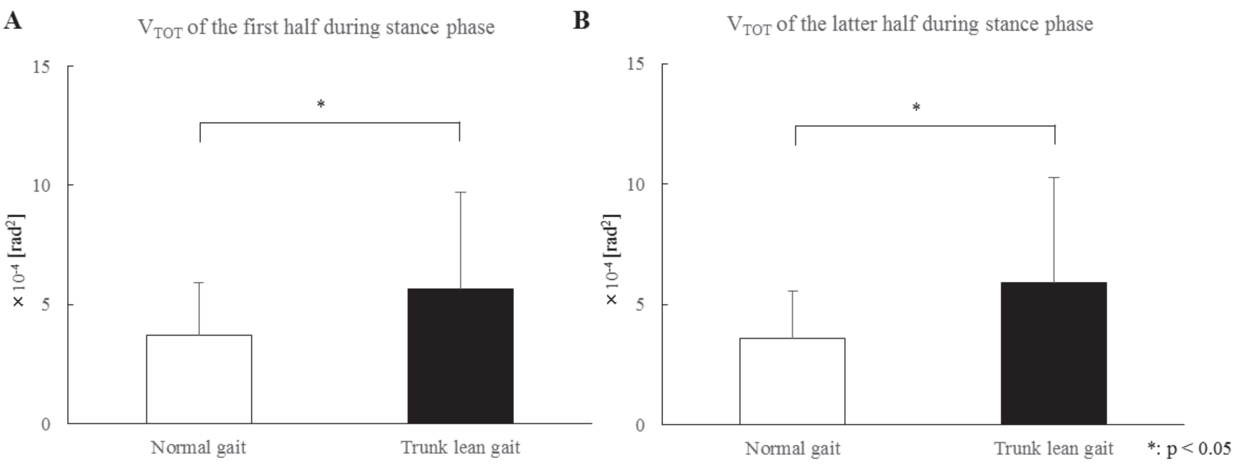

A
V_TOT of the first half during stance phase
15
10
5
0
× 10-4 [rad2]
*
Normal gait
Trunk lean gait
B
V_TOT of the latter half during stance phase
15
10
5
0
× 10-4 [rad2]
*
Normal gait
Trunk lean gait
*: p < 0.05

Fig. 6

Table

Table 1. Patient characteristics

Characteristics	Mean ± SD or n
Age (y)	72.1 ± 4.6
Height (m)	1.54 ± 0.08
Body mass (kg)	56.6 ± 6.7
BMI (kg/m^2)	24.0 ± 2.4
Gender (n)	
Female	14
Male	6
K/L grade	
Grade 1/2/3/4 (n)	10/3/4/3
WOMAC score	
Pain (0–20)	2.5 ± 2.7
Stiffness (0–8)	0.9 ± 1.5
Physical function (0–68)	7.6 ± 6.6

BMI: body mass index; K/L: Kellgren/Lawrence; WOMAC: Western Ontario and McMaster Universities Osteoarthritis Index; SD: standard deviation

Table 2. Gait parameters under normal and trunk lean gait conditions

	Normal gait (mean ± SD)	Trunk lean gait (mean ± SD)	P-value
Walking speed (m/s)	1.14 ± 0.21	0.76 ± 0.24	<0.001
Stance time (s)	0.63 ± 0.06	0.79 ± 0.12	<0.001
Step width (mm)	121.6 ± 33.0	171.0 ± 39.2	<0.001

SD: standard deviation

Table 3. KAM data under normal and trunk lean gait conditions

	Normal gait (mean ± SD)	Trunk lean gait (mean ± SD)	P-value
Peak KAM (N·m/kg)	0.56 ± 0.21	0.41 ± 0.15	<0.001
KAM impulse (N·m/kg·s)	0.19 ± 0.06	0.16 ± 0.06	<0.001

KAM: knee adduction moment; SD: standard deviation

Table 4. Data related to KAM under normal and trunk lean gait conditions at the first peak KAM during the stance phase

	At the first peak KAM		
	Normal gait (mean ± SD)	Trunk lean gait (mean ± SD)	P-value
Lever arm along the ML axis (mm)	37.7 ± 27.9	21.7 ± 23.2	<0.01
GRF vector magnitude (N/kg)	9.56 ± 1.77	9.81 ± 1.36	0.24
Knee joint center position along the ML axis (mm)	−22.4 ± 17.1	−9.7 ± 15.1	<0.001
COP displacement along the ML axis (mm)	27.2 ± 19.0	10.8 ± 14.6	<0.001
COM displacement along the ML axis (mm)	83.3 ± 12.5	84.6 ± 14.7	0.55

KAM: knee adduction moment; GRF: ground reaction force; COM: center of mass; COP: center of pressure; SD: standard deviation

Conflict of Interest Statement

The authors declare no conflicts of interest.