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RESEARCH ARTICLE

Spinal Networks and Spinal Cord Injury: A Tribute to Reggie Edgerton

Interjoint coordination between the ankle and hip joints during quiet standing in individuals with motor incomplete spinal cord injury

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Abstract

Individuals with motor incomplete spinal cord injuries (iSCI) often have impaired abilities to maintain upright balance. For able-bodied (AB) individuals, the ankle and hip joint accelerations are in antiphase to minimize the postural sway during quiet standing. Here we investigated how interjoint coordination between the ankle and hip joints was affected in individuals with iSCI, leading to their larger postural sway during quiet standing. Data from 16 individuals with iSCI, 14 age- and sex-matched AB individuals, and 13 young AB individuals were analyzed. The participants performed quiet standing during which kinematic and kinetic data were recorded. Postural sway was quantified using center-of-pressure velocity and center-of-mass acceleration. Individual ankle and hip joint kinematics were quantified, and the interjoint coordination was assessed using the cancellation index (CI), goal-equivalent variance (GEV), nongoal-equivalent variance (NGEV), and uncontrolled manifold (UCM) ratio. Individuals with iSCI displayed greater postural sway compared with AB individuals. The contribution of ankle angular acceleration toward one's sway was significantly greater for those with iSCI compared with AB groups. CI and the UCM ratios were not statistically different between the groups, while GEV and NGEV were significantly greater for the iSCI group compared with the AB groups. We demonstrated that individuals with iSCI show larger postural sway compared with the AB individuals during quiet standing, primarily due to larger ankle joint acceleration. We also demonstrated that the interjoint coordination between ankle and hip joint is not affected in individuals with iSCI, which is not successfully able to reduce the large COM acceleration.

NEW & NOTEWORTHY There are limited studies investigating the biomechanics of standing balance for individuals with motor incomplete spinal cord injury (iSCI). Through our study, we found that these individuals with iSCI demonstrated increased postural sway primarily due to increased ankle joint accelerations. In addition, the ankle-hip coordination was equivalent between able-bodied individuals and those with motor incomplete spinal cord injury, which was not able to reduce the large body acceleration.

hip strategy; human biomechanics; incomplete spinal cord injury; interjoint coordination; quiet standing

INTRODUCTION

Motor incomplete spinal cord injury (iSCI) may cause sensory and motor impairment below the level of injury and often affects the control of the lower extremities, resulting in decreased ability to regulate posture during tasks such as standing and walking. The frequency of falls in this population is often greater compared with the older adult population or individuals with other neurological disorders, such as stroke or Parkinson's disease (1). Although there have been a number of clinical studies investigating upright balance in individuals with iSCI (1–3), only a few studies investigated the biomechanics during quiet standing, in this population. For example, Lemay et al. (4) investigated quiet standing using center-of-pressure (COP) based measures and demonstrated that individuals with iSCI showed larger COP velocity during quiet standing while relying primarily on visual information. Further investigation on the kinematics of quiet standing would help

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us to better understand impaired standing balance in individuals with iSCI and may guide how to improve their balance.

The dynamics of the human body during quiet standing is often approximated as a single-link inverted pendulum model, where the body pivots only about the ankle joint (5–8). However, other studies adopted more realistic model using a double-link inverted pendulum model, where the hip and ankle joints act as the two pivot axes (9–14). With the use of this model, coordinated activities by the ankle and hip joints, i.e., the interjoint coordination, have been investigated. For example, previous literature demonstrated that the antiphase relationship between the two joint angular accelerations reduces the whole body translational COM acceleration (9, 12). The deterioration of this antiphase interjoint coordination, quantified using the cancellation index (CI), in the older adult population resulted in larger COM acceleration (12). The uncontrolled manifold (UCM) analysis has been also used to examine the coordinated activity between multiple joints during quiet standing (11, 15–17). In their analyses, the joint configurations distributed along the UCM subspace, quantified using the goal-equivalent variance (GEV), indicated coordinated joint actions that do not affect the variance of the COM kinematics. Contrarily, the joint configurations distributed along the orthogonal subspace, quantified using the nongoal-equivalent variance (NGEV), indicated coordinated joint action affecting the variance of the COM kinematics (11, 15–17). The ratio of GEV to NGEV (UCM ratio) can represent the coordinated joint kinematics to maintain stable posture. Hsu et al. (15) demonstrated that the UCM ratio is lower for older adults compared with younger adults during quiet standing, since the older adults exhibited larger NGEV compared with the younger adults while maintaining similar GEV.

To date, there is no study that has investigated how the interjoint coordination during quiet standing is affected following iSCI. Here we investigated the ankle-hip joint coordination during quiet standing in individuals with iSCI using the CI and the UCM analysis. In this study, we quantified the postural sway using the COM acceleration and the COP velocity. Since individuals with iSCI experience motor and/or sensory deficits and display increased postural sway (4), we hypothesized that the interjoint coordination is impaired in individuals with iSCI resulting in the increase of postural sway.

METHODS

Participants

Data used in this study were common to another study published by Chan et al. (18, 19). Fourteen young able-bodied (AB) individuals (Y-group), 15 age- and sex-matched AB individuals (M-group), and 21 individuals with iSCI (S-group) participated in the project by Unger et al. (20) and Chan et al. (18, 19). Recruited AB individuals were free from any known health issues that affected standing and walking. The age-matched participants were ±3 yr of age from the corresponding participants with iSCI. We recruited individuals with iSCI who were able to stand independently for 30 s without any gait aid. Subsequently, individuals with iSCI underwent two baseline assessments, separated by 2 wk. Here, we primarily used the data during their first baseline assessment, except for two individuals for whom we used the data from the second baseline assessment due to technical issues with motion capture data during the first baseline assessment. The demographics and clinical- and self-reported scores of the recruited individuals with iSCI are summarized in Tables 1 and 2, respectively (for detailed testing procedures see Refs. 18, 20). The recruited participants had relatively high motor functions in terms of mobility while there were considerable variations in their functions shown in their clinical scores (refer to Table 2). One individual from each of the young- and matched-AB groups were excluded from analysis due to technical issues during data collection, and five individuals among the S-group were excluded from the analysis because motion capture data were not recorded for three participants and because two participants were not able to complete the standing task independently. Therefore, in this study, we used the data from 13 young AB (Y-group; mean age 24.5 ± 5.1 yr; 7 male and 6 female; mean mass: 62.0 ± 11.6 kg; mean height: 173 ± 12 cm), 14 age- and sex-matched AB individuals (M-group; mean age 56.1 ± 12.9 yr; 4 male and 10 female; mean mass: 57.4 ± 10.9 kg; mean height: 164 ± 8 cm), and 16 individuals with iSCI (S-group; mean age 56.9 ± 15.9 yr; 4 male and 12 female; mean mass: 71.8 ± 18.6 kg; mean height: 166 ± 12 cm). Written informed consent was obtained from all participants, and the experimental procedure was approved by the Research Ethics Boards of the University Health Network and University of Toronto.

Experimental Procedure

The participants were asked to perform two 150-s trials of quiet standing on a force platform (AccuSway Dual Top, Advanced Mechanical Technology, Inc., Watertown, MA) with their eyes open (EO) and closed (EC). The participants stood with their heels 17 cm apart and 14° between left and right feet with their arms across their chest (21). All participants wore a slack safety harness that was connected to the ceiling to prevent participants from falling.

A motion capture system with six infrared cameras (Raptor-E system and Cortex 3.1, Motion Analysis Corp., Rohnert Park, CA) was used to record three-dimensional locations of the reflective markers at the sampling frequency of 200 Hz. The markers from right medial and lateral malleoli, greater trochanter, and glenohumeral joints were used for the subsequent analyses. The force platform was used to measure ground reaction force and moment components at the sampling frequency of 2,000 Hz.

Analyses

MATLAB (MATLAB 2014a, MathWorks, Natick, MA) was used for the offline analyses. The first and the last 15 s of data were removed to avoid any potential transients. In this study, all analyses focused on the movement in the anterior-posterior direction as the postural sway amplitude is greater in this direction compared with the medial-lateral direction during quiet standing (8). The motion capture and the force plate signals were filtered using fourth-ordered, zero phase lag, low-pass Butterworth filter with a cut-off frequency of 4 Hz (7).

COP and whole body COM displacement.

The COP was calculated from the filtered force plate signals. The body dynamic was modeled using a double-

Table 1. *Demographics of the recruited individuals with motor incomplete spinal cord injury*

Subject	Sex	Age, yr	Mass, kg	Height, cm	Injury Level†	Time since Injury, yr	Mechanism of Injury
PBT05	F	32	49.9	171	C4	3.5	Stenosis
PBT06	M	70	88.2	185	T1	1.8	Osteomyelitis
PBT08	M	60	109.2	188	C5	3.2	Traumatic
PBT10	F	43	47.3	154	T6	3.9	Surgery, Meningioma
PBT12	F	87	61.1	168	T4	2.6	Meningioma
PBT13	F	57	102.0	165	C2	2.9	Transverse myelitis
PBT14	F	59	62.3	150	C1	1.1	Traumatic
PBT16	F	55	47.5	155	C5	9.1	Traumatic
PBT17	F	38	55.6	155	T4	1.3	AVM
PBT18	F	54	83.3	169	C4	13	Traumatic
PBT20	F	56	73.7	148	L5	1.2	Surgery
PBT21	F	69	60.2	174	T4	4.8	Virus
PBT22	M	88	77.2	179	C6	5.3	Blood clot
PBT23	F	38	79.9	166	T11	6.8	Traumatic
PBT24*	M	51	81.9	176	C3	7.9	Traumatic
PBT25	F	53	68.9	158	C4	39	Traumatic
Means (SD)	4 M 12 F	56.9 (15.9)	71.8 (18.6)	166 (12)	NA	6.71 (9.21)	NA
PBT01‡	F	61	51.3	NA	C3	1.0	Surgery
PBT02‡	M	64	115.9	NA	T6	6.8	Staph Infection
PBT04‡	F	54	69.4	NA	T10	1.0	Surgery
PBT15‡	M	49	46.2	169	T5	21.0	Tumor
PBT15‡	M	56	69.1	158	L1	16.3	Virus

AVM, arteriovenous malformation; C, cervical; F, female; L, lumbar; M, male; NA, not applicable; T, thoracic. *Participant whose eyes closed quiet standing task was excluded data analysis. †Neurological level of injury. ‡Participants that were excluded from data analysis.

link inverted pendulum, pivoting about the ankle and hip joints. The ankle angle (θ_A) was defined as the angle between the ankle-hip segment and the vertical axis, and the hip angle (θ_H) was defined as the angle between the ankle-hip segment and the hip-shoulder segment (Fig. 1*A*). Assuming that the joint angles were small during quiet stance, the anterior-posterior component of the whole body COM was estimated as the weighted sum of the joint angles shown by the following equation (9, 12):

$$COM_{AP} = K_A\theta_A + K_H\theta_H, \text{where } K_A = \frac{m_A r_A + m_H l_A + m_H r_H}{m_A + m_H} \text{ and } K_H = \frac{m_H r_H}{m_A + m_H}. \tag{1}$$

where COM_{AP} is the anterior-posterior component of the COM and K_A and K_H are the relative weights of ankle and hip

Table 2. *Clinical scores and self-reported scores for the recruited individuals with motor incomplete spinal cord injury*

	Clinical Scores				Self-Reported Scales				
Subject	Mini-BES test score (/28)	LE MMT score (/120)	CB&M scale (/96)	Gait speed without aid, m/s	FES-I scale (/64)	ABC scale, %	Walking aid used for gait assessment	Fall history†	Fear of falling
PBT05	25	87.5	89	1.29	31	70.6	None	1	N
PBT06§	19	104.5	42	1.24	39	60.3	Canes	0	N
PBT08	25	115	70	1.28	31	70	None	0	N
PBT10	24	104.5	78	1.1	39	65	None	1	Y
PBT12	19	103	34	0.96	21	96.3	Canes	0	N
PBT13	21	89	29	0.72	48	53.8	Canes	0	Y
PBT14	4	75	3	0.43	36	48.4	4WW	0	Y
PBT16	25	90	26	0.88	36	68.1	Canes	1	Y
PBT17	5	75	NA	0.75	25	70	4WW	1	N
PBT18	13	78.5	27	0.91	55	31.3	Canes	0	Y
PBT20	17	70	33	0.94	62	30	None	0	Y
PBT21§	3	72	NA	0.42	44	36.3	4WW	0	N
PBT22	12	81.5	20	0.83	26	81.2	None	0	N
PBT23	22	101.5	63	1.03	29	56.3	None	1	Y
PBT24*	15	97	52	1.29	34	51.3	Poles or 4WW	0	Y
PBT25	15	89.5	33	0.95	37	49.4	Poles	1	Y
Means (SD)	16.5 (7.5)	89.6 (13.6)	37.4 (27.0)	0.94 (0.27)	37.1 (10.9)	58.6 (17.9)	NA	NA	9Y 7N
PBT01‡	10	82.5	5	0.67	42	50.6	4WW	1	Y
PBT02‡	1	76	NA	0.16	26	52.5	4WW	0	N
PBT04‡	1	79.5	NA	0.38	44	52.2	4WW	0	N
PBT15‡	2	75	2	0.34	35	48.1	Canes	12	N
PBT15‡	0	66	NA	0.2	53	66.9	2WW	0	Y

ABC, Activity-specific Balance Confidence scale; CB&M, Community Balance & Mobility; FES-I, Falls Efficacy Scale International; LE MMT, lower-extreminity manual muscle testing; Mini-BESTest, Mini-Balance Evaluation Systems Test; 2WW, 2-wheeled walker; 4WW, 4-wheeled walker. *Participant whose eyes closed quiet standing task was excluded data analysis. †Retrospective falls in the previous 3 mo. ‡Participants that were excluded from data analysis. §Participant whose clinical data was from the second baseline assessment.

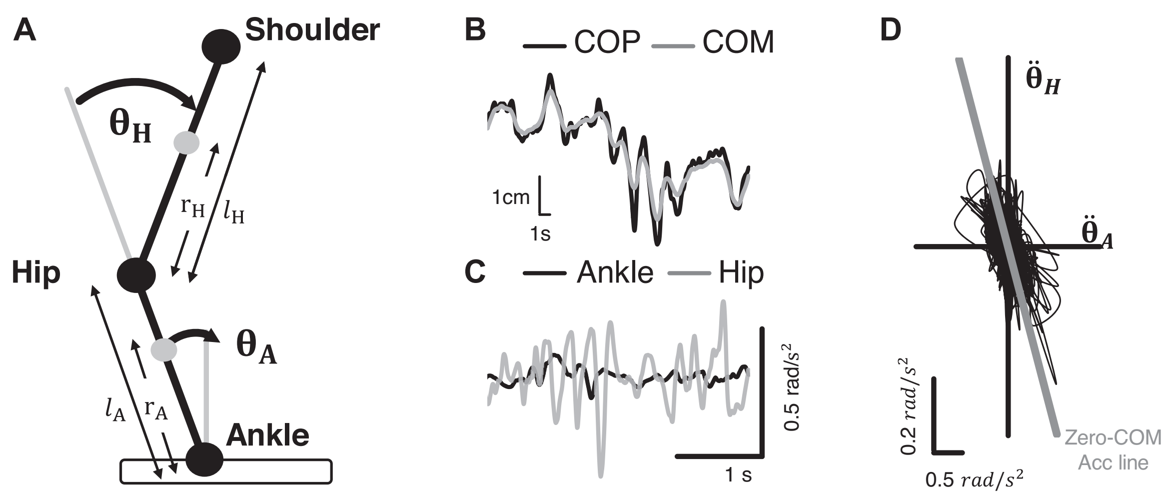

Figure 1. *A*: double-link inverted pendulum model of human body during quiet standing and with the definition of the ankle and hip joint angles (θ_A and θ_H), segment length (*l*), and the length between the joint and the segment center-of-mass (*r*). Positive joint angles are defined as a clockwise rotation. *B*: example time series of the center-of-mass (COM; gray line) and center-of-pressure (COP; black line) from the quiet standing task with eyes open. *C*: example time series of the ankle (gray line) and hip (black line) joint acceleration from the quiet standing task with eyes open. *D*: distribution of the ankle and hip joint acceleration and the zero-COM-acceleration line (gray line) derived from the double-link inverted pendulum model.

defined by the standard anthropometric data (12, 22). The segment masses, lengths, and distances to the segment COMs are defined as m, l, and r, respectively (Fig. 1*A*). The calculated COP and COM displacements nearly coincided, with faster oscillation of COP around the COM data (Fig. 1*B*), which aligns with previous findings (7, 8).

Postural and joint sway measures.

Standard deviation of the COP velocity and COM acceleration were used to quantify postural sway. COP velocity was calculated by numerically differentiating the COP displacement data, and COM acceleration was calculated by dividing the filtered horizontal ground reaction force by the participant's mass. Horizontal force was filtered separately from the filter described above but instead using a fourth-order band-pass Butterworth filter with the cut-off frequencies of 0.15 Hz and 4 Hz. The Romberg ratio was calculated for the postural sway measures by calculating the ratio of the outcome measures during EC and EO (4, 23). A ratio >1.0 indicates a greater amount of movement during the EC condition.

The amount of joint sway was quantified using the standard deviation of the ankle and hip joint accelerations. Joint angular accelerations were calculated by numerically differentiating the joint angles. In addition, the standard deviation of the weighted angular accelerations (i.e., $K_A\ddot{\theta}_A$ and $K_H\ddot{\theta}_H$) were quantified to compare the relative contribution of each joint angular acceleration toward the whole body COM acceleration.

Quantification of the ankle-hip joint coordination.

The ankle and hip joint angular accelerations show anti-phase action (Fig. 1*C*), indicated by a negatively correlated relationship on the joint angular acceleration plane (Fig. 1*D*) (9, 12). We analyzed this interjoint coordination using CI and UCM analysis. In addition, correlation between the interjoint coordination measures and postural sway measures were calculated to investigate whether changes in interjoint coordination is associated with changes in postural sway.

Cancellation index. From *Eq. 1*, COM acceleration is given as:

$$C\ddot{O}M = K_A\ddot{\theta}_A + K_H\ddot{\theta}_H \tag{2}$$

According to *Eq. 2*, when the ankle and hip joint acceleration ratio equals the specific ratio of $-K_A/K_H$, the COM acceleration is zero, which composes the zero-COM acceleration line (Fig. 1*D*) (9). The deviation from this hypothetical relation can be quantified as the CI (12):

$$CI = \frac{\sqrt{K_A^2 var(\ddot{\theta}_A) + K_H^2 var(\ddot{\theta}_H)}}{\sqrt{K_A^2 var(\ddot{\theta}_A) + K_H^2 var(\ddot{\theta}_H) + 2K_AK_H cov(\ddot{\theta}_A, \ddot{\theta}_H)}}. \tag{3}$$

The CI is greater or equal to 1, where CI = 1 indicates absence of coordinated behavior, CI > 1 indicates reduced anti-phase relationship between the joint accelerations, and CI < 1 indicates an in-phase relationship. We compared the CI among the three groups.

Uncontrolled manifold analysis. The UCM analysis, as established by Scholz and Schöner (24), was used to divide the distribution on the ankle-hip joint acceleration coordinate into two components, i.e., the UCM subspace and the orthogonal subspace. The UCM subspace was quantified as the variance along the zero-COM acceleration line, i.e., GEV. The orthogonal subspace was quantified as the variance orthogonal to the zero-COM acceleration line, i.e., NGEV. Finally, the ratio between GEV and NGEV, termed UCM ratio, was previously used to quantify the degree of interjoint coordination during quiet standing (11, 15–17). We calculated the GEV, NGEV, and UCM ratio and compared each among the three groups. The two approaches examine the interjoint coordination differently: CI examines the degree of interjoint coordination by quantifying the covariance between ankle and hip joint accelerations regardless of the zero-COM acceleration line, while UCM analysis examines how well interjoint coordination align to the zero-COM acceleration line and reduce the overall COM acceleration by quantifying the variance along and orthogonal to the zero-COM-acceleration line.

Statistical Analysis

IBM SPSS Statistics 26 (IBM, Natick, MA) was used to perform statistical analyses. Outcome measures for postural and joint sway measures and interjoint coordination were compared among the Y-group, M-group, and S-group for each of the conditions separately. The Shapiro-Wilk test was used to determine the normality of the data. Homogeneity of variance was examined using Levene's test. When both normality and homogeneity of variance were satisfied, a one-way ANOVA with the Bonferroni correction was used to compare among participant groups. When either the normality or homogeneity of variances was violated, the Kruskal-Wallis H test with the Dunn-Bonferroni comparison test was used to compare the outcomes among the three groups. The CI and UCM ratios were log transformed before the statistical analyses. Spearman's rank correlation coefficients were used to characterize the correlation between postural sway measures and interjoint coordination measures.

RESULTS

Postural Sway

Figure 2, *A* and *B*, shows the postural sway measures for each participant group and for all participants. Kruskal-Wallis H tests revealed significant differences among the three participant groups during EO and EC conditions for COP velocity [EO: $\chi^2(2) = 16.70$, $P < 0.001$; EC: $\chi^2(2) = 15.56$, $P < 0.001$] and COM acceleration [EO: $\chi^2(2) = 16.56$, $P < 0.001$; EC: $\chi^2(2) = 17.64$, $P < 0.001$]. Post hoc tests revealed significant difference between S-group and both Y- and M-groups for COP velocity during EO (Y-S groups: $P = 0.003$; M-S groups: $P = 0.001$) and EC conditions (Y-S groups: $P = 0.002$; M-S groups: $P < 0.001$). Similar results were found for COM acceleration during EO (Y-S groups: $P = 0.003$; M-S groups: $P = 0.001$) and EC conditions (Y-S groups: $P = 0.002$; M-S groups: $P < 0.001$). Figure 2*C* show the Romberg ratio for COP velocity and COM acceleration, and no significant differences were found across participants groups.

Ankle and Hip Joint Angular Accelerations

Figure 3, *A* and *B*, shows the ankle and hip joint angular accelerations for all participants. Kruskal-Wallis H tests revealed significant differences among the three participant groups for the ankle joint acceleration [EO: $\chi^2(2) = 13.01$, $P = 0.001$; EC: $\chi^2(2) = 18.18$, $P < 0.001$] and the hip joint acceleration [EO: $\chi^2(2) = 9.64$, $P = 0.008$; EC: $\chi^2(2) = 13.20$, $P = 0.001$]. Post hoc tests revealed that the ankle acceleration was significantly larger for the S-group than both Y- and M-groups during EO (Y-S group: $P = 0.015$; M-S group: $P = 0.003$) and EC conditions (Y-S group: $P = 0.004$ M-S group: $P < 0.001$). Additionally, the S-group displayed significantly higher hip acceleration compared with the M-group during EO task (Y-S group: $P = 0.114$ M-S group: $P = 0.008$) and to both Y- and M-groups during EC condition (Y-S group: $P = 0.045$; M-S group: $P = 0.001$).

Figure 3, *C* and *D*, shows the group averages of the weighted ankle and hip joint accelerations for the three participant groups. Kruskal-Wallis H tests revealed significant differences among the three participant groups for the weighted ankle acceleration [EO: $\chi^2(2) = 13.89$, $P = 0.001$; EC: $\chi^2(2) = 19.60$, $P < 0.001$] and hip joint acceleration [EO: $\chi^2(2) = 7.79$, $P = 0.020$; EC: $\chi^2(2) = 11.39$, $P = 0.003$]. Post hoc tests revealed that the weighted ankle acceleration was significantly larger for S-group than both Y- and M-groups during EO (Y-S group: $P = 0.024$; M-S group: $P = 0.001$) and EC conditions (Y-S group: $P = 0.006$; M-S group: $P < 0.001$). The S-group also displayed significantly larger hip acceleration compare to the M-group during EO (Y-S group: $P = 0.146$; M-S group: $P = 0.023$) and during EC conditions (Y-S group: $P = 0.166$ M-S group: $P = 0.002$).

Interjoint Coordination

Figure 4, *A–D*, shows the group averages of the CI, GEV, NGEV, and UCM ratio for the three participant groups. CI was not statistically different between groups (EO: $\chi^2 = 1.16$, $P = 0.56$; EC: $F_{2,38} = 0.987$, $P = 0.38$). GEV (EO: $\chi^2 = 9.73$, $P = 0.008$; EC: $\chi^2 = 13.62$, $P = 0.001$) and NGEV

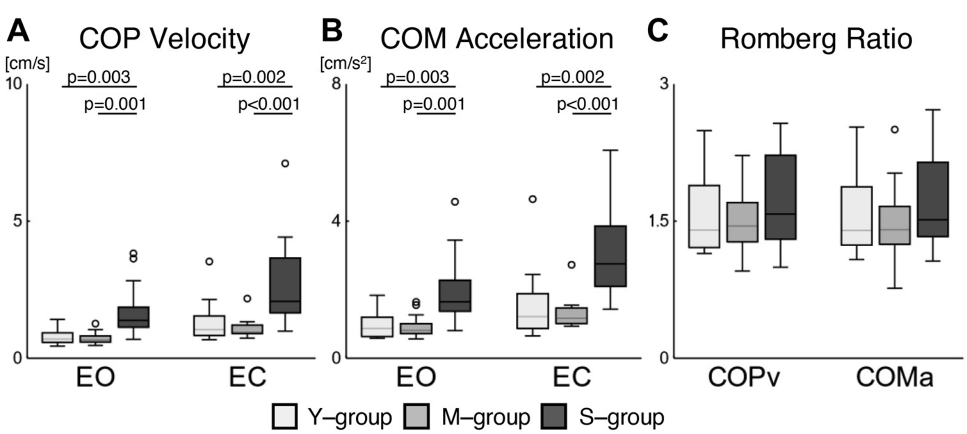

Figure 2. Box-plot diagram showing the group distribution of the postural sway measures for center-of-pressure (COP) velocity (COPv; *A*), center-of-mass (COM) acceleration (COMa; *B*), and Romberg ratio during quiet standing tasks with eyes open (EO) and eyes closed (EC; *C*) for young able-bodied group (Y-group), for age- and sex-matched able-bodied group (M-group), and for those with incomplete spinal cord injury (S-group). The box plot contained the 25th and 75th percentile with the center line denoting the median values. The outliers are shown as an open circle, and the whisker extends to the farthest points that were not outliers.

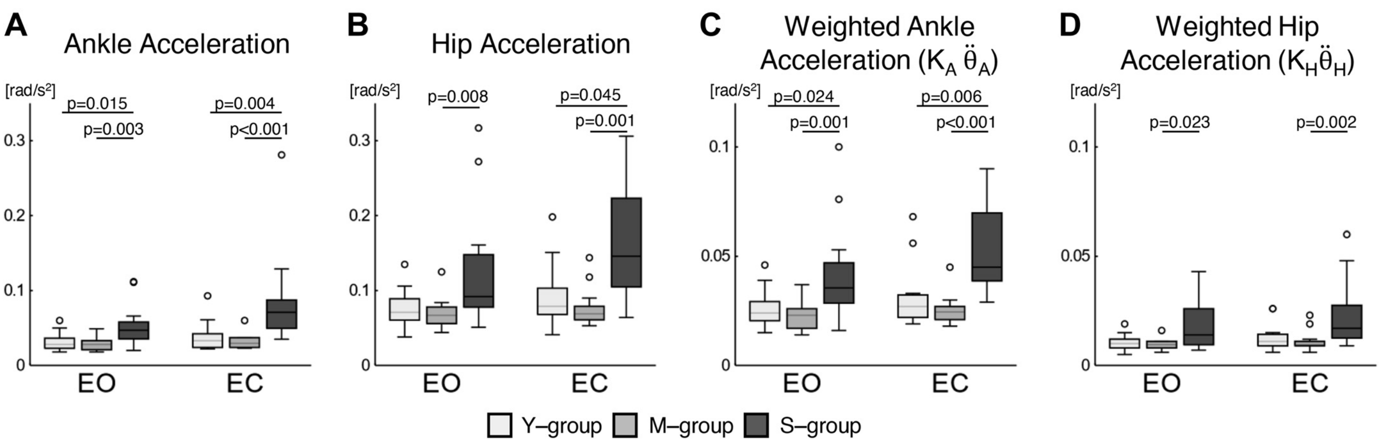

Figure 3. Box-plot diagram showing the group distribution of the individual joint accelerations for ankle acceleration (*A*), hip acceleration (*B*), weighted ankle acceleration (*C*), and weighted hip acceleration (*D*) during quiet standing tasks with eyes open (EO) and eyes closed (EC) for young able-bodied group (Y-group), for age- and sex-matched able-bodied group (M-group), and for those with incomplete spinal cord injury (S-group). The box plot contained the 25th and 75th percentile with the center line denoting the median values. The outliers are shown as an open circle, and the whisker extends to the farthest points that were not outliers. K_A and K_H, relative ankle and hip weights; θ_A and θ_H, ankle and hip joint angles.

(EO: $\chi^2 = 13.44$, $P = 0.001$; EC: $\chi^2 = 18.44$, $P = 0.008$) were statistically different among the three groups while the UCM ratio was not statistically different [EO: $\chi^2(2) = 0.11$, $P = 0.94$; EC: $\chi^2(2) = 0.67$, $P = 0.72$]. Post hoc tests revealed that the GEV was larger for S-group than the M-group during EO (Y-S groups: $P = 0.104$; M-S groups: $P = 0.008$) and was larger than both Y- and M-groups during EC condition (Y-S groups: $P = 0.030$; M-S groups: $P = 0.001$). NGEV was significantly larger for the S-group compared with both Y- and M- groups during EO (Y-S groups: $P = 0.014$; M-S groups: $P = 0.002$) and EC conditions (Y-S groups: $P = 0.002$; M-S groups: $P < 0.001$).

Figure 5, *A–D*, shows the Spearman's rank correlation coefficients between COM acceleration and CI, GEV, NGEV, and UCM ratio. Across the three participant groups, low Spearman's rho between COM acceleration with CI (EO: $P = 0.18$, $P = 0.24$; EC: $\rho = 0.07$, $P = 0.64$) and with UCM ratio (EO: $\rho = -0.02$, $P = 0.88$; EC: $\rho = -0.10$, $P = 0.54$) was observed. In contrast, high Spearman's rho between COM acceleration with GEV (EO: $\rho = 0.75$, $P < 0.01$; EC: $\rho = 0.73$, $P < 0.01$) and NGEV (EO: $\rho = 0.77$, $P < 0.01$; EC: $\rho = 0.84$, $P < 0.01$) was observed.

DISCUSSION

The COP velocity and COM acceleration were greater for individuals with iSCI compared with AB individuals (Fig. 2). Both ankle and hip joint angular acceleration was significantly greater for the iSCI group compared with both young and matched AB groups (Fig. 3, *A* and *B*), while the contribution of joint acceleration to the COM acceleration was shown to be more prominent in the ankle joint compared with the hip joint (Fig. 3, *C* and *D*). We also found that NGEV and GEV were significantly larger in individuals with iSCI (Fig. 4, *B* and *C*) and that they were highly correlated with the COM acceleration (Fig. 5, *B* and *C*).

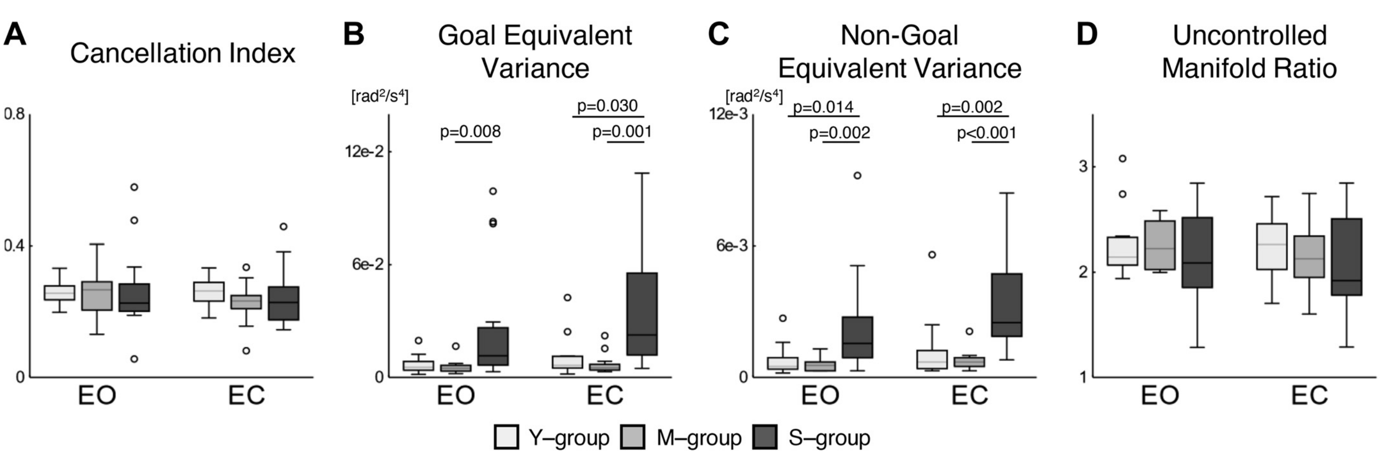

Figure 4. Box-plot diagram showing the group distribution of the ankle-hip joint coordination measures for cancellation index (CI; *A*), goal equivalent variance (GEV; *B*), nongoal equivalent variance (NGEV; *C*), and uncontrolled manifold (UCM; *D*) ratio during quiet standing tasks with eyes open (EO) and eyes closed (EC) for young able-bodied group (Y-group), for age- and sex-matched able-bodied group (M-group), and for those with incomplete spinal cord injury (S-group). The values for CI and UCM were log transformed. Each box plot contained the 25th and 75th percentile with the center line denoting the median values. The outliers are shown as an open circle, and the whisker extends to the farthest points that were not outliers.

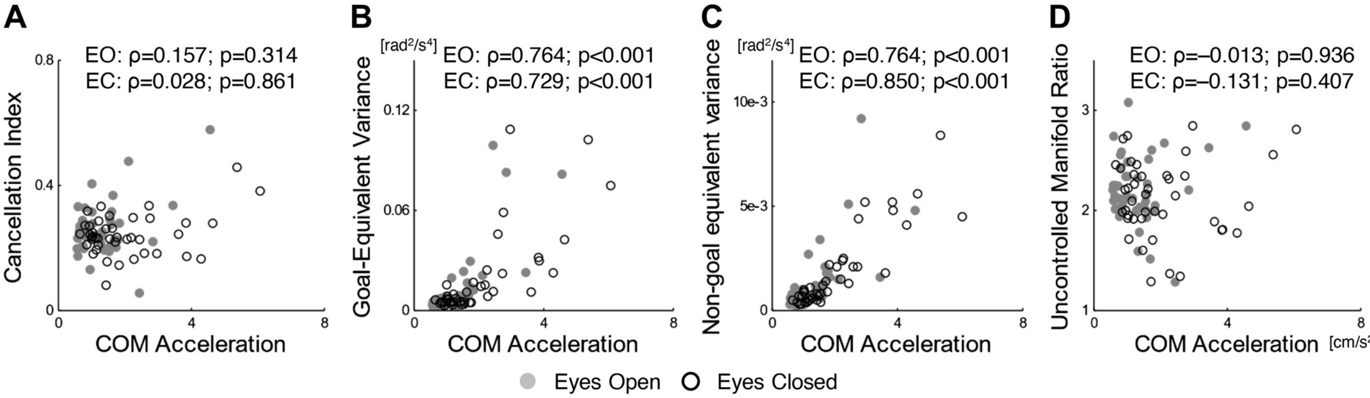

Figure 5. The Spearman correlation between center-of-mass (COM) acceleration and cancellation index (CI; *A*), goal equivalent variance (GEV; *B*), non-goal equivalent variance (NGEV; *C*), and uncontrolled manifold (UCM; *D*) ratio during quiet standing tasks with eyes open (EO) and eyes closed (EC). Each circle represents individual values, and the corresponding Spearman's rank correlation coefficient (ρ) and the *P* values for EO and EC trials are shown.

Larger Ankle Angular Acceleration Causes Larger COM Acceleration in Individuals with iSCI

We found that participants with iSCI showed larger postural sway, i.e., larger COP velocity and COM acceleration. The results align with findings from Lemay et al. (4) showing that the COP velocity was larger in individuals with iSCI compared with AB individuals. Although they showed that the Romberg ratio for COP velocity was larger in individuals with iSCI, indicating higher reliance on visual information in individuals with iSCI, our results did not show significant differences for either of the postural sway measures, which indicated that the visual reliance is not different between the groups. However, compared with Lemay et al. (4), the Romberg ratios of both Y- and M-groups in the present study were higher [Y-group: median: 1.40 and interquartile range: 1.23–1.84; M-group: median: 1.45 and interquartile range: 1.30–1.69; Lemay et al. (4): median: 1.22 and interquartile range: 1.02–1.43], while those with iSCI had similar Romberg ratios [S-group: median: 1.58 and interquartile range: 1.32–2.10; Lemay et al. (4): median: 1.69 and interquartile range: 1.38–2.38]. Also, the postural sway in the AB groups in Lemay et al. (4) (median: 7.70 mm/s; interquartile range: 6.34–9.37 mm/s) swayed larger than the AB groups in current study while standing with EO (Y-group: median: 7.00 mm/s; interquartile range: 5.91–8.94 mm/s; M-group: median: 6.62 mm/s; interquartile range: 5.93–8.03 mm/s). The difference of the AB participants' characteristics as well as longer task duration [150 s in the current study against 45 s in Lemay et al. (4)] may have led to this different finding within AB participants.

The hip angular acceleration was much larger than the ankle angular acceleration in both AB and individuals with iSCI (Fig. 3, *A* and *B*). This corresponded with previous studies of AB individuals (9, 12). However, the contribution of the ankle angular acceleration to the COM acceleration was twice as large compared with that of the hip angular acceleration (Fig. 3, *C* and *D*). Thus regulating the ankle angular acceleration has a larger effect in controlling the COM acceleration even if the hip angular acceleration is larger than the ankle angular acceleration. Comparing AB individuals and those with iSCI, both joint accelerations are similarly larger in individuals with iSCI, while only the weighted ankle angular acceleration is more clearly larger in the iSCI group. This suggests that larger ankle joint acceleration is the primary factor in the larger COM acceleration during quiet standing in individuals with iSCI.

Interjoint Coordination Is Not Affected by iSCI

Both CI and UCM ratio did not indicate the interjoint coordination differed between AB-groups and S-group (Fig. 4, *A* and *D*), which does not support our hypothesis that the interjoint coordination is impaired in individuals with iSCI.

CI quantifies the degree of reciprocal relationship between the ankle and hip angular accelerations (12). Our results indicate that this reciprocal relationship was similar between AB-groups and the S-group. In addition, CI does not correlate with the COM acceleration (Fig. 5*A*), suggesting that the reciprocal relationship may not be the primary determinant of regulating the postural sway.

The UCM ratio quantifies how the joint accelerations are distributed along the zero-COM acceleration axis (i.e., GEV) compared with the distribution on the orthogonal axis (i.e., NGEV). The UCM ratio did not differ between the AB-groups and S-group, suggesting the shapes of the distributions on the angular-acceleration plane were similar between groups. Hsu et al. (15) demonstrated that the UCM ratio is smaller in older adults compared with younger adults as the older adults exhibited larger NGEV, which suggested that interjoint coordination deteriorates with age. We hypothesized a similar deterioration could be induced by iSCI; however, this is not supported by our findings. This may have been because the distribution was proportionally larger in S-group indicated by larger GEV and NGEV (Fig. 4, *B* and *C*). The interjoint coordination orthogonal to the zero-COM acceleration line quantified by NGEV typically increases the COM acceleration, which must relate to a result that the NGEV highly correlated with the COM acceleration (Fig. 5*C*). This may reflect the causal relation between the interjoint activity and the COM acceleration. On the contrary, the distribution along the zero-COM acceleration line quantified by the GEV is not supposed to affect the COM acceleration, although we found a high correlation between the GEV and the COM acceleration (Fig. 5B). Theoretically, this does not reflect a causal relation, and we assume that this may be a result of collinearity among GEV, NGEV, and the COM acceleration.

In case of a completely rigid hip joint, the ankle joint angular acceleration purely determines the COM acceleration. The hip joint angular acceleration further increases or decreases the COM acceleration when the hip joint is flexible (i.e., less rigid). For example, in the case the hip joint angular acceleration is at the ratio of $-K_A/K_H$ to the ankle angular acceleration or the ankle-hip relation is completely on the zero-COM acceleration line, the COM acceleration is zero regardless of the magnitude of the ankle joint acceleration. On the contrary, when the ankle-hip joint angular accelerations are not distributed on the zero-COM acceleration line, the COM acceleration will increase. Based on the current results with individuals with iSCI, the ankle angular acceleration is larger than AB individuals while the interjoint coordination is similar between groups. These show that the compromised regulation of ankle joint acceleration in individuals with iSCI induces increased COM acceleration. In addition, as the interjoint coordination is at a similar level between groups, the interjoint coordination does not successfully compensate increased ankle angular acceleration, resulting in larger COM acceleration.

Limitation and Implication

First, the above-mentioned results are observational and based on the resultant kinematics and kinetics only. Therefore, we are not able to conclusively indicate the control mechanism of standing balance. For example, the larger ankle joint acceleration in individuals with iSCI can be caused by impaired controls of not only ankle but also hip joints. Additional experimental approaches are required to understand the effect of iSCI on the neural control system of quiet standing. Secondly, the participants with iSCI in this study had relatively high motor function and could independently stand without any aids, while their clinical scores and self-reported scores showed variations (Tables 1 and 2). Therefore, our conclusion is applicable to those populations and not to the general iSCI population. Furthermore, the participants analyzed in this study were predominantly female (12/16) and include both traumatic (9/16) and nontraumatic cases of the SCI (7/16), such as from infections or vascular disorders, which may have caused a conclusion that does not represent the iSCI population. Finally, this study assumed that the body dynamics during quiet standing could be modeled using a double-link inverted pendulum model, where we did not consider the knee joint or upper body joints that may contribute to maintaining standing balance (14, 25, 26).

Larger postural sway is often associated with deteriorated balance ability. For example, it has been shown that larger postural sway predicts future falls in the elderly population (27). In our previous study using the same data set, we demonstrated that the COP velocity is negatively correlated with both of mini-balance evaluation systems test (mini-BESTest) scores and lower extremity muscle strengths scores (19), suggesting that the larger postural sway reflects their deteriorated ability in standing balance and motor function. Furthermore, in another study using the same data set, we demonstrated that the participants with lower community balance & mobility score (CB&M) (<48) tend to show larger COP velocity while the participants with higher CB&M (>48) show similar COP velocity compared with AB individuals (28). Thus our result that the postural sway is larger in our participants with iSCI indicate that their balance ability is deteriorated even though the overall motor functions in our participants were relatively high among the general SCI population. Our results suggest that this deterioration in balance ability is primarily due to the ankle joint acceleration but not due to the interjoint coordination. Therefore, focusing on the ankle kinematics may be a potential approach in their rehabilitation to improve standing balance. However, the larger postural sway may be a compensatory strategy to increase sensory feedback (29) and may not directly indicate deteriorated balance ability. The current study relying only on kinematics does not provide any supports to either opinion, and further research is required to investigate the relation of the current results to fall.

Similar to our previous studies mentioned above (19, 28), we tried to find out relationships between clinical scores and interjoint coordination measures, but we did not find significant relationships. For example, the correlation coefficients between mini-BESTest score and the CI were EO: $\rho = -0.097$, $P = 0.731$; EC: $\rho = -0.499$, $P = 0.069$, and the correlation coefficients between mini-BESTest score and the UCM ratio were EO: $\rho = -0.065$, $P = 0.819$; EC: $\rho = -0.243$, $P = 0.402$. Similarly, the correlation coefficients between CB&M score and the CI were EO: $\rho = -0.159$, $P = 0.571$; EC: $\rho = -0.376$, $P = 0.185$, and the correlation coefficient between CB&M score and the UCM ratio was EO: $\rho = -0.144$, $P = 0.685$; EC: $\rho = -0.563$, $P = 0.402$. In addition, neither CI nor UCM ratio were different between participants with higher CB&M scores (>48) and lower CB&M scores (<48). These results further support the above-mentioned suggestion that the interjoint coordination was not deteriorated in individuals with iSCI as the interjoint coordination was unrelated to the severity of motor functions detected by these clinical scores. Larger sample size with wider variations in motor functions may show relationships between the clinical scores and interjoint coordination measures. Nonetheless, postural sway measures, such as COM acceleration and COP velocity, were more sensitive measures to detect the deterioration in standing balance compared with the interjoint coordination in individuals with iSCI.

Conclusions

We demonstrated that the larger postural sway in individuals with iSCI (i.e., COM acceleration and COP velocity) during quiet standing is primarily due to the larger ankle joint acceleration. We also demonstrated that the interjoint coordination between ankle and hip joint is not affected in individuals with iSCI, which does not successfully compensate the abovementioned increased COM acceleration.

Rehabilitation that focuses on the ankle joint may particularly help in reducing postural sway during standing in individuals with iSCI and in assessing balance issues and fall risks, while further research is required with larger sample size including an investigation on the relation of the current finding and fall risks.

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DISCLOSURES

No conflicts of interest, financial or otherwise, are declared by the authors.

AUTHOR CONTRIBUTIONS

J.W.L., K.E.M., and K.M. conceived and designed research; J.W.L., K.C., J.U., J.Y., and K.E.M. performed experiments; J.W.L. analyzed data; J.W.L. and K.M. interpreted results of experiments; J.W.L. prepared figures; J.W.L. drafted manuscript; J.W.L., K.C., J.U., J.Y., K.E.M., and K.M. edited and revised manuscript; J.W.L., K.C., J.U., J.Y., K.E.M., and K.M. approved final version of manuscript.

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