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Temporal relationship between trunk and thigh contributes to balance control in load carriage walking

Sheng-Che Yen [a,*], Wen Ling [b], Richard Magill [b], Andrew McDonough [c], Gregory M. Gutierrez [b]

[a] Sensory Motor Performance Program, Rehabilitation Institute of Chicago, Suite 1406, 345 E, Superior Street, Chicago, IL 60611, United States
[b] Department of Physical Therapy, New York University, NY 10010, United States
[c] Graduate Programs in Health Sciences, Seton Hall University, NJ 07079, United States

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ABSTRACT

Load carriage walking (LCW) challenges a person's balance as the load increases their forward trunk inclination, shifting the center of mass (COM) forward with respect to the base of support (BOS). We examined LCW to understand whether and how healthy people adjust the temporal relationship (TR) between the trunk and leg for balance control. Ten subjects were recruited to perform unloaded walking and LCW. The TR between the trunk and leg was measured by the continuous relative phase. The maximum forward displacement of the COM with respective to the BOS ($FDCOM_{BOS}$) was recorded during the stance phase. We found that the TR was shifted in LCW, and the shift was associated with a decrease in the maximum $FDCOM_{BOS}$. The findings suggest that the TR between the trunk and leg contributes to balance control, and it may be a variable that needs to be addressed in gait rehabilitation.

1. Introduction

Balance control during walking requires the maintenance of a proper relative position between the body center of mass (COM) and the base of support (BOS) [1]. To achieve this goal, the trunk and leg need to work together in a coordinated manner. Since the trunk makes up approximately 50% of the body weight [2], any uncoordinated movement between the trunk and leg could shift the COM away from the BOS and cause a loss of balance [3]. However, how the trunk and leg are coordinated for balance control during walking is not fully understood.

Coordination during walking has been defined as controlling the cyclic temporo-spatial relationship between body segments [4]. A number of studies have evaluated how people adapt the spatial relationship between the trunk and leg to maintain walking balance [3,5,6]. Specifically, people tend to adopt a "crouched" posture (i.e., increases in hip flexion, knee flexion, and ankle dorsiflexion) when the trunk is constrained in an anteriorly inclined position. However, we are not aware of any study which has investigated whether and how people adapt the temporal relationship (TR) between the trunk and leg to maintain walking balance.

This study examined load carriage walking (LCW) as a model to generate insight into this question. Here LCW refers to walking while carrying a load using a backpack. We focused our examination in the sagittal plane, since previous findings suggest that people are more prone to loss of balance in the sagittal plane than in the frontal plane in LCW. Specifically, people tend to increase forward trunk inclination (FTI) in LCW [7–9], which brings the COM forward with respect to the BOS, making it more difficult to maintain balance in the sagittal plane [3]. On the other hand, people tend to limit the medial–lateral excursion of the COM in LCW, which may help keep the COM close to the BOS in the frontal plane [7]. Additionally, the BOS in the sagittal plane, as represented by step length, tends to decrease in LCW due to a decrease in pelvic motion [10]. Thus, LCW presents an opportunity for studying balance control in the sagittal plane.

We hypothesized that people would adjust the TR between the trunk and leg to maintain walking balance during LCW. We expected to observe the following outcomes: (1) LCW would be associated with an increase in the maximum forward displacement of the COM with respect to the BOS ($FDCOM_{BOS}$) during the initial double-limb-support (IDLSP), single-limb-support (SLSP), and terminal double-limb periods (TDLSP) of the stance phase; (2) the TR between the trunk and leg would be shifted during each period of the stance phase in LCW; (3) the shift in the TR would be associated with a reduction in the maximum $FDCOM_{BOS}$ during each period of the stance phase in LCW. The TR between the trunk and leg was measured by continuous relative phase (CRP) between the trunk and thigh [11,12], as the thigh drives leg movement during walking.

* Corresponding author. Tel.: +1 3122382228.
E-mail address: syen@ric.org (S.-C. Yen).

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2. Methods

2.1. Subjects

A convenience sample of 10 subjects (5 males/5 females; age = 30.1 ± 3.7 years; height = 170.3 ± 12.5 cm; weight = 67.7 ± 14.2 kg) participated and provided informed consent (approved by the local committee on activities involving human subjects). No subjects had any neuromuscular or musculoskeletal problems affecting their trunk, arms, legs, and walking.

2.2. Instrumentation

Modular Light Weight Load Carrying Equipment (MOLLE), a standard backpack system used by the United States Marine Corps, was used. A block of foam and a number of sandbags were put into the main rucksack of the MOLLE to attain the required loads. A 5-camera Qualisys motion analysis system (Qualisys AB, Göthenborg, Sweden) was used to capture subjects' walking at a sampling rate of 120 Hz. A GAITRite system (CIR Systems, Inc., Clifton, NJ) was used to monitor subjects' walking speed at a sampling rate of 80 Hz.

2.3. Procedures

Data collection was conducted indoors in a laboratory setting. Before data collection, a total of 20 reflective makers were attached to each subject to create a 3-degrees-of-freedom, 14-segment model (Fig. 1A) [13]. During data collection, subjects were asked to walk at a speed of 4.8 (±10%) km/hr under 3 load conditions: no load, 10%, and 20% of body weight (BW). Each condition included 12 data collection trials. Prior to the data collection trials, several practice trials were provided for subjects to become familiarized with the loads and walking speed. During each trial, subjects were asked to walk naturally from a starting line, over a four-meter GAITRite walkway, and then circle back to the starting line 3 times in a continuous manner. Data were collected without subjects' awareness in the second or third pass. The walking speed was calculated automatically by the GAITRite software (v3.4) immediately after each trial. For safety, the order of conditions was fixed from unloaded to 10%, then 20% of the BW, allowing subjects to experience the load in a gradual manner to minimize the risk of back and leg injuries. A 2-min break was provided between trials and a 10-min break was provided between conditions to reduce the potential for fatigue.

2.4. Data processing

Based on the camera setup, the Qualisys system was able to capture all 20 markers in the middle of the walkway for 6–8 steps in each trial. From these steps, we randomly selected 4 consecutive steps (a left cycle and a right cycle) for data analysis. Marker data were smoothed using a fourth-order Butterworth filter at a cut-off frequency of 6 Hz. Based on the smoothed marker data, the COM trajectory over each walking cycle was calculated using Hatze's mathematical model [13]. To identify the IDLSP, SLSP, and TDLSP, heel contact and toe-off were detected automatically using the vertical velocity profile of the foot center proposed by O'Connor et al. [14]. Additionally, visual inspection of stick figures was done to verify the heel contact/toe-off determined by auto-detection. The foot center was defined as the midpoint between the heel and forefoot markers (Fig. 1A). The maximum $FDCOM_{BOS}$ during SLSP was measured as the maximum anterior displacement of the COM relative to the center of the support foot. The maximum $FDCOM_{BOS}$ during IDLSP and TDLSP was measured as the maximum anterior displacement of the COM relative to the midpoint between left and right foot centers. The maximum $FDCOM_{BOS}$ was averaged across both left and right cycles for each trial for the statistical analysis.

Fig. 1B shows how we measured the trunk and thigh positions for each side. The CRP between the trunk and thigh was calculated based on the position data [11,12]. First, we differentiated the position data of trunk and thigh against time to obtain the velocity data. Second, both position and velocity data were linearly normalized to be between 1 (maximum FTI or thigh flexion) and −1 (maximum backward trunk inclination or thigh extension). Next, a phase plane plotting normalized velocity (*Y* axis) against normalized position (*X* axis) over a walking cycle was constructed for each segment. Then, each segment's temporal position on the phase plane, or the phase angle, was calculated as:

$$\Phi(t) = \arctan\frac{v(t)}{S(t)} \tag{1}$$

where $v(t)$ is the normalized velocity at given time t, and $s(t)$ is the normalized position at the same time t. The CRP between the trunk and thigh was calculated as:

$$\Phi(t)_{\text{trunk-thigh}} = \left|\Phi(t)_{\text{trunk}} - \Phi(t)_{\text{thigh}}\right| \tag{2}$$

A CRP of 0° (in-phase) indicates that the two body segments are at the same temporal position on the phase plane; a CRP of 180° (anti-phase) indicates that the two body segments are at opposite temporal positions on the phase plane [15].

The raw CRP data were a series of CRP points over each walking cycle (CRP trajectory). For statistical analysis, the corresponding CRP points were averaged (ACRP) for each phase (IDLSP, SLSP, and TDLSP). The ACRPs were then averaged across both left and right cycles for each trial for the statistical analysis.

2.5. Statistical analysis

Linear mixed models with repeated measures were conducted to examine whether the $FDCOM_{BOS}$ and ACRP varied as a function of load, period, and the interaction between load and period. A first-order autoregressive model was used to structure the correlation between repeated observations, assuming that events (conditions by trials by periods) occurring closer to each other in time during data

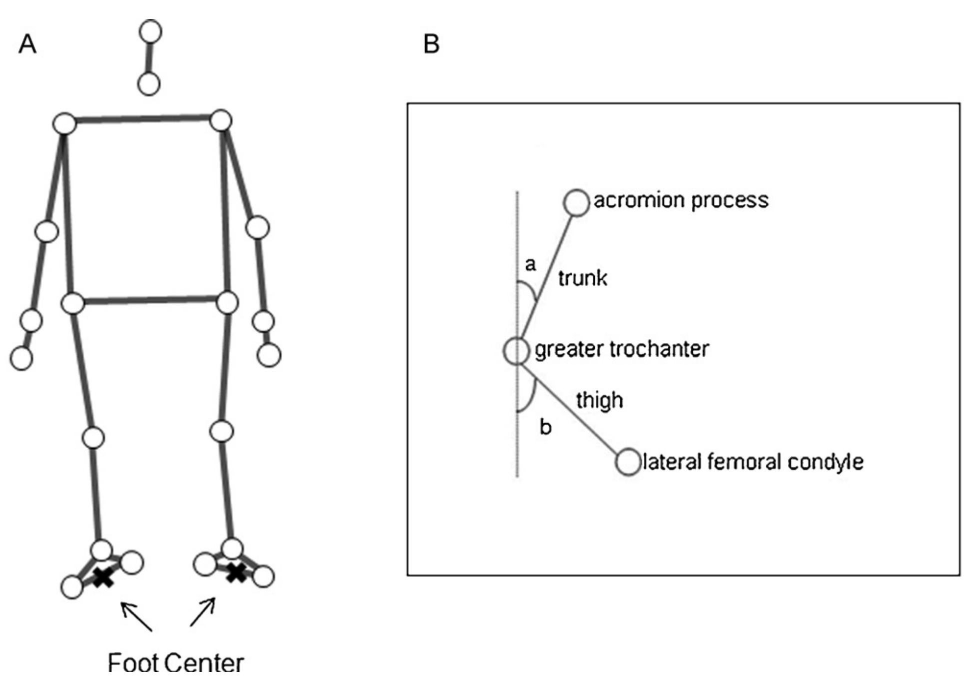

Fig. 1. (A) A total of 20 markers were placed in the following locations: forehead, chin, and bilaterally over the acromion process, lateral humeral epicondyle, distal radius, third metacarpal head, greater trochanter, lateral femoral condyle, calcaneous, lateral malleolus, and fifth metatarsal head, to form a 3-degrees-of-freedom, 14-segment model. The 14 segments include: the head, trunk, upper arms, forearms, hands, thighs, shanks, and feet. The cross signs represent the foot center on each side, which is located at the mid-point between the fifth metatarsal head and calcaneous. (B) The positions of the trunk (a) and thigh (b) at each side were measured as the angle of intersection between the corresponding segment and a vertical line passing through the greater trochanter.

collection should have a stronger correlation [16]. A multiple regression analysis was carried out to examine if the maximum $FDCOM_{BOS}$ varied as a function of ACRP, controlling for load and period. The α level was set at 0.05 for all statistical procedures, which were conducted using the Statistical Package for Social Sciences (SPSS) version 16.0 (SPSS, Inc., Chicago, IL).

3. Results

3.1. Trunk adaptation

The trunk demonstrated a greater forward inclination (Fig. 2A), a restricted movement range, and a lower velocity (Fig. 2B) during the stance phase in LCW compared with those in unloaded walking. After the position and velocity data were normalized to be between 1 and -1, we observed some timing changes in trunk oscillation due to the additional load. First, forward inclination was delayed during the IDLSP and SLSP – the upward trajectory is slightly shifted to the right along the X (time) axis in both loaded conditions (Fig. 2C). Second, backward inclination occurred sooner during the TDLSP; the downward trajectory is shifted to the left along the time axis in both loaded conditions. Third, the direction of trunk velocity switched from forward to backward earlier during the SLSP in both loaded conditions (Fig. 2D; dashed arrow) than that in the unloaded condition (solid arrow). With these timing changes, the position of the trunk in the loaded conditions was farther from the maximum FTI (farther from 1) at the end of the IDLSP and across most of the SLSP and TDLSP compared with that in the unloaded condition. The changes in the normalized position and velocity also caused a shift in the temporal position on the phase plane. For example, the temporal positions were different at the end of the IDLSP and at the end of the TDLSP between the unloaded and loaded conditions (Fig. 2E).

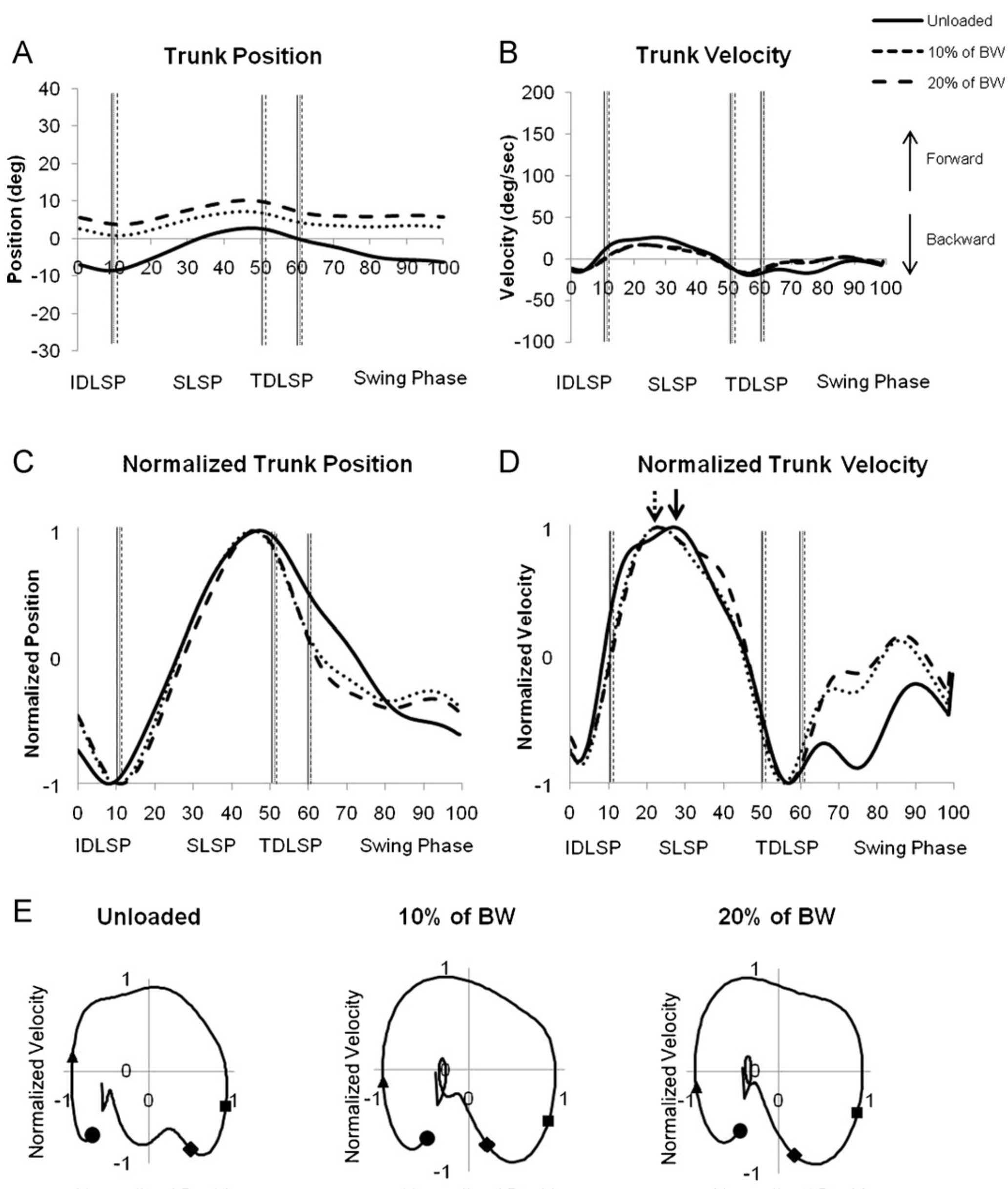

Fig. 2. The mean trunk trajectory for position (A), velocity (B), normalized position (C), and normalized velocity (D) during each period of the stance phase in the three conditions. IDLSP, initial-double-limb-support period; SLSP, single-limb-support period; TDLSP, terminal-double-limb-support period. In (D), the solid arrow and dashed arrow indicate the point at which the direction of trunk velocity switches from forward to backward during the SLSP in the unloaded condition and loaded conditions, respectively. The phase plane in each condition is shown in (E). On each phase plane, the dark circle, triangle, square, and diamond represent the beginning of the IDLSP, the end of the IDLSP, the end of the SLSP, and the end of the TDLSP in each condition, respectively.

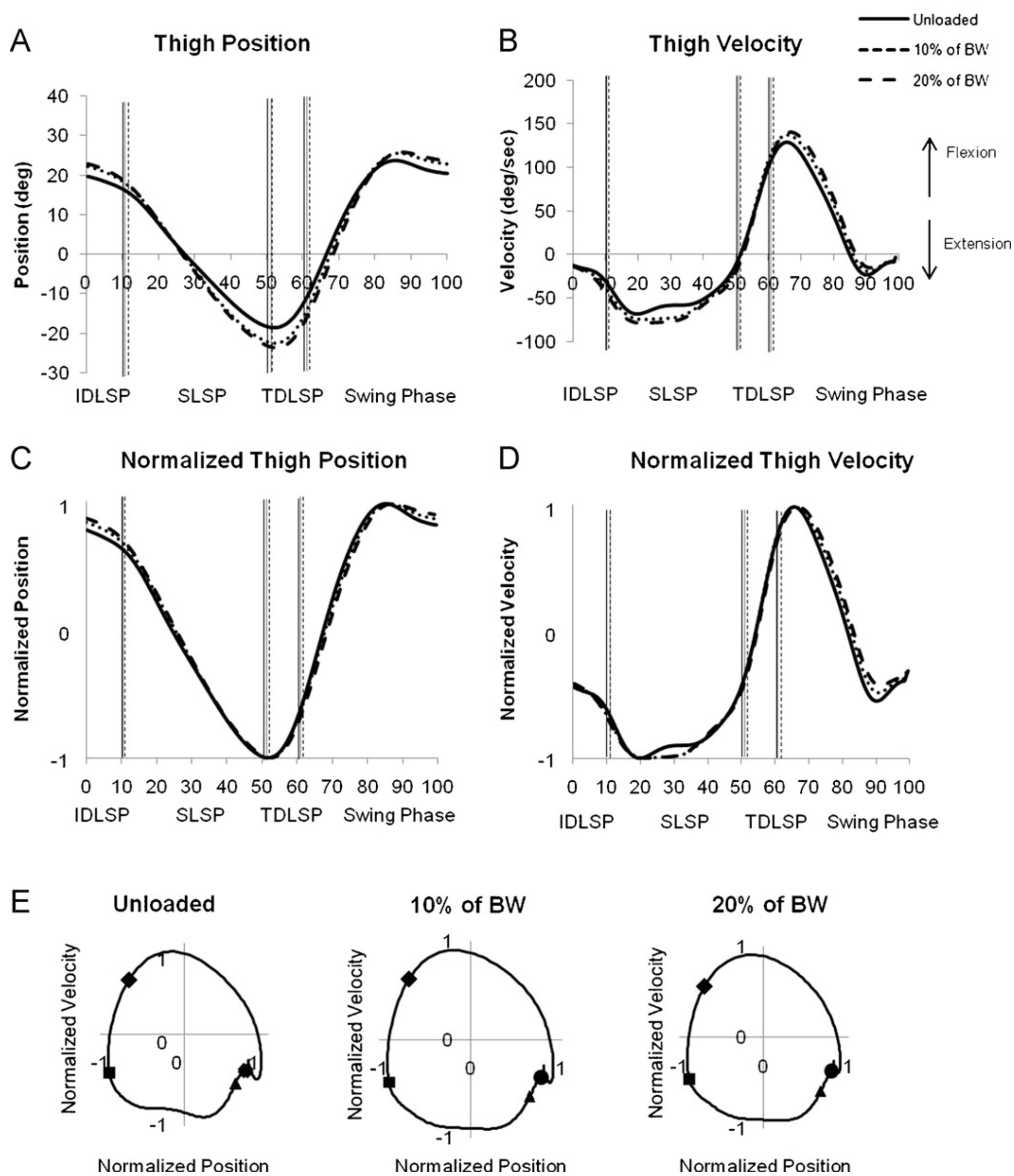

Fig. 3. The mean thigh trajectory for position (A), velocity (B), normalized position (C), and normalized velocity (D) during each period of the stance phase in the three conditions. IDLSP, initial-double-limb-support period; SLSP, single-limb-support period; TDLSP, terminal-double-limb-support period. The phase plane in each condition is shown in (E). On each phase plane, the dark circle, triangle, square, and diamond represent the beginning of the IDLSP, the end of the IDLSP, the end of the SLSP, and the end of the TDLSP in each condition, respectively.

3.2. Thigh adaptation

The thigh showed less adaptation to the load than did the trunk. The thigh movement range (Fig. 3A) and the extension velocity (Fig. 3B) during the SLSP were slightly increased in both loaded conditions. No apparent difference was observed among the three conditions after the position and velocity were normalized to be between 1 and -1 (Fig. 3C and D), except that the relative position of the thigh was farther from maximum extension during the IDLSP in both loaded conditions (Fig. 3C). Fig. 3E shows that the temporal position of the thigh on the phase plane was slightly different between unloaded and loaded conditions at the end of the IDLSP.

3.3. TR

Subjects' mean CRP trajectory and ACRP during each period of stance phase in three conditions are presented in Fig. 4A and B, respectively. In the unloaded condition, the TR was close to anti-phase (180°) during each period. In each loaded condition, the TR shifted towards in-phase. We found a significant load effect on ACRP ($p < 0.001$). The shift was more apparent during the IDLSP and TDLSP than during the SLSP. We also found a significant interaction effect between load and period on ACRP ($p < 0.001$).

3.4. COM displacement

Fig. 4C presents subjects' mean maximum $FDCOM_{BOS}$ during each period of stance phase in the three conditions. As seen in Fig. 4C, the maximum $FDCOM_{BOS}$ increased as the load increased during each period. The amount of increase was slightly greater during the IDLSP and TDLSP than during the SLSP. Both the load effect ($p < 0.001$) and the interaction between load and period ($p < 0.001$) were significant. The maximum $FDCOM_{BOS}$ consistently occurred at the end of each period across trials and subjects.

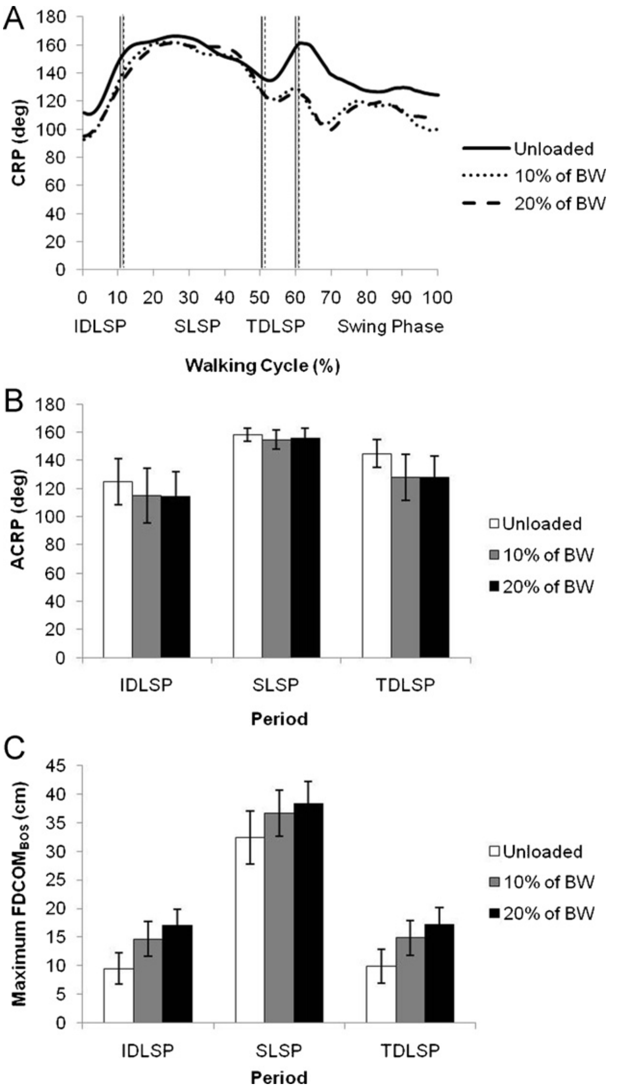

Fig. 4. (A) The mean CRP trajectory over a walking cycle in the three conditions. (B) The mean ACRP during each period of the stance phase in the three conditions. (C) The mean maximum FDCOM$_{BOS}$ during each period of the stance phase in the three conditions. IDLSP, initial-double-limb-support period; SLSP, single-limb-support period; TDLSP, terminal-double-limb-support period. The error bars in (B) and (C) represent standard deviation.

3.5. COM displacement and TR

There was a positive relationship between the maximum FDCOM$_{BOS}$ and the ACRP ($B = 0.06$, $p < 0.001$). Based on the B coefficient, for every degree of decrease in the ACRP (a shift of TR towards in-phase), there was a corresponding 0.06 cm of decrease in the maximum FDCOM$_{BOS}$.

4. Discussion

Carrying a load during walking increased the maximum FDCOM$_{BOS}$ in each period of the stance phase, making it more difficult to maintain balance. Our results suggest that subjects adjusted the TR between the trunk and leg to maintain balance in LCW. Shifting the TR towards a more in-phase pattern was associated with a reduction in the maximum FDCOM$_{BOS}$. To clarify this association, a biomechanical model showing the positions of the trunk and leg at the end of IDLSP, SLSP, and TDLSP (where the maximum FDCOM$_{BOS}$ occurred) in LCW was created in Fig. 5.

In this model, the hip joint represents the axis of rotation between the trunk and leg, with the COM located in the lower trunk. The trunk moves into a more anteriorly inclined position during load carriage walking. Based on our data, the hip was approximately 14 cm and 11 cm in front of the midpoint between the left and right foot centers at the end of the IDLSP (Fig. 5A) and TDLSP (Fig. 5C), respectively, and was approximately 30 cm in front of the center of the support foot at the end of the SLSP (Fig. 5B). The effect of higher load (from 10% to 20% of the BW) in these measures was minimal (less than 1 cm).

Based on the model, the FDCOM$_{BOS}$ can be expressed as:

$$\text{FDCOM}_{\text{BOS}} = L_4 - L_3 - L_2 \times \sin\,\theta_2 + L_1 \times \sin\,\theta_1, \quad \text{at the end of IDLSP (Fig. 5A)} \tag{3}$$

$$\text{FDCOM}_{\text{BOS}} = L_1 \times \sin\,\theta_1 + L_2 \times \sin\,\theta_2 - L_3, \quad \text{at the end of SLSP (Fig. 5B)} \tag{4}$$

$$\text{FDCOM}_{\text{BOS}} = L_1 \times \sin\,\theta_1 + L_2 \times \sin\,\theta_2 - L_3 - L_4, \quad \text{at the end of TDLSP (Fig. 5C)} \tag{5}$$

where θ_1 and θ_2 respectively denote the trunk and thigh positions, L_1 denotes the distance between the hip and the COM, L_2 denotes the leg length (from the hip to the ankle), L_3 denotes the horizontal distance between the ankle and the foot center of the corresponding leg for the period, and L_4 denotes the horizontal distance between the foot center of the corresponding leg for the period and the midpoint between the left and right foot centers. The TR mainly affected θ_1 and θ_2 as the variable was derived from the trunk and thigh positions.

The TR was shifted during LCW because the load changed the timing of trunk oscillation (Fig. 2C–E). Specifically, the trunk reached a position farther from the maximum FTI (i.e., a smaller θ_1) at the end of IDLSP, SLSP, and TDLSP. This adaptation decreased the value of $L_1 \times \sin\theta_1$ in Eqs. (3)–(5) and thus reduced the maximum FDCOM$_{BOS}$. On the other hand, the load did not change the timing of thigh oscillation (Fig. 3C–E), and therefore did not affect the thigh position θ_2 relative to its maximum flexion/extension position at the end of IDLSP, SLSP, and TDLSP. This kept the value of $L_2 \times \sin\theta_2$ constant in Eqs. (3)–(5) and thus did not affect the maximum FDCOM$_{BOS}$. This suggests that a shift in the TR could alter the positional coupling between the trunk and leg (θ_1 versus θ_2) during walking and in turn could affect the relative position between the COM and BOS, although the change in the positional coupling in this case was mainly driven by one of the segments (the trunk).

This is the first study showing how the trunk and leg adjust their temporal relationship in response to a balance threat (increased FTI in LCW). By analyzing the CRP, we revealed that shifting the TR between the trunk and leg towards a more in-phase pattern (i.e., delaying the occurrence of FTI late in the IDLSP, advancing the occurrence of backward trunk inclination during the SLSP, and keeping the timing of thigh oscillation similar to regular walking) could reduce the maximum FDCOM$_{BOS}$ and benefit balance control in LCW in healthy people. Previous studies have shown that walking with an increased FTI could induce spatial adaptations (i.e., a "crouched" posture) to keep the COM close to the BOS [3,5,6]. Our results add value to the previous studies, providing a more complete picture of how the trunk and leg are coordinated to maintain walking balance in both spatial and temporal domains.

Clinically, aging and certain neurologic disorders (e.g., Parkinson's disease and spinal pathologies) could result in abnormal FTI and in turn affect walking balance [17–19]. We postulated that individuals with such issues may benefit from a therapeutic program that shifts the TR between the trunk and leg towards a more in-phase pattern. Future study is warranted to clarify this postulation.

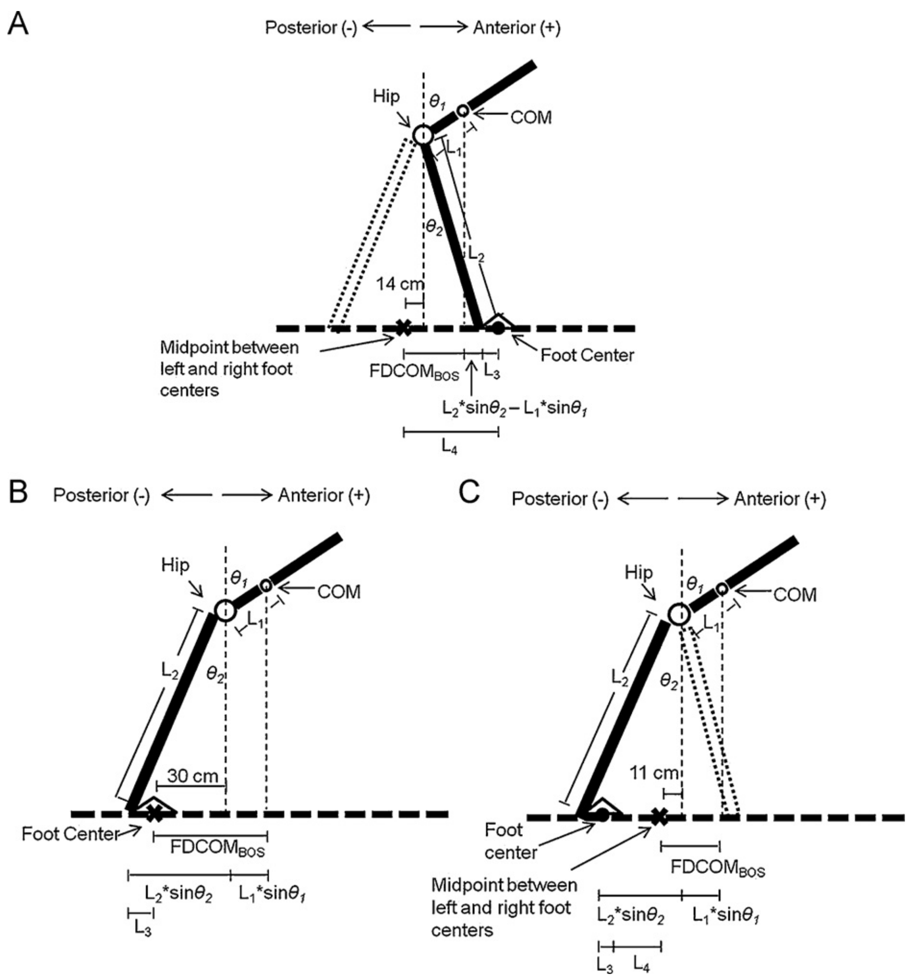

Fig. 5. A biomechanical model showing the positions of the trunk, leg, COM, and foot center at the end of the IDLSP (A), SLSP (B), and TDLSP (C). θ_1 = the trunk position; θ_2 = the thigh position, L_1 = the distance between the hip and the COM, L_2 = the leg length (distance from the hip to the ankle); L_3 = the horizontal distance between the ankle and the foot center of the corresponding leg for the period; L_4 = the horizontal distance between the foot center of the corresponding leg for the period and the midpoint between the left and right foot centers.

5. Conclusion

By examining LCW, we found that healthy people may adjust the TR between the trunk and leg for balance control. A shift in the TR could change the positional coupling between the trunk and leg during walking and in turn affect the relative position between the COM and BOS. The TR between the trunk and leg may be a variable that can be addressed in gait rehabilitation in neurologic patients to improve their balance control during walking.

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Conflict of interest statement

We, the authors of this manuscript, affirm that we have no financial and personal relationships with other people or organizations that could inappropriately influence our work.

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