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Relationship between Mediolateral Margin of Stability and Load Weight during Walking with a Unilateral Load

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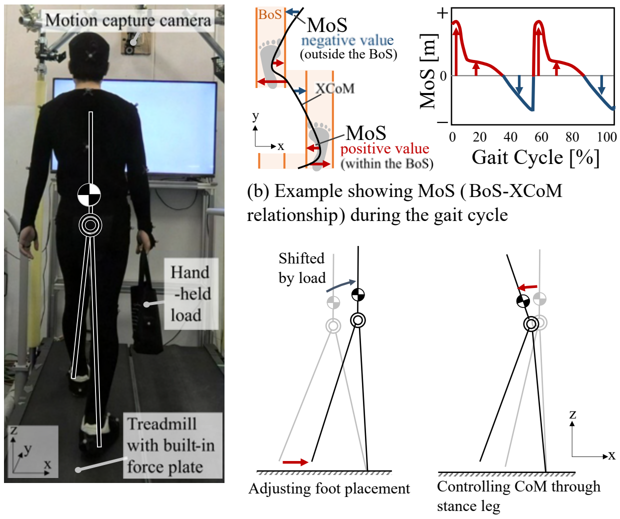
***Abstract*— Carrying a unilateral load shifts the body's center of mass (CoM) laterally, making foot placement adjustments critical for maintaining frontal-plane stability. The margin of stability (MoS), which quantifies the distance between the extrapolated CoM and base of support, provides a dynamic measure for assessing such stability control strategies. Although previous studies have examined segmental effects of asymmetric load carriage, they are limited in capturing whole-body dynamic stability. In this study, the impact of unilateral load carrying on dynamic stability was analyzed using MoS. Furthermore, the respective contributions of foot placement adjustments and trunk movement were explored. The findings revealed that, under moderate load conditions, MoS was primarily maintained through adjustments in foot placement. Conversely, under high load conditions, significant inter-individual variability was observed, with strategies ranging from foot placement adjustments to repositioning the CoM via trunk movement. Additionally, variations in trunk tilt direction were associated with distinct strategies aimed at reducing shoulder joint load. These findings provide new insights into the movement mechanisms underlying stability maintenance during unilateral luggage carrying and may lead to the development of assistive devices tailored to individual physical characteristics.**



(a) Overview of the experimental environment

(b) Example showing MoS (BoS-XCoM relationship) during the gait cycle

(c) Each movement pattern expressed using a link model

Fig. 1: Overview of the experimental setup, stability analysis, and movement pattern modeling

I. INTRODUCTION

Carrying loads while walking is a common daily life activity; when the load is carried unilaterally, it disrupts the body's left–right symmetry and may negatively impact gait stability. Numerous studies have examined the effects of asymmetric load carrying on dynamic stability during walking. For instance, Liu et al.[1] reported that by increasing the carried load, the maximum Lyapunov exponent (maxLE) is significantly elevated, indicating reduced local dynamic stability. Wang et al.[2] demonstrated that carrying a load equivalent to 20% of body weight in one hand increases the velocity of the medial-lateral center of pressure, suggesting a compromise in balance control. DeVita et al.[3] observed that asymmetric load carriage increases hip moments and trunk tilt, thereby disturbing gait symmetry through compensatory postural adjustments. However, these studies employed evaluation metrics limited to segmental measures, which are insufficient for capturing whole-body movement patterns. Therefore, this study focuses on the margin of stability (MoS) as an indicator, which is capable of comprehensively assessing dynamic walking stability over time. MoS is defined based on the relationship between the extrapolated center of mass (XCoM) and the base of support (BoS), enabling quantitative evaluation of stability at specific phases within the gait cycle [4]. Importantly, MoS visualizes the extent to which XCoM deviates from BoS, rendering it well-suited to capture changes in whole-body balance control under asymmetric loading conditions (Fig. 1(b)).

During normal walking, frontal-plane stability is maintained through appropriate regulation of the relationship between the body's center of mass (CoM) and lateral boundary of BoS[5]. This regulation can be achieved either by adjusting the foot placement of the swing leg or by controlling the movement of CoM through the stance leg [6]. A similar movement pattern is presumed to occur when a load is carried unilaterally (Fig. 1(c)). Foot placement adjustment requires only the movement of the swing leg mass and is considered to impose a lower de-

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mand on motor control when compared with regulating the CoM through the stance leg. Conversely, CoM control via the stance leg is presumed to be more complicated, as the CoM remains displaced toward the load-bearing side. Consequently, adjustments in foot placement are critical for maintaining stability under unilateral load carrying. MoS dynamically reflects the interplay between the center of mass position and foot contact location, and thereby, serves as an effective metric for evaluating stability during unilateral load carrying.

This study aims to quantitatively evaluate the impact of unilateral load carrying on gait stability using MoS and elucidate its relationship with foot placement. Specifically, we focus on the moment immediately preceding foot contact of the swing foot, where variations in MoS due to load are most pronounced. This enables us to investigate the relationships among carried weight, MoS, CoM movement patterns, and individual differences. Furthermore, we explore stabilization strategies during unilateral load carrying by considering CoM regulation mediated by trunk motion and adjustments in foot placement.

The primary contributions of this study are as follows:

- We demonstrated pronounced inter-individual differences in the tendency of MoS to change with varying load conditions, providing a conceptual framework for understanding stability regulation during walking while carrying a unilateral load.
- We identified multiple CoM displacement patterns underlying these individual differences and characterized the distinctive control mechanisms associated with each.
- We revealed that increasing load magnitude may induce a transition from strategies that exploit passive center-of-gravity displacement to those involving active trunk movement.

II. EXPERIMENTAL METHODS

A. Purpose of the Experiment

The objective of this experiment was to elucidate the impact of carrying a load with one hand on walking dynamics. Specifically, changes in movement characteristics with variations in load weight were validated by establishing multiple load conditions. Additionally, the walking motions under various conditions were compared and analyzed. This experiment was conducted with the approval of the Experimental Ethics Committee of the Faculty of Engineering, Kyushu University (Approval No.: 2025-04).

B. Experimental Setup

The experiment was conducted in the environment shown in Fig. 1(a). A split-belt instrumented treadmill (ITR5018-11, Bertec Corporation) was employed. In this case, walking on one side of the belt was utilized to prevent the widening of stride, which may occur when traversing both belts, thereby ensuring that the natural walking patterns of the participants remained unimpeded. The belt speed was calibrated to 0.9 m/s, a velocity validated in preliminary assessments to be comfortable for all participants in the experiment. Notably, all participants independently selected the same speed during these assessments.

Equipped with a force plate, the treadmill facilitated the acquisition of ground reaction force data, which was instrumental in determining the foot contact timing and extracting the gait cycle. Motion measurement was conducted using an optical motion capture system (OptiTrack, NaturalPoint, Inc.). Participants donned specialized motion capture suits, with infrared reflective markers affixed to each joint and trunk. This enabled the acquisition of three-dimensional motion data through the use of seven infrared cameras. Consistency in gaze direction across trials was maintained by displaying a visual target on a monitor positioned in front of the participants. They were instructed to focus on this target while walking.

C. Experimental Participants

Three healthy young males participated in the experiment. Participant A was 23 years old, 172 cm tall, and weighed 58 kg; Participant B was 21 years old, also 172 cm tall, and weighed 52 kg; and Participant C was 22 years old, measured 171 cm in height, and weighed 57 kg. Prior to participation, all participants were informed of the purpose and procedures of the experiment and provided their consent.

D. Experimental Conditions

In addition to normal walking without carrying any load, five conditions were set where participants walked while holding a bag containing weights of 5, 10, 15, and 20 kg in their right hand. In each condition, participants completed two 10-s walking trials on the treadmill. The acquired data were divided into gait cycles, and 11–13 cycles were analyzed for each condition. Gait cycles containing missing data due to marker loss or tracking errors were excluded from the analysis. To mitigate the effects of fatigue on performance, a minimum rest period of 3 min was provided between each trial.

E. Analysis Method

In this study, MoS in the frontal plane was utilized as a key indicator to evaluate dynamic stability during walking. The MoS is based on the theoretical framework proposed by Hof et al. [4]. The XCoM is calculated using the position (x_{ML}) and velocity ($\dot{x}_{\mathrm{ML}}$) of the CoM of the combined system, including the body and load. In this study, CoM is defined as the center of gravity of the entire system. Furthermore, MoS is then computed based on the relative distance between XCoM and BoS.

First, the CoM position for each body segment was determined using the three-dimensional coordinates of multiple markers affixed to the body. The CoM positions

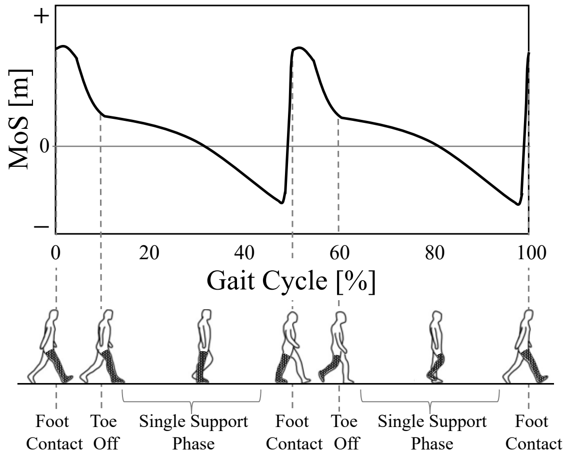


Fig. 2: Example of MoS during a gait cycle

and weight ratios for each segment were obtained using the Matsui measurement results [7]. Furthermore, the CoM position of the load held in the hand was measured using markers. By incorporating the mass of this load, we calculated the overall frontal plane CoM $x_{\mathrm{ML,\,total}}$ for the body and load combined. This calculation was performed using the following equation.

$$x_{\mathrm{ML,\,total}} = \frac{\sum_{i=1}^{n} m_i \, x_{\mathrm{ML},i} + m_{\mathrm{load}} \, x_{\mathrm{ML,\,load}}}{\sum_{i=1}^{n} m_i + m_{\mathrm{load}}} \tag{1}$$

where m_i denotes the mass of body segment i, $x_{\mathrm{ML},i}$ denotes its center of gravity position in the frontal plane, m_{load} denotes the mass of the load, and $x_{\mathrm{ML,\,load}}$ denotes the center of gravity position of the load.

Furthermore, XCoM was calculated based on the center of gravity position and its velocity as follows:

$$\mathrm{XCoM_{ML}} = x_{\mathrm{ML,\,total}} + \frac{\dot{x}_{\mathrm{ML,\,total}}}{\sqrt{g/l}} \tag{2}$$

where $\dot{x}_{\mathrm{ML,\,total}}$ denotes the velocity of the body and load in the frontal plane, g denotes gravitational acceleration ($9.81\,\mathrm{m\,s^{-2}}$), and l denotes the leg length, which was measured for each experimental participant using motion capture.

BoS is defined as a polygon formed by three markers affixed to the standing foot (heel, first metatarsal head, and fifth metatarsal head). The outer edges of this polygon in the lateral direction are considered as the boundaries of the BoS. The MoS is defined as the distance between XCoM and outer edge of the BoS. This distance is calculated using the side of the BoS that is closest to the XCoM among its left and right ends as follows:

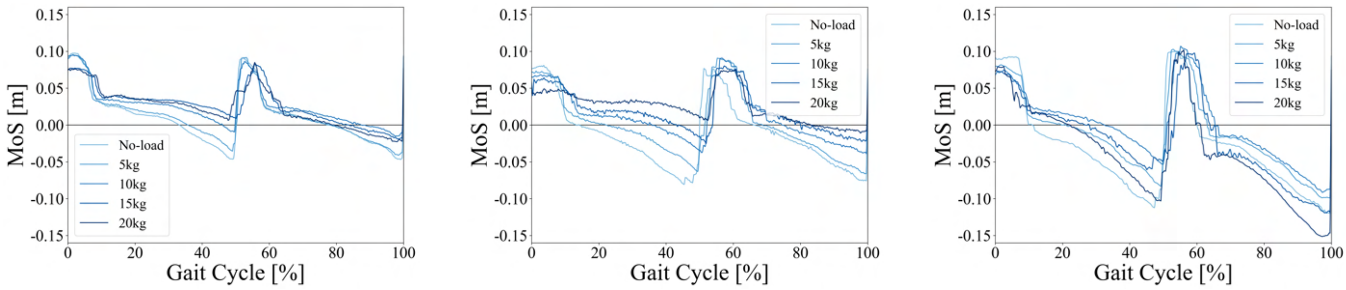
$$\begin{aligned} \mathrm{MoS} = \min\bigl(&\mathrm{XCoM_{ML}} - x_{\mathrm{BoS,\,left}}, \\ &x_{\mathrm{BoS,\,right}} - \mathrm{XCoM_{ML}}\bigr) \end{aligned} \tag{3}$$

where $x_{\mathrm{BoS,\,left}}$ and $x_{\mathrm{BoS,\,right}}$ denote the coordinates of the left and right endpoints of BoS, respectively. An example of the MoS during a gait cycle is shown in Fig. 2. This figure illustrates the relationship between XCoM and BoS in the mediolateral direction during walking. A positive MoS indicates that XCoM is situated within the BoS, whereas a negative MoS signifies that XCoM is positioned outside the BoS.

III. EXPERIMENTAL RESULTS

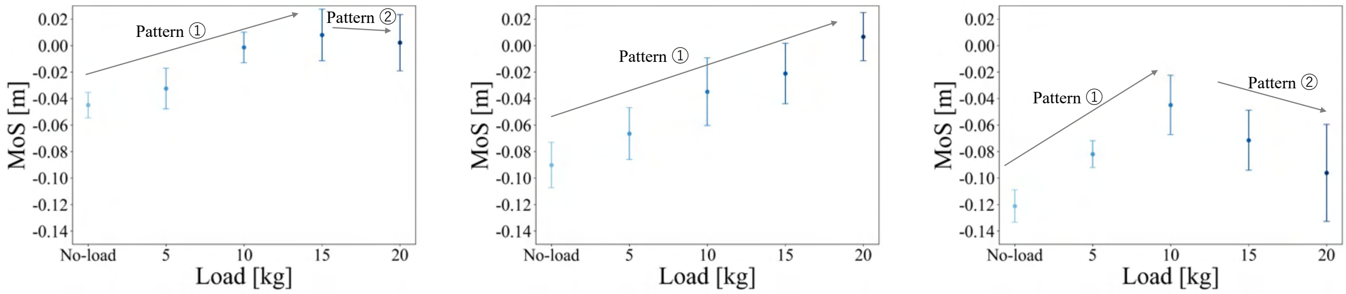
Figure 2 shows an example of MoS over a gait cycle, starting from the initial contact of the right foot to the subsequent right foot contact. The walking cycle was normalized such that the moment of right foot contact was designated as 0%. The subsequent right foot contact occurred at 100%. The maxima observed near 0% and 50% corresponded to the grounding of the right and left feet, respectively. Conversely, the minima near 10% and 60% corresponded to the toe-off moments of the right and left feet, respectively. Intervals 10–50% and 60–100% represent the single-leg support phase of the right and left feet, respectively.

The results for participants A, B, and C are shown in Fig. 3(a), (b), and (c), respectively. The load conditions are represented using darker colors to signify higher loads. First, we examined the condition where no load was carried. At the moment of foot contact, the MoS reached its local maximum value, attributed to the expansion of the BoS. Conversely, at the moment of toe-off, the MoS decreased as the BoS contracted. As the participant transitioned into the single-leg support phase, the MoS decreased over time, ultimately becoming negative owing to the shift in the CoM toward the next



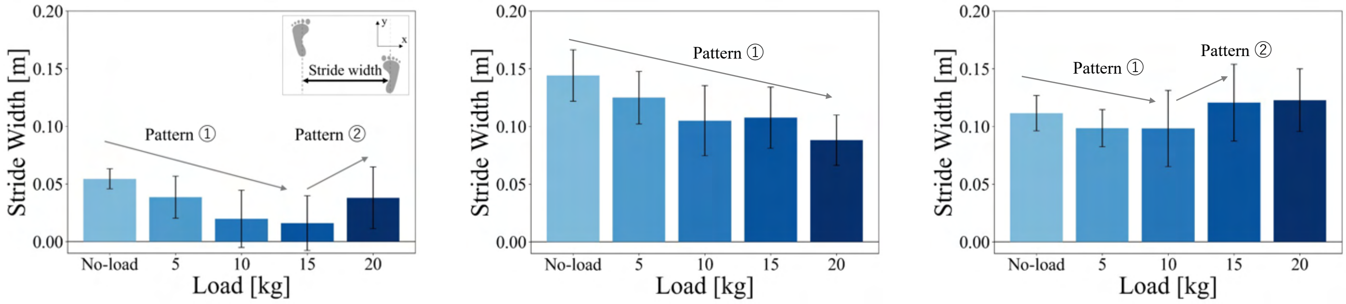
(a) Participant A
(b) Participant B
(c) Participant C

Fig. 3: MoS during a gait cycle: (a) Participant A, (b) Participant B, and (c) Participant C



(a) Participant A (b) Participant B (c) Participant C

Fig. 4: MoS immediately before left foot contact under each handheld load condition: (a) Participant A, (b) Participant B, and (c) Participant C



(a) Participant A (b) Participant B (c) Participant C

Fig. 5: Stride width under each handheld load condition: (a) Participant A, (b) Participant B, and (c) Participant C

step. This decline reached its minimum just prior to foot contact, indicating that the MoS was at its lowest point in the gait cycle just before foot contact. These observed trends align with findings from previous studies[8][9].

Next, we examined the impact of increased load on the MoS. At the moment of foot contact, a tendency for MoS to decrease with increasing load was observed. During the single-leg support phase, a tendency for MoS to increase was observed under load conditions of up to approximately 10 kg; however, the results varied among participants at higher loads. Notably, the difference in MoS owing to varying load conditions was particularly significant just before left foot contact. In this experiment, the left foot served as the non-weight-bearing side, and this timing represents the final stage where adjustments to the foot contact position of the swing leg were reflected. Therefore, this phase is critical for understanding the motor adjustments involved in maintaining stability while carrying an object. Therefore, this study focused on MoS immediately prior to left foot contact. Furthermore, the impacts of varying loads were examined on shifts in CoM during walking. The relationship between MoS and load immediately before left foot contact is shown in Fig. 4. Two distinct patterns emerged from the MoS change pattern (Pattern①, ②). Pattern① encompassed conditions up to 15 kg for Participant A (Fig. 4(a)), all conditions for Participant B (Fig. 4(b)), and conditions up to 10 kg for Participant C (refer to

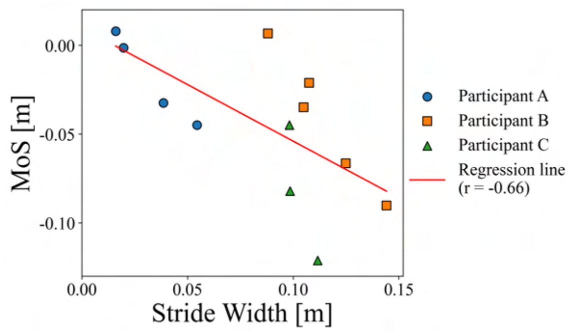


Fig. 6: Correlation between MoS and stride width

Fig. 4(c)). In this pattern, MoS increased with load. In contrast, Pattern② included Participant A's 20-kg condition and Participant C's conditions above 15 kg; no clear increase trend, such as that in Pattern①, was observed. In particular, MoS decreased for Participant C's conditions exceeding 15 kg.

This difference may be due to adjustments in the foot contact position, specifically changes in the stride width. In fact, when stride width was examined across load conditions (Fig. 5), it decreased with increasing load in Pattern①, where an increase in MoS was observed. Furthermore, analysis of the correlation between MoS and stride width (Fig. 6) revealed a strong association between the two. This suggests that foot contact position

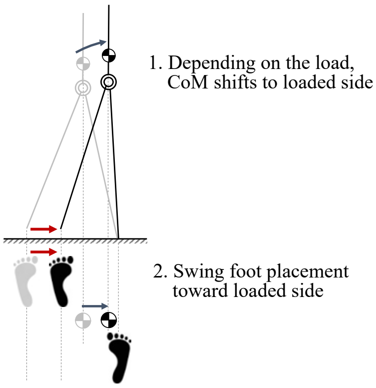


Fig. 7: Link model representation for Pattern①

adjustment played a dominant role in conditions where MoS increased. Conversely, in Pattern②, where MoS did not increase with increasing load, the stride length was larger when compared with that in Pattern①. This discrepancy can be attributed to CoM control via the support leg, as well as adjustments made to the foot contact position. Based on the aforementioned results, it is suggested that in unilateral load carrying, foot contact position adjustment is primarily performed under moderate load conditions. However, under high load conditions, large individual differences are observed, where CoM control, via the supporting leg, is used in addition to foot contact position adjustment.

IV. DISCUSSION

A. Classification of Movement Patterns

This section examines two distinct CoM adjustment patterns (Pattern① and Pattern②) observed during unilateral load carrying. The first pattern (Pattern①) is characterized by an increase in MoS as the load increases. When carrying a load unilaterally, the CoM of the body shifts toward the load-bearing side. Under conditions of relatively light loads, the MoS is primarily maintained through adjustments in foot placement, with minimal muscle activation or trunk movement. This phenomenon is illustrated using the inverted pendulum model (Fig. 7), which demonstrates that when a load is carried in the right hand, the CoM shifts to the right. Furthermore, owing to the geometric constraint of maintaining a constant limb length, the left foot placement also shifts rightward, resulting in the observed decrease in stride length.

The second pattern (Pattern②) is characterized by a lack of increase in MoS despite the increasing load. Pattern② showed no increase in stride length with increasing load, suggesting that alternative strategies, such as trunk movement, may be employed to adjust CoM. Fig. 8 presents postural changes under loaded conditions for Participants A and C, in whom both patterns were observed. Although no marked postural changes were evident in Pattern①, a pronounced lateral trunk tilt was observed in Pattern②. Trunk inclination toward the load-bearing side was observed in participant A, whereas participant C inclined the trunk toward the side opposite to the load. To quantitatively examine this difference, we compared the trunk tilt angles across load conditions and observed that participant A exhibited a greater tilt angle toward the loaded side in Pattern② compared with Pattern① (Fig. 9a). Conversely, Participant C demonstrated minimal variation in the mean tilt angle between the two patterns; however, the standard deviation was larger in Pattern②, indicating increased variability in trunk motion throughout the gait cycle (Fig. 9b). This variability included instances of significant tilt toward the unloaded side. These movements can be effectively modeled using a two-link framework, which extends the inverted pendulum by incorporating additional links for the hip and trunk (Fig. 10). In both cases, CoM shifted toward the unloaded side, which in turn was accompanied by a shift in the foot contact position toward the same side, resulting in an increase in stride width during the transition from Pattern① to Pattern②.

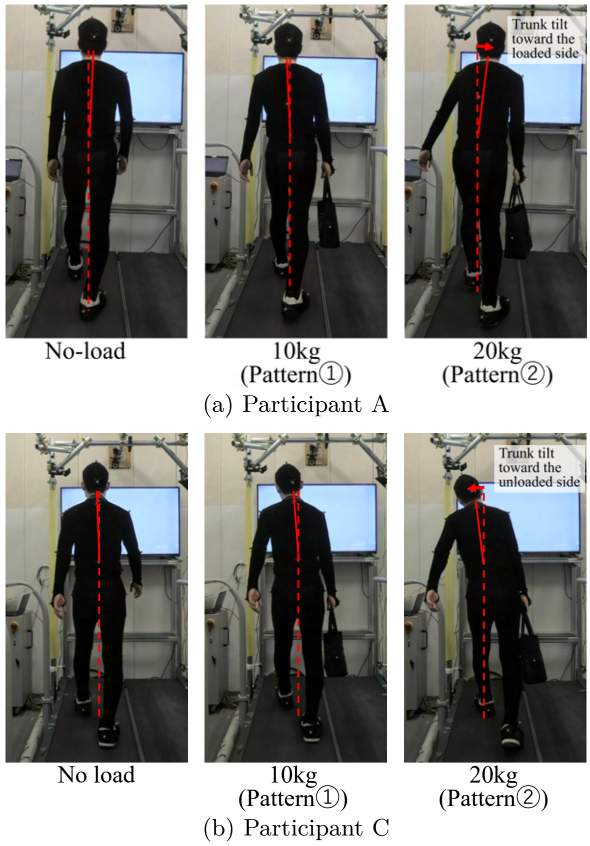
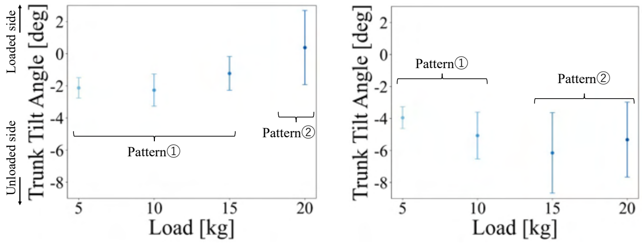


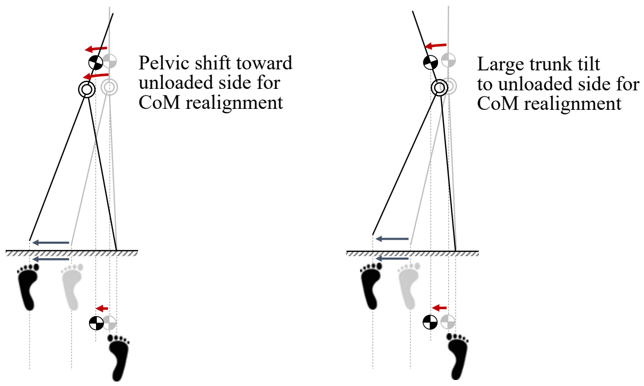
Fig. 8: Postural changes and classification of movement patterns observed in (a) Participant A and (b) Participant C

These findings suggest that adjustments in foot placement (Pattern①) and CoM regulation through trunk tilt (Pattern②) may be selectively employed based on the magnitude of the load. Human locomotion is generally understood to minimize energy expenditure[10].



(a) Participant A (b) Participant C

Fig. 9: Trunk tilt angle immediately before left foot contact under each handheld load condition: (a) Participant A and (b) Participant C



(a) Trunk tilt to the loaded side (Participant A) (b) Trunk tilt to the unloaded side (Participant C)

Fig. 10: Link model representation for Pattern②

Adjustments that primarily involve foot placement can passively accommodate CoM displacement and are presumed to require relatively low energy. In contrast, movements that necessitate active repositioning of the trunk and pelvis directly manipulate the CoM and are likely associated with higher energy costs. Therefore, we can assume that strategies focused on foot placement are preferred under lighter loads, whereas trunk-mediated strategies are adopted when the load exceeds a certain threshold. Furthermore, variations in trunk tilt direction likely indicate alternative mechanisms for alleviating shoulder joint loading. Tilting toward the load-bearing side may decrease shoulder torque by aligning the upper limb more closely with the vertical axis. Conversely, tilting toward the unloaded side may facilitate load redistribution by enabling contact between the upper arm and trunk. These strategy choices are believed to be influenced by individual factors, such as muscle strength and anthropometric characteristics.

B. Limitation

The two movement patterns identified in this study may not be universally applicable across all individuals. For example, individuals with differing muscle strength, balance ability, or flexibility may adopt alternative strategies even under identical load conditions. Such individual differences may influence trunk and pelvic kinematics, potentially producing movement characteristics that are not fully explained by the two patterns described here. Therefore, the movement strategies presented in this study cannot be generalized to all conditions.

V. Conclusion

In this study, we quantitatively evaluated the impact of unilateral load carrying on dynamic stability during walking by utilizing MoS. Furthermore, we investigated the roles of foot contact position adjustment and trunk movement. Our findings indicated that under moderate load conditions, MoS was predominantly maintained via adjustments in foot contact position. Conversely, under high load conditions, we observed significant individual variability, with a pattern of CoM repositioning facilitated by trunk movement. Additionally, our analysis suggested that variations in trunk tilt direction may be associated with different strategies for reducing shoulder joint load. These insights are anticipated to enhance our understanding of the movement mechanisms that underpin stability maintenance while carrying luggage. Additionally, they hold promise for informing the design of assistive devices tailored to individual characteristics.

Future research should broaden the range of participant demographics by including older adults and individuals with physical limitations. This will enable investigation on the effects of different individual characteristics on stabilization strategy selection. Moreover, data should be collected under more varied experimental conditions to systematically assess the applicability and limitations of the two movement patterns identified in this study.

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