https://doi.org/10.31489/3081-0531/2025-1-2/4-14
UDC 612.76:531.36:616-053.9 Received: 14.08.2025 | Accepted: 07.10.2025 | Published online: 30 December 2025

Zelong Gai[1], Zhaoli Meng[2*], Haibin Liu[3]

[1, 2, 3]Dalian University of Technology, School of Sport and Health Sciences, Dalian, China
([*]Corresponding author email:mengzl@dlut.edu.cn)

[1]ORCID 0009-0001-8297-476X
[2]ORCID 0000-0002-9338-305X
[3]ORCID 0000-0001-9197-4959

A Systematic Review of the Biomechanical Impact of Load Carriage on Gait in Older Adults

Objective: To examine the biomechanical effects of load carriage on gait patterns, joint kinematics, and muscle activity during walking in older adults. Methods: A systematic literature search was conducted across five databases (CNKI, Wanfang, VIP, PubMed, and Web of Science) through June 2025. Eight studies met the inclusion criteria. The methodological quality of included studies was assessed using the ROBINS-I Version 2 tool. Results: Asymmetrical load carriage during walking increases step frequency and step width, shortens step length and the gait cycle, induces lateral trunk tilt, and leads to asymmetric muscle activation between body sides. With increasing load, adverse effects on trunk posture and muscle activation become more pronounced, including a significant increase in contralateral hip joint torque. Symmetrical load carriage up to 5% of body weight has no significant effect on gait and may improve static postural stability in older adults. Conclusion: Both asymmetrical and heavier load carriage impose greater biomechanical demands on gait in older adults. Older adults are advised to carry loads symmetrically and keep the weight below 5% of body mass to maintain gait stability and reduce fall risk.

Keywords: Load Carriage, Older Adults, Gait, Systematic Review, Biomechanics, Dynamic Stability, Fall Risk

Introduction

The aging process causes a progressive decline in physical functions, including motor ability and neurological integrity, thereby significantly increasing fall risk among older adults [1]. Globally, falls are a significant public health concern, causing approximately 684,000 deaths annually and ranking as the second leading cause of unintentional injury-related mortality [2]. Robinovitch et al. reported that 24% of all falls in older adults occurred during normal forward walking—the highest proportion among all fall scenarios [3].

Various factors contribute to fall risk during walking, including motor ability, psychological state, and environmental conditions [4]. Load carriage constitutes an environmental modification and is a routine aspect of daily life for many older adults. Carrying external loads shifts the body's center of mass by introducing additional external forces. To maintain balance, individuals must adjust their gait and posture accordingly. Consequently, the biomechanical characteristics of load-carrying gait differ significantly from those of unloaded walking [5].

There are currently divergent perspectives regarding the effects of load carriage on gait in older adults. One perspective posits that load carriage compromises gait stability in older adults. For example, Nagaraja et al. found that during load-bearing walking, the trunk and pelvis deviate from the neutral position, increasing asymmetry in frontal and transverse plane movements and disrupting the rhythmic coordination between trunk and pelvic motion [6]. These biomechanical changes clearly impair the ability of older adults

Buketov University

to maintain stable gait. In contrast, an alternative viewpoint argues that load carriage may not negatively affect gait in older adults and could even enhance postural stability during quiet standing [7].

Load carriage during walking is sometimes unavoidable for older adults, such as when carrying items from routine shopping trips. Therefore, understanding the biomechanical alterations induced by load carriage is crucial for reducing fall incidence and promoting both physical and mental well-being in later life.

Existing research on gait in older adults has primarily focused on walking under cognitive load, whereas empirical evidence concerning walking under external physical load remains limited. Although studies on load-bearing gait in older adults are scarce, those available vary in testing tools, evaluation metrics, and measurement dimensions, resulting in inconsistent findings.

Methods and materials

This review was conducted and reported following the PRISMA (Preferred Reporting Items for Systematic Reviews and Meta-Analyses) guidelines [8].

Search Strategy

A comprehensive literature search was conducted across five databases—PubMed, Web of Science, CNKI, Wanfang, and VIP—with the search period extending through June 2025. The search terms included combinations of the following keywords: “older”, “load”, “carrying”, “bags”, “weight-bearing”, “walking”, “gait”, “biomechanics”, and “biomechanical”. Using PubMed as an example, the detailed search strategy was as follows: ((older[Title/Abstract]) AND (load[Title/Abstract] OR carrying[Title/Abstract] OR bags[Title/Abstract] OR weight-bearing[Title/Abstract])) AND (walking[Title/Abstract] OR gait[Title/Abstract] OR biomechanics[Title/Abstract] OR biomechanical[Title/Abstract])

The search was restricted to controlled experimental studies published in peer-reviewed academic journals in either Chinese or English.

Inclusion and Exclusion Criteria

Inclusion Criteria:

(1) Participants were healthy older adults aged 60 years or older.

(2) Participants walked in a straight line, at a self-selected speed, within a defined experimental setting while facing forward throughout the task.

(3) Outcome measures included, but were not limited to:

(a) kinematic indicators (e.g., joint angles, center of pressure displacement, and center of mass sway amplitude);

(b) kinetic indicators (e.g., joint moments, ground reaction forces (GRF), and joint power);

(c) gait parameters (e.g., step length, stride length, gait cycle, step width, cadence, gait speed, and gait variability coefficient).

Exclusion Criteria:

(1) Participants were older adults diagnosed with medical conditions such as Parkinson’s disease, Alzheimer’s disease, or hypertension.

(2) Studies in which the intervention was unrelated to load carriage or involved non-standard walking tasks.

(3) Studies published in languages other than Chinese or English; conference abstracts, retrospective analyses, reviews, or those with inaccessible full texts.

(4) Duplicate publications or those assessed as having low methodological quality. (5) Studies lacking extractable or usable data.

Study Selection and Data Extraction

All retrieved records were imported into EndNote X9 for reference management.

(1) Duplicate records were identified and removed.

(2) Titles and abstracts were screened to exclude irrelevant studies.

(3) Full texts of potentially eligible articles were reviewed to exclude studies not meeting the inclusion criteria.

(4) Remaining eligible studies were included in the final analysis.

Extracted data included article title, first author, year of publication, country of origin, participant characteristics (age, sex, height, and weight), sample size, study design and grouping, and outcome measures.

Buketov University

Quality Assessment of Included Studies

The methodological quality of the included studies was evaluated using the ROBINS-I Version 2 tool [9].

The tool evaluates bias risk across seven domains: confounding, classification of interventions, participant selection, deviations from intended interventions, missing data, outcome measurement, and selection of reported results. Each domain is assessed using a structured set of signaling questions designed to guide risk-of-bias judgments [10].

Results and Discussion

Results of Literature Search

A total of 2,437 records were retrieved from five databases: CNKI (n = 114), Wanfang (n = 19), VIP (n = 6), PubMed (n = 615), and Web of Science (n = 1,683). An additional three studies were identified through manual reference screening. After removing 177 duplicates, 2,263 unique records remained for screening. After title and abstract screening, 2,196 records were excluded, leaving 67 articles for full-text review. Following full-text review, 59 articles were excluded, and eight studies were included in the final analysis (Bampouras et al., 2016; Narouei et al., 2023; Allahverdipour et al., 2021; Badawy et al., 2019; Walsh et al., 2018; Kong et al., 2014; Matsuo et al., 2008; Tengyu et al., 2018) [7, 11–16].

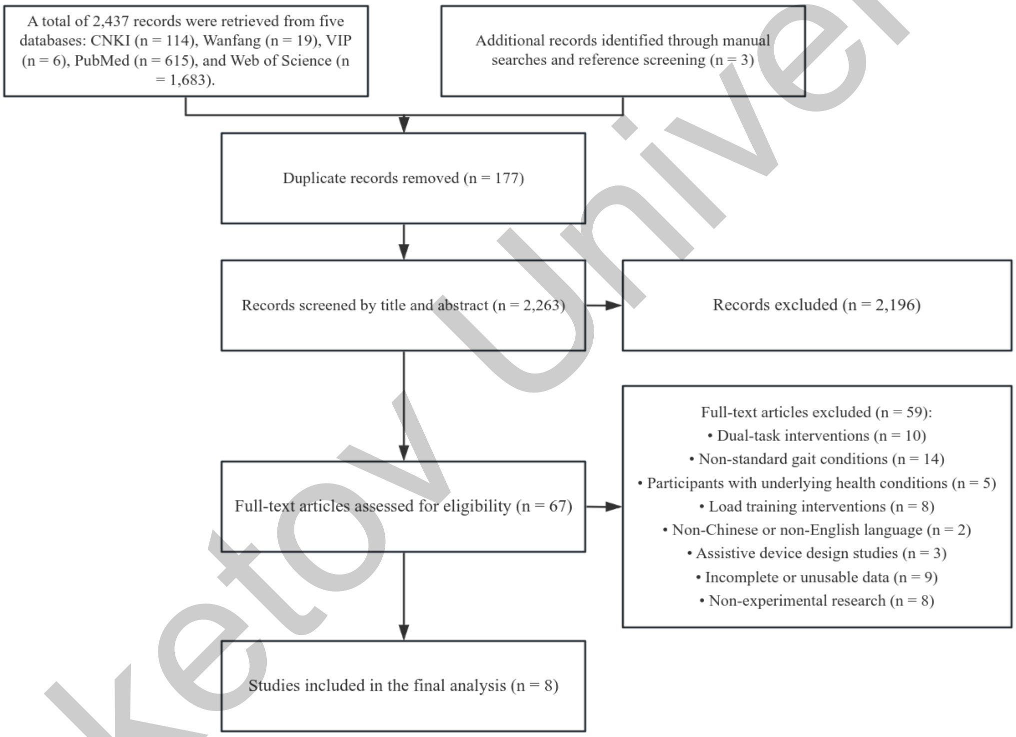

Figure 1. Flow Diagram of the Study Screening Process

Risk of Bias Assessment Results

According to the ROBINS-I Version 2 assessment criteria, Walsh et al. and Kong et al. clearly reported inclusion and exclusion criteria and effectively controlled for potential confounding factors. Consequently, the risk of bias due to confounding was judged to be low. Allahverdipour et al. (2021) and Kong et al. (2014) clearly described the inclusion procedures and participant recruitment methods, ensuring appropriate participant selection [12, 14, 15]. As a result, the risk of bias in participant selection was rated as low. However, both studies employed measurement tools susceptible to inaccuracies, resulting in a moderate risk of bias in outcome assessment. In all eight included studies, intervention conditions were clearly defined and appropriately classified. Participants adhered strictly to the assigned interventions, with no notable deviations from the intended procedures. All relevant outcomes were reported and supported by appropriate statistical validation. Accordingly, the risks of bias related to intervention classification, deviations from intended interven-

Buketov University

 ISSN 3081-0531 (Print) ISSN 3081-054X (Online)

tions, missing data, and selective outcome reporting were all judged to be low. Overall, the included studies were assessed to have a moderate risk of bias, which is considered acceptable for inclusion in this review.

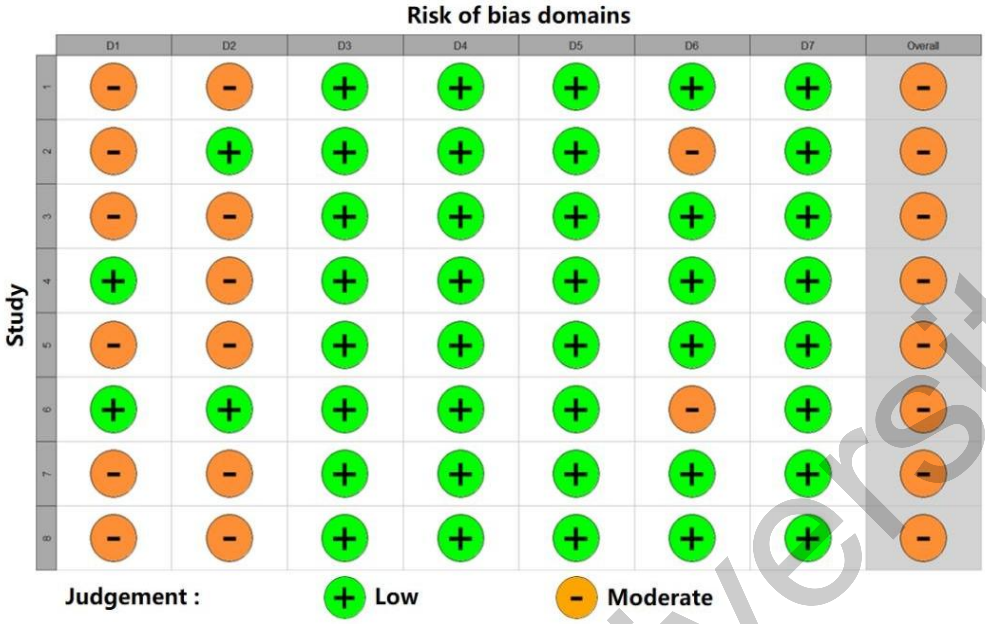

Figure 2. Risk of Bias Assessment Across Included Studies Using ROBINS-I Tool

Note: D1 = Risk of bias due to confounding; D2 = Risk of bias in selection of participants into the study; D3 = Risk of bias in classification of interventions; D4 = Risk of bias due to deviations from intended interventions; D5 = Risk of bias due to missing data; D6 = Risk of bias arising from measurement of the outcome; D7 = Risk of bias in selection of the reported result. 1: Narouei, S., 2023; 2: Allahverdipour, H., 2021; 3: Badawy, M., 2019; 4: Walsh, G. S. (2018); 5: Bampouras, T. M., 2016; 6: Kong, P. W., 2014; 7: Matsuo, T., 2008; 8: Zhang, T., 2018

Basic Characteristics of Included Studies

Table 1 summarizes the basic characteristics of the included studies. All eight studies were published between 2008 and 2023. All eight studies were published between 2008 and 2023. The studies originated from the United Kingdom (n = 2), Japan (n = 2), and one each from the United States, China, Singapore, and Iran. Collectively, the studies involved 278 healthy older adults. Regarding load carriage methods, four studies investigated bilateral hand-carrying, four involved unilateral carrying, two involved backpack-style carrying, one involved lower-limb loading, one included pushing and pulling carts, and one assessed front-carrying (e.g., cradling). Five studies implemented symmetrical loading, while another five assessed asymmetrical (unilateral) loading patterns.

Table 1

Summary of General Characteristics of the Included Studies

First Author	Year of Publication	Country	Sample Size (M/F)	Age (Years)	Height	Weight (kg)	Load Type (Weight)
Narouei, S.	2023	Japan	29(7,22)	67.96±6.86	157.63±8.48cm	53.78±9.04	No load; symmetrical load (2% body weight)
Allahverdipour, H.	2021	Iran	42(21,21)	≥60	Not reported	Not reported	No load; symmetrical loads (2, 4, 6 kg)

Buketov University

Continuation of Table 1

First Author	Year of Publication	Country	Sample Size (M/F)	Age (Years)	Height	Weight (kg)	Load Type (Weight)
Badawy M.	2019	USA	20(20,0) 5(5,0)	59.7±3.5	175.3±4.4cm	70.2±6.6	No load; asymmetrical load (5.67 kg, 10.21 kg)
Walsh, G. S.	2018	UK	14(7,7)	65.0±6.0	1.70±0.10m	74±13	No load; stable/unstable asymmetrical load (15% body weight)
Bampouras, T. M.	2016	UK	19(0,19) 9(0,9)	71.0±6.0	1.65±0.06m	66.3±10.1	No load; asymmetrical loads (1.5 kg, 3 kg); symmetrical loads (3 kg, 6 kg)
Kong, P. W.	2014	Singapore	52(24,28) 32(14,18)	69.4±7.0	1.57±0.08m	62.5±10.0	No load; pushing load (10 kg), pulling load (15 kg); symmetrical load (4 kg)
Matsuo, T.	2008	Japan	11(0,11) 6(0,6)	59.7±1.4	1.61±0.06m	56.8±4.7	No load; single-hand carrying (3 kg, 8 kg)
Zhang, T.	2018	China	15(8,7)	71.43±4.72	Not reported	Male: 70.21 ± 6.10 kg; Female: 61.02 ± 9.47 kg	No load; symmetrical/asymmetrical load (10% body weight)

All included studies employed an experimental design.

Selection of Research Instruments and Outcome Measures

The research instruments and outcome measures employed in the included studies are summarized in Table 2. Four studies employed three-dimensional motion capture systems; three utilized surface electromyography (sEMG); one used a high-speed camera with video analysis software; one implemented the Optojump system; and one relied on a stopwatch. The outcome measures were primarily categorized into gait parameters, joint kinematics, muscle activity, and miscellaneous variables. In terms of gait parameters, walking speed was assessed in three studies; cadence, stride time (ST), and total double support (TDS) were each reported in two studies; while step length (SL), step width (SW), coefficient of variation (CoV), and step asymmetry (SA) were each evaluated in one study. For joint kinematics, the angles of the hip, knee, ankle, and trunk joints were measured in two studies; center of pressure (CoP) displacement and peak hip abduction torque were each reported in one study. The Timed Up and Go (TUG) test duration was measured in one study.

Buketov University

ISSN 3081-0531 (Print) ISSN 3081-054X (Online)

Table 2

Measurement Instruments and Outcome Variables in the Included Studies

Included Study	Instrument	Outcome Measures			
		Gait Parameters	Joint Kinematics	Muscle Activity	Others
Narouei, S., 2023	A three-dimensional motion capture system, and surface electromyography (sEMG) system	Cadence, and toe off timing	Hip, knee, and ankle joint angles	Normalized Average Value of EMG： rectus femoris (RF)	
Allahverdipour, H., 2021	Stopwatch				Timed Up and Go (TUG)
Badawy M., 2019	Surface electromyography (sEMG) system	Speed		Average and peak % maximum voluntary contraction (MVC)：left/right rectus abdominis (RA), left/right external oblique (EO), left/right internal oblique (IO), left/right latissimus dorsi (LD), left/right upper erector spinae (UES), and left/right lower erector spinae (LES)	
Walsh, G. S., 2018	A three-dimensional motion capture system, and surface electromyography (sEMG) system	Step width (SW), and stride time (ST)	Hip, knee, and ankle joint angles in the sagittal, frontal, and transverse planes	Mean electromyographic (EMG) activity：RF, vastus medialis (VM), biceps femoris (BF), tibialis anterior (TA), gastrocnemius medialis (GM), and soleus (SOL)	
Bampouras, T. M., 2016	Optojump, and treadmill	Stride length (SL), coefficient of variation (CoV), total double support (TDS), step asymmetry (SA), and gait stability Ratio (GSR)			
Kong, P. W., 2014	High-speed video camera, video analysis software, and timing gates	Start-up time, and speed			
Matsuo, T., 2008	A three-dimensional motion capture system, and three-dimensional force plate		Maximum trunk/head lateral flexion to contralateral side, maximum upper arm elevation for contralateral side, contralateral/ipsilateral maximum hip abduction torque, and continuous relative phase (CRP)		
Zhang, 2018	A three-dimensional motion capture system, and three-dimensional force plate	Cadence, speed, SL, ST, stance phase (SP), and COV	Center of pressure (CoP) displacement		

Buketov University

In experimental research in sports and human movement sciences, researchers typically select appropriate instruments and precise outcome measures based on study objectives, equipment accuracy, reliability, and cost, as well as the demands of data processing and analysis.

In this study, various technologies used to measure kinematic parameters operate on a shared fundamental principle: capturing the trajectories of anatomical landmarks to calculate motion-related variables. Their primary differences lie in technological sophistication and system evolution. Historically, motion capture began with Eadweard Muybridge's pioneering use of sequential photography to capture the dynamic movement of galloping horses—an approach later widely adopted for studying human locomotion. With the advent and widespread adoption of video technology, researchers began employing high-speed cameras and camcorders to record gait behavior. By analyzing sequential frames, they extracted spatiotemporal characteristics of movement. Kong et al. used high-speed video cameras and video analysis software to measure participants' initiation time and gait speed [15]. The emergence of digital and sensor technologies has enabled optical motion capture systems to record human walking trajectories with significantly enhanced precision. Bampouras et al. employed the Optojump system to measure step length (SL) and total support duration (TSD) during treadmill walking in older adults [7]. Similarly, Narouei et al. utilized a three-dimensional motion capture system to acquire gait parameters and joint angles [11]. The key distinction between the two systems lies in data dimensionality. The Optojump system measures temporal parameters by detecting light interruptions between emitter and receiver bars, enabling the assessment of contact time, flight time, and stride timing during walking, running, and jumping. In contrast, a three-dimensional motion capture system incorporates a Z-axis, enabling the capture of reflective marker trajectories on joints to derive multi-planar joint angles at the hip, knee, ankle, and trunk.

Among the included studies, three employed similar methodologies to measure lower-limb sEMG signals during load-carrying walking in older adults. However, due to differing experimental objectives, the selected muscles and evaluation metrics varied slightly. All three studies included the rectus femoris (RF) as a target muscle. Badawy et al. additionally included muscles such as the external oblique (EO), internal oblique (IO), and latissimus dorsi (LD) to investigate trunk muscle activation during movement [13]. Walsh et al. focused on comprehensive lower-limb muscle activity and thus included a broader array of lower-extremity muscles in their analysis. Regarding evaluation metrics, Narouei et al. and Walsh et al. [11, 14] conducted time-domain analysis of sEMG to calculate the average electromyographic amplitude, reflecting the mean activation levels of the measured muscles. In contrast, Badawy et al. normalized the EMG values of each muscle to the percentage of maximum voluntary contraction (MVC), emphasizing the relative intensity of maximal muscle recruitment [13].

Biomechanical Effects of Various Load-Carrying Methods on Gait in Older Adults

In the included studies, researchers frequently designed a range of load-carrying conditions to comprehensively assess the effects of load carriage on gait in older adults. Load carriage methods are typically categorized as symmetrical or asymmetrical based on the distribution of weight. In symmetrical loading, weight is evenly distributed across the sagittal or frontal plane; in contrast, asymmetrical loading results in unequal weight distribution across the left/right or anterior/posterior axes of the body. Zhang, reported that under asymmetrical load conditions equivalent to 10% of body weight—such as single-handed carrying, shoulder-loading, or front-holding—older adults exhibited significantly increased cadence and decreased normalized step length (SL) and stride time (ST) [17]. These findings suggest that asymmetrical load carriage adversely affects gait performance in older adults. Gait speed and step length are commonly used indicators of postural stability during linear walking. To improve balance, older adults often adopt a compensatory strategy marked by higher cadence and reduced step length—commonly referred to as a "shuffling gait" [18]. This strategy shortens step length, thereby limiting the displacement of the center of mass, reducing body acceleration, and minimizing ground reaction forces. However, existing evidence suggests that this strategy does not necessarily lower fall risk in older adults. In fact, increased cadence elevates the likelihood of foot-to-foot contact, thereby raising the risk of self-induced tripping [19, 20]. Asymmetrical load carriage introduces torsional forces on the trunk, increasing the demand on the anti-rotational capacity of muscles around the ankle, hip, and other joints in older adults. Additionally, it complicates the biomechanical control required to maintain the center of mass over the base of support [21].

Narouei et al., Bampouras et al., and Badawy et al. (2019) reported no significant effects of symmetrical load carriage on gait parameters [7, 11, 13]. Similarly, Allahverdipour et al. found that symmetrical loading had no significant effect on the time to complete the Timed Up and Go (TUG) test [12]. Previous research

Buketov University

 ISSN 3081-0531 (Print) ISSN 3081-054X (Online)

has demonstrated that bilateral load carriage, compared to unilateral loading, significantly reduces spinal strain, with biomechanical impact estimated to be about half that of single-handed carrying [22]. According to Bampouras et al., symmetrical loading reduced fear of falling in older adults, enhanced perceived walking stability, and may have modulated neural mechanisms involved in postural control [7]. The sensation of "weighted grounding" associated with symmetrical loading was reported to enhance confidence in balance maintenance during quiet standing.

In summary, asymmetrical load carriage impairs gait stability in older adults, as evidenced by increased cadence and decreased step length. Carrying a symmetrical load equivalent to 5% of body weight does not adversely affect gait in older adults and may even enhance postural stability during quiet standing.

Effects of Various Load-Carrying Methods on Joint Kinematics and Muscle Activation During Gait in Older Adults

Matsuo et al. demonstrated that asymmetrical load carriage led to lateral tilting of the trunk and head, unequal shoulder heights, and imbalanced extension torques between the left and right lower limbs [16]. During gait, trunk movement drives upper-limb swinging through shoulder motion, while pelvic rotation initiates lower-limb movement. At this stage, angular momentum is balanced between the upper and lower limbs, contributing to dynamic gait stability [23]. Abnormal trunk and lower-limb movements disrupt angular momentum balance during normal gait. To compensate, other body segments perform compensatory actions, which pose greater challenges for older adults with age-related muscle weakness. In addition to load distribution (symmetrical vs. asymmetrical), some studies also classified load carriage by the load's physical state: stable (e.g., solid objects) versus unstable (e.g., liquids or sand-filled containers). Walsh et al. found that unstable load carriage had a more pronounced impact on trunk stability than stable load conditions [14]. Under unstable load conditions, the Local Divergence Exponent (LDE) in the frontal plane increases, joint angles become irregular, and frontal-plane instability intensifies. Notably, older adults inherently demonstrate lower mechanical stability in the frontal plane compared to the sagittal (anterior–posterior) plane [24–26].

Asymmetrical load carriage leads to imbalanced activation of bilateral muscles, significantly impairing the coordination of homologous muscle groups on both sides of the body. Badawy et al. reported that muscles on the loaded side produced greater force than those on the unloaded side [13]. Among the trunk muscles tested, four showed significantly increased activation under asymmetrical loading: the left external oblique (EO), left lower erector spinae (LES), right latissimus dorsi (LD), and upper erector spinae (UES). The increased activity of these muscles helped maintain an upright trunk posture during gait. Regarding lower-limb muscles, the rectus femoris (RF) showed elevated activation during both posterior load carriage and symmetrical bilateral leg loading. The rectus femoris is a biarticular muscle spanning both the hip and knee joints, primarily responsible for transferring mechanical energy between them (Thiru et al., 1999). This accounts for the increased activation of the rectus femoris under load-carrying conditions. Walsh et al., after evaluating a broader range of lower-limb muscles, reported increased activation of the soleus (SOL) [14]. However, activation levels of the gastrocnemius medialis (GM) and vastus medialis (VM) did not differ significantly between loaded and unloaded conditions. Compared to the GM and VM, the soleus plays a distinct role in gait, contributing more to gravitational support and forward propulsion [27, 28].

In summary, asymmetrical load carriage in older adults induces lateral trunk tilt during walking and causes imbalanced muscle activation between the two sides of the body. Unstable loads increase trunk sway amplitude and compromise postural stability in the frontal plane. Among lower-limb muscles, the rectus femoris (RF) demonstrated the highest level of activation.

Effects of Different Load Weights on Gait Performance in Older Adults

Zhang reported that when older adults carried a unilateral load equivalent to 10% of their body weight, stride time (ST) significantly decreased, while cadence significantly increased compared to unloaded walking [17]. In contrast, Bampouras et al. found that unilateral upper-limb loading using a shopping bag had no significant effect on dynamic gait stability [7]. This suggests that frontal plane perturbations induced by such loading can be compensated for through effective motor control strategies. However, the discrepancy in findings may be attributed to differences in load design: the maximum unilateral load used in the Bampouras et al. study was less than 5% of body weight [7].

Furthermore, Zhang reported that carrying a backpack load equal to 10% of body weight did not significantly alter gait parameters in older adults [17]. In contrast, Walsh et al. found that carrying 15% of body

Buketov University

weight on the back increased step width (SW), potentially compromising postural stability [14]. SW is a critical gait parameter for evaluating fall risk in older adults. When older adults are afraid of falling, they tend to increase step width and reduce stride length to enlarge their base of support and limit displacement of the center of mass, thereby improving stability [4]. Similarly, McAndrew et al. found that variations in SW are strongly correlated with gait stability [29]. A wider step width generally enhances dynamic stability during walking by providing a broader base of support, which aids in maintaining balance and reducing fall risk. Carrying a backpack load equal to 10% of body weight did not negatively impact gait stability in older adults. However, when the load increased to 15% of body weight, older adults reported a perceived decline in gait stability. This perceived instability triggered compensatory mechanisms, such as increased step width, to improve postural control during walking.

A cross-study comparison of the included literature indicates that different load magnitudes have distinct impacts on gait parameters in older adults. Unilateral hand-held loads equivalent to 10% of body weight were found to increase cadence and decrease gait cycle duration. Carrying a backpack load amounting to 15% of body weight was associated with increased step width during walking. However, the exact load threshold at which gait alterations occur is influenced by various factors, including sex, physical fitness, and functional capacity. Further experimental research is required to accurately determine this threshold.

Effects of Varying Load Magnitudes on Joint Kinematics and Muscle Activation During Gait in Older Adults

Matsuo et al. reported that increasing asymmetrical load progressively induced lateral tilt of the head and trunk, exacerbated shoulder height asymmetry, increased torque in the contralateral hip joint, and decreased torque in the ipsilateral hip [16]. As external load increases, additional musculoskeletal components are recruited to maintain frontal plane alignment and trunk equilibrium. Moreover, adequate hip abduction torque is essential for stabilizing the supporting lower limb. The trunk and limbs operate synergistically during gait, and their coordination is crucial for maintaining postural stability. Arm swing and trunk rotation dissipate ground reaction forces and enhance gait stability. The trunk and shoulders function as biomechanical dampers during locomotion, playing a critical role in maintaining postural balance. Trunk rotation and sway help reduce gait oscillations and improve dynamic postural stability [30]. Increased loading typically restricts natural arm swing during walking in older adults. The upper limbs assist mediolateral stability by regulating trunk rotation. In the absence of arm swing, postural stability significantly deteriorates, particularly in older adults [31]. The lower limbs and trunk coordinate synergistically to maintain both vertical and mediolateral stability during gait. During the stance and swing phases, they regulate the body's center of mass (CoM), while trunk motion ensures smooth and stable locomotion [32]. Increased loading amplifies the contribution of the trunk and limbs, whose coordinated movements are essential for maintaining postural stability.

As asymmetrical load increases, the body deviates further from its normal gait pattern, as indicated by trunk lean toward the loaded side, decreased hip torque on the ipsilateral side, and increased torque on the contralateral side.

Conclusions

This study systematically reviewed the biomechanical effects of load carriage on gait in older adults and summarized the experimental methodologies employed to assess gait parameters, joint kinematics, and muscle activity. It synthesized findings based on different load types and magnitudes, highlighting their effects on gait stability, joint kinematics, and muscle coordination during walking in older adults. Current evidence indicates that optical motion capture systems are predominantly used to assess gait and joint kinematics, whereas surface electromyography (sEMG) is commonly applied to measure muscle activation. Load carriage compromises gait stability, alters joint kinematics, and disrupts muscle coordination—effects that are especially pronounced under asymmetrical loading conditions.

Study Limitations

Although this study provides a comprehensive review of the biomechanical effects of load carriage on gait in older adults—specifically in terms of gait parameters, joint kinematics, and muscle activity—several limitations should be acknowledged: (1) The number of included studies was relatively small, which limits the generalizability of the findings and necessitates cautious interpretation. (2) Most included studies focused solely on level-ground walking. The biomechanical implications of load carriage under more challenging

Buketov University

locomotor tasks—such as obstacle crossing or stair negotiation—remain insufficiently studied and warrant further investigation.

Future Research Directions

In view of the biomechanical effects of load carriage on gait in older adults, future research should focus on the following key areas: (1) Investigate the effects of different load magnitudes, while considering individual factors such as sex and physical fitness, to establish safe load thresholds for walking in older adults. (2) Design and assess exercise-based interventions to mitigate the negative impact of load carriage on gait performance and reduce the risk of falls in older adults.

Funding

Authors state no funding involved.

References

1 Glorioso, C. & Sibille, E. (2011). Between destiny and disease: genetics and molecular pathways of human central nervous system aging. *Progress in neurobiology, 93*(2), 165–181. https://doi.org/10.1016/j.pneurobio.2010.11.006

2 World Health Organization (2021). Falls. World Health Organization. World Health Organization. Retrieved from https://www.who.int/news-room/fact-sheets/detail/falls

3 Robinovitch, S. N., Feldman, F., Yang, Y., Schonnop, R., Leung, P. M., Sarraf, T., Sims-Gould, J., & Loughin, M. (2013). Video capture of the circumstances of falls in elderly people residing in long-term care: an observational study. *Lancet, 381*(9860), 47–54. London, England. https://doi.org/10.1016/S0140-6736(12)61263-X

4 Kim, J., Byun, M., & Kim, M. (2020). Physical and Psychological Factors Associated with Poor Self-Reported Health Status in Older Adults with Falls. *International journal of environmental research and public health, 17*(10), 3548. https://doi.org/10.3390/ijerph17103548

5 Sturdy, J. T., Rizeq, H. N., Whittier, T. T., Daquino, C. J., Silder, A., Corman, A. C., Sessoms, P. H., & Silverman, A. K. (2025). Joint moments and muscle excitations increase with body-mass normalized backpacks across walking slopes. *Gait & posture,* 120, 234–241. https://doi.org/10.1016/j.gaitpost.2025.04.007

6 Nagaraja, S., Rubio, J. E., Tong, J., Sundaramurthy, A., Pant, A., Owen, M. K., Samaan, M. A., Noehren, B., & Reifman, J. (2025). Effects of an active ankle exoskeleton on the walking biomechanics of healthy men. *Frontiers in bioengineering and biotechnology,* 13, 1533001. https://doi.org/10.3389/fbioe.2025.1533001

7 Bampouras, T. M. & Dewhurst, S. (2016). Carrying shopping bags does not alter static postural stability and gait parameters in healthy older females. *Gait & posture,* 46, 81–85. https://doi.org/10.1016/j.gaitpost.2016.02.017

8 Moher, D., Liberati, A., Tetzlaff, J., Altman, D. G., & PRISMA Group (2009). Preferred reporting items for systematic reviews and meta-analyses: the PRISMA statement. *PLoS medicine, 6*(7), e1000097. https://doi.org/10.1371/journal.pmed.1000097

9 Risk of Bias Tools (2024). ROBINS-I V2 tool.Risk of Bias Tools Retrieved from https://sites.google.com/site/riskofbiastool/welcome/robins-i-v2

10 Sterne, J. A., Hernán, M. A., Reeves, B. C., Savović, J., Berkman, N. D., Viswanathan, M., Henry, D., Altman, D. G., Ansari, M. T., Boutron, I., Carpenter, J. R., Chan, A. W., Churchill, R., Deeks, J. J., Hróbjartsson, A., Kirkham, J., Jüni, P., Loke, Y. K., Pigott, T. D., Ramsay, C. R., … Higgins, J. P. (2016). ROBINS-I: a tool for assessing risk of bias in non-randomised studies of interventions. *BMJ* (Clinical research ed.), 355, i4919. https://doi.org/10.1136/bmj.i4919

11 Narouei, S., Akatsu, H., & Watanabe, K. (2023). Acute effects of ankle weight loading on regional activity of rectus femoris muscle and lower-extremity kinematics during walking in older adults. *Kinesiology, 55*(1), 70–79. https://hrcak.srce.hr/ojs/index.php/kinesiology/article/view/21056

12 Allahverdipour, H., Dianat, I., Mameh, G., & Asghari Jafarabadi, M. (2021). Effects of Cognitive and Physical Loads on Dynamic and Static Balance Performance of Healthy Older Adults Under Single-, Dual-, and Multi-task Conditions. *Human factors, 63*(7), 1133–1140. https://doi.org/10.1177/0018720820924626

13 Badawy, M., Schall, M. C., Jr, Zabala, M. E., Coker, J., Davis, G. A., Sesek, R. F., & Gallagher, S. (2019). Trunk muscle activity among older and obese individuals during one-handed carrying. *Applied ergonomics,* 78, 217–223. https://doi.org/10.1016/j.apergo.2019.03.007

14 Walsh, G. S., Low, D. C., & Arkesteijn, M. (2018). Effect of stable and unstable load carriage on walking gait variability, dynamic stability and muscle activity of older adults. *Journal of Biomechanics,* 75, 75–81. https://doi.org/10.1016/j.jbiomech.2018.04.018

15 Kong, P. W. & Chua, Y. K. (2014). Start-up time and walking speed in older adults under loaded conditions during simulated road crossing. *Experimental aging research, 40*(5), 589–598. https://doi.org/10.1080/0361073X.2014.956630

16 Matsuo, T., Hashimoto, M., Koyanagi, M., & Hashizume, K. (2008). Asymmetric load-carrying in young and elderly women: relationship with lower limb coordination. *Gait & posture, 28*(3), 517–520. https://doi.org/10.1016/j.gaitpost.2008.02.001

Buketov University

17 Zhang, T., Zhang, J., Ji, R., et al. (2018). A comparative analysis of the effects of different load-bearing methods on walking in older adults. *Chinese Journal of Sports Medicine, 37*(12), 1005–1010. https://doi.org/10.16038/j.1000-6710.2018.12.006

18 Fan, Y., Li, Z., Han, S., Lv, C., & Zhang, B. (2016). The influence of gait speed on the stability of walking among the elderly. *Gait & posture,* 47, 31–36. https://doi.org/10.1016/j.gaitpost.2016.02.018

19 Barak, Y., Wagenaar, R. C., & Holt, K. G. (2006). Gait characteristics of elderly people with a history of falls: a dynamic approach. *Physical therapy, 86*(11), 1501–1510. https://doi.org/10.2522/ptj.20050387

20 Pavol, M. J., Owings, T. M., Foley, K. T., & Grabiner, M. D. (1999). Gait characteristics as risk factors for falling from trips induced in older adults. *The journals of gerontology. Series A, Biological sciences and medical sciences, 54*(11), M583–M590. https://doi.org/10.1093/gerona/54.11.m583

21 Pijnappels, M., Reeves, N. D., Maganaris, C. N., & van Dieën, J. H. (2008). Tripping without falling; lower limb strength, a limitation for balance recovery and a target for training in the elderly. *Journal of electromyography and kinesiology: official journal of the International Society of Electrophysiological Kinesiology, 18*(2), 188–196. https://doi.org/10.1016/j.jelekin.2007.06.004

22 McGill, S. M., Marshall, L., & Andersen, J. (2013). Low back loads while walking and carrying: comparing the load carried in one hand or in both hands. *Ergonomics, 56*(2), 293–302. https://doi.org/10.1080/00140139.2012.752528

23 Vera-Garcia, F. J., Moreside, J. M., & McGill, S. M. (2011). Abdominal muscle activation changes if the purpose is to control pelvis motion or thorax motion. *Journal of electromyography and kinesiology: official journal of the International Society of Electrophysiological Kinesiology, 21*(6), 893–903. https://doi.org/10.1016/j.jelekin.2011.08.003

24 Bauby, C. E., & Kuo, A. D. (2000). Active control of lateral balance in human walking. *Journal of Biomechanics, 33*(11), 1433–1440. https://doi.org/10.1016/S0021-9290(00)00101-9

25 Rankin, B. L., Buffo, S. K., & Dean, J. C. (2014). A neuromechanical strategy for mediolateral foot placement in walking humans. *Journal of neurophysiology, 112*(2), 374–383. https://doi.org/10.1152/jn.00138.2014

26 Schrager, M. A., Kelly, V. E., Price, R., Ferrucci, L., & Shumway-Cook, A. (2008). The effects of age on medio-lateral stability during normal and narrow base walking. *Gait & posture, 28*(3), 466–471. https://doi.org/10.1016/j.gaitpost.2008.02.009

27 Annaswamy, T. M., Giddings, C. J., Della Croce, U., & Kerrigan, D. C. (1999). Rectus femoris: its role in normal gait. *Archives of physical medicine and rehabilitation, 80*(8), 930–934. https://doi.org/10.1016/s0003-9993(99)90085-0

28 Cronin, N. J., Avela, J., Finni, T., & Peltonen, J. (2013). Differences in contractile behaviour between the soleus and medial gastrocnemius muscles during human walking. *The Journal of experimental biology, 216*(Pt 5), 909–914. https://doi.org/10.1242/jeb.078196

29 McAndrew Young, P. M., Wilken, J. M., & Dingwell, J. B. (2012). Dynamic margins of stability during human walking in destabilizing environments. *Journal of biomechanics, 45*(6), 1053–1059. https://doi.org/10.1016/j.jbiomech.2011.12.027

30 Pontzer, H., Holloway, J. H., 4th, Raichlen, D. A., & Lieberman, D. E. (2009). Control and function of arm swing in human walking and running. *The Journal of experimental biology, 212*(Pt 4), 523–534. https://doi.org/10.1242/jeb.024927

31 Ortega, J. D., Fehlman, L. A., & Farley, C. T. (2008). Effects of aging and arm swing on the metabolic cost of stability in human walking. *Journal of biomechanics, 41*(16), 3303–3308. https://doi.org/10.1016/j.jbiomech.2008.06.039

32 Meyns, P., Bruijn, S. M., & Duysens, J. (2013). The how and why of arm swing during human walking. *Gait & Posture, 38*(4), 555–562. https://doi.org/10.1016/j.gaitpost.2013.02.006

Information about the authors

Zelong, Gai — MSc (Human Movement Science), School of Sport and Health Sciences, Dalian University of Technology, Dalian, China; e-mail: gaizelong@mail.dlut.edu.cn, ORCID ID: 0009-0001-8297-476X

Haibin, Liu — PhD (Biomedical Engineering), Associate Professor, Vice Dean, School of Sport and Health Sciences, Dalian University of Technology, Dalian, China; e-mail: liuhaibin@dlut.edu.cn, ORCID ID: 0000-0001-9197-4959

Zhaoli, Meng (contact person) — MSc (Human Movement Science), Professor, Supervisor of Master's Students; School of Sport and Health Sciences, Dalian University of Technology, Dalian, China; e-mail: mengzl@dlut.edu.cn, ORCID ID: 0000-0002-9338-305X

Buketov University