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Research Article

Enhancing Above-Knee Prosthetic Design for Inclusive Workplaces: Ergonomic Considerations in Manual Material Handling

Khoirul Muslim [1, *], Adinda Shafira [1], Rifko Rahmat Kurnianto [1], Sandro Mihradi [2], Auditya Purwandini Sutarto [3]

[1] Department of Industrial Engineering and Management, Institut Teknologi Bandung, Bandung, Indonesia
[2] Faculty of Mechanical and Aerospace Engineering, Institut Teknologi Bandung, Bandung, Indonesia
[3] Department of Industrial Engineering, Universitas Qomaruddin, Gresik, Indonesia

*Corresponding Author: kmuslim@itb.ac.id

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ABSTRACT

Employment is crucial for economic sustainability and social inclusion, yet individuals with disabilities face significant barriers. Globally, only 44% of disabled individuals are employed compared to 75% of those without disabilities. Manual material handling (MMH) relies heavily on stability and control in demanding industries such as manufacturing and logistics. Such demands create challenges for individuals with above-knee prostheses, as most current designs focus on walking and do not adequately support the postural and load-bearing requirements of MMH tasks. This study aims to evaluate the performance of transfemoral prosthesis designs during MMH, analyzing the effects of container type, load mass, and their interaction on gait efficiency, discomfort, and stability. Eight male unilateral above-knee amputees (24–39 y) carried handled and handle-less boxes loaded from 4 to 10 kg in a randomised within-subject trial. Gait deviation, perceived discomfort, and steadiness were captured with self-report measures. Two-way analysis of variance analyses showed a significant container × load interaction: handle-less 10 kg loads produced the greatest lateral trunk lean toward the prosthetic side, whereas lighter handled loads minimised deviation. Increasing load also elevated discomfort in the back, waist, stump and contralateral arm and reduced perceived stability. Observed lateral lean and impact-related knee extension suggest three priority modifications: (1) add socket adduction within an ischial-containment design to improve femoral stabilisation, (2) increase knee-swing friction to soften terminal impact, and (3) fit dual-keel feet to cushion heel strike. Implementing these changes may reduce gait errors and fatigue, raising safe lifting capacity for transfemoral prosthesis users in MMH task. Nonetheless, the male-only sample may not capture gender-specific gait strategies; future trials should include female participants and a larger cohort to verify generalisability. These preliminary findings still offer insights into improving prosthetic designs to enhance safety, functionality, and inclusion in industrial MMH tasks.

Keywords: disability, ergonomics, material handling, prosthetic, stability

INTRODUCTION

Lower-limb disabilities encompass a wide range of conditions, including lower extremity musculoskeletal disorders (MSDs) and amputations. Globally, an estimated 65 million people live with limb amputations, and 1.5 million amputations are performed annually, with 60% involving lower limbs [1]. These disabilities include above-knee and transtibial conditions, significantly impacting mobility and daily functioning [2], [3]. Beyond physical limitations, individuals with lower-limb disabilities often experience biomechanical challenges that affect their stability, balance, and ability to perform physically demanding tasks [4]. These functional limitations not only restrict movement but also pose significant challenges in securing and maintaining employment, particularly in industries requiring manual labor, prolonged standing, or frequent mobility [5].

For people with disabilities, employment offers economic independence and social inclusion, yet they face significant barriers to entering and remaining in the workforce [6], [7]. Globally, only 44% of individuals with disabilities are employed compared to 75% without disabilities [8]. with this disparity being more pronounced in developing countries [9], [10]. In Indonesia, only 0.53% of disabled individuals are gainfully employed [11]. National Socioeconomic Survey data indicates that 63.28% of individuals with disabilities in the productive age group experience at least one type of limitation [11], with 7.44% reporting difficulty walking or climbing stairs, often linked to above-knee lower-limb disabilities. WHO data highlights that systemic barriers, including the absence of tailored ergonomic interventions and inadequate assistive technology in physically demanding sectors like manufacturing, logistics, and agriculture [4], and social stigma, contribute to lower labor market participation among individuals with disabilities [12]. These challenges are exacerbated by the mismatches between task assignments and individuals' actual capabilities, often based on inaccurate assumptions about what tasks they can perform [4].

These challenges highlight the need for innovative solutions, including practical tools and technologies, to empower individuals with disabilities in diverse work environments. Improving access to physically demanding jobs for prosthesis users aligns with broader goals such as SDG 8, which promotes inclusive employment and decent work opportunities [13]. Among the priority areas, enabling participation in physically demanding tasks, particularly manual material handling (MMH), can play a pivotal role in addressing these gaps, fostering greater autonomy and workforce integration. MMH tasks, which involve lifting, carrying, pushing, and pulling loads, are common in industrial sectors and require stability, control, and force distribution to prevent injuries [14]. Although clinicians have developed prosthetic designs to enhance walking mobility, they often fall short in meeting the postural and load-bearing demands of MMH [15], [16], [17], making it difficult for above-knee users to perform such tasks effectively [4].

Therefore, proper construction, alignment, and weight distribution are crucial to restoring are crucial to restoring mobility, mainly when performing dynamic load-bearing tasks like MMH. Without these considerations, users face fatigue, discomfort, and instability, increasing their risk of injury and limiting occupational performance [17–19]. To enhance prosthetic performance in real-world MMH conditions, factors such as load mass, movement strategy, and container design should be systematically addressed. Prior studies have explored gait mechanics and general prosthetic performance, particularly in walking context [15], [17], [18], [19], [20], [21]. However, research on the challenges of above-knee prosthetic users in dynamic MMH tasks remains understudied.

Although load mass significantly impacts walking efficiency, stability, and comfort [17], [18], [20], existing work has not assessed how external loads affect gait mechanics and postural stability during MMH. For instance, Kahle et al. [18] identified key predictors of walking ability, such as amputation level and physical fitness, but did not evaluate lifting or carrying tasks. Similarly, Pienaar [17] discussed mobility limitations and reduced satisfaction among transfemoral prosthesis users without addressing occupational load-handling demands. Köhler et al. [15] further examined socket adduction and trunk stabilization but focused on static walking, overlooking dynamic load-bearing and upper-body compensatory strategies. While Anderst et al. [21] explored socket design for comfort, their study also excluded scenarios involving external loads.

Beyond prosthetic design, MMH performance may also be affected by the form of the load and the quality of coupling between the load and the user. Simple design features, such as handles, can significantly enhance this coupling and improve lifting and carrying efficiency [22], [23]. Selecting appropriate container types in conjunction

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with well-fitted prosthetic components may determine how much load users can safely manage [24], [25]. Thus, understanding the interplay between load mass, form, and coupling is important for optimizing transfemoral prosthetic performance in MMH tasks.

Despite the growing need for prosthetic solutions in physically demanding occupations, limited research has examined how real-world load characteristics affect gait, comfort, and stability among above-knee prosthesis users during MMH. This gap presents a critical challenge in aligning prosthetic design with the functional demands of industrial tasks. Therefore, this study aims to evaluate the effects of container type and load mass on the gait efficiency, perceived discomfort, and stability of above-knee prosthesis users during manual material handling tasks. To achieve this aim, the study addresses the following research questions; 1). How do variations in container type and load mass influence gait efficiency, perceived discomfort, and steadiness in individuals using transfemoral prostheses during MMH tasks? 2). What design implications can be drawn from these findings to optimize prosthetic components—such as the socket, knee joint, and foot—for improved performance in load-bearing environments?

Prior Work on Prosthetic Design

Recognizing these challenges, the biomechanical research team at the Faculty of Mechanical and Aerospace Engineering, Bandung Institute of Technology (FTMD ITB), has been working since 2014 to address the unique needs of individuals with above-knee disabilities. Their 4-bar linkage D2 knee joint prosthetic shows potential for improving prosthesis performance in work-related tasks, which may contribute to more inclusive employment environments. However, to further refine this technology, it is crucial to understand its performance under real-world MMH conditions, considering variations in load mass and container types.

As illustrated in Figure 1, the transfemoral prosthesis evaluated in this study comprises seven components. At the top, the belt provides additional support by securing the residual limb's prosthesis, helping maintain proper positioning during movement. The socket acts as the crucial interface, ensuring a snug and comfortable fit against the residual limb, which is vital for mobility and long-term wear. The socket adaptor connects the socket to the knee joint, allowing component alignment adjustments for a customized fit. The knee joint, featuring a 4-bar linkage D2 mechanism, is critical for stability and a more natural gait, especially during weight-bearing activities like standing, walking, and MMH. Unlike single-axis prosthetic knees, the D2 linkage allows a moving center of rotation that improves foot clearance during swing and enhances stance-phase stability—both crucial in tasks involving lifting and asymmetrical loading. Prior research confirms that such mechanisms reduce trunk compensation and better distribute loading during dynamic tasks [15], [26]. The shank functions as the artificial lower leg, transferring forces between the knee joint and the foot to ensure smooth and efficient movement. The foot adaptor connects the shank to the foot, enhancing mechanical stability. Finally, the foot simulates natural foot motion, aiding balance and shock absorption.

A reliable prosthesis minimizes gait deviation and supports comfort and stability—key outcomes assessed in this study through both objective and subjective measures [20], [26]. Specifically, we evaluated how container type and load mass influence prosthetic performance. Findings will inform targeted design improvements in socket alignment, knee mechanics, and weight distribution to enhance functionality under MMH conditions. By adopting a human-centered (soft systems) approach, this research intends to develop adaptable prosthetic solutions that meet the specific needs of this population in real-world job environments. The study's outcomes are expected to contribute to inclusive workplace policies, enhance workforce participation for individuals with disabilities, and advance industrial ergonomics. Although this study was conducted in Indonesia, the core challenges, including limited assistive technology, biomechanical constraints in manual labor, and workplace exclusion, are common across

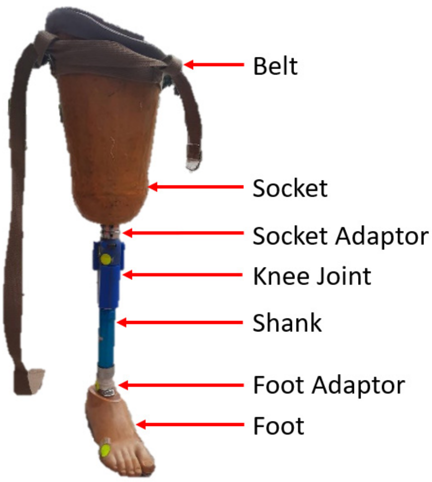

Figure 1. Components of Transfemoral Prosthesis for Above-Knee Disabilities

diverse contexts [27], [28], [29]. Accordingly, the findings offer valuable insights for global efforts to improve prosthetic functionality and integration in sectors where manual handling is essential but often excluded from current prosthetic design domains.

METHODS

Study Design

This experimental study utilized a within-subject design, exposing each participant to all treatment conditions. The treatments consisted of combinations of two independent variables: container type and load mass. Each participant completed all treatment conditions, which consisted of eight combinations derived from two independent variables: container type (with or without handles) and load mass (4 kg, 6 kg, 8 kg, and 10 kg). The selected load masses were chosen to simulate weights commonly encountered during daily or occupational tasks [30]. These weight levels were also appropriate for individuals using transfemoral prostheses and aligned with recommendations from [31] who advised that carried loads should not exceed 20% of a participant's body weight. To avoid asymmetry-related variability, the loads were standardized using uniform book-shaped objects.

To control for order effects, the treatment sequence was randomized using the Latin Square Design method [32], as shown in Table 1. Descriptions of treatment codes (A–H) are provided in Table 2.

Table 1. Order of Experimental Treatments Based on the Latin Square Design Method

Participant							
Treatment Code	**2**	**3**	**4**	**5**	**6**	**7**	**8**
A	B	C	D	E	F	G	H
B	C	D	E	F	G	H	A
C	D	E	F	G	H	A	B
D	E	F	G	H	A	B	C
E	F	G	H	A	B	C	D
F	G	H	A	B	C	D	E
G	H	A	B	C	D	E	F
H	A	B	C	D	E	F	G

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Table 2. Description of Treatment Codes

Code	Load (kg)	Container Type
A	4	Box without handle
B	4	Box with handle
C	6	Box without handle
D	6	Box with handle
E	8	Box without handle
F	8	Box with handle
G	10	Box without handle
H	10	Box with handle

Note: Each treatment involved walking 5 meters while carrying the specified load and container.

Participants

The study recruited eight male participants with unilateral above-knee disabilities, representing the productive working-age population. Participants had diverse amputation histories, including cases resulting from accidents, infections, and congenital conditions, with durations ranging from 2 to 30 years. All participants had been using a prosthetic limb for at least one year (range: 2–17 years) and were able to walk independently with their prosthesis. Eligibility criteria were adapted from [31], [33]. Inclusion criteria required participants to: (1) have an above-knee amputation, (2) be capable of independently using a prosthetic limb, (3) have a minimum of one year of prosthesis experience, and (4) fall within the working-age range of 15–64 years. Exclusion criteria included the presence of comorbidities or medical conditions that could affect gait, balance, or general mobility. The participants' ages ranged from 24 to 39 years, with a mean (standard deviation) of 31.63 (5.85) years. Their average height was 163.43 (3.87) cm, stump height was 35.57 (8.79) cm, pelvic height was 89.63 (2.26) cm, body weight without the prosthetic leg was 58.76 (10.00) kg, and prosthetic leg weight was 3.04 (0.16) kg. All participants provided informed consent following ethical procedures approved by the Research Committee at ITB.

Instruments

This experimental study employed three key instruments: The Prosthesis Evaluation Questionnaire or Prosthetic Observational Gait Scale (POGS), the Rate of Perceived Discomfort (RPD) scale, and the Perception of Steadiness scale. POGS, specifically designed for individuals with lower body disabilities, comprises 16 criteria related to gait deviation such as arm swing, trunk lean, hip/knee motion, and step symmetry) [33]. It provides a standardized approach for assessing gait efficiency and is recognized for high interobserver repeatability. The POGS is simple to implement in clinical settings, requiring only basic tools like a goniometer, without the need for advanced technology [34]. Incorporating subjective feedback alongside POGS further ensures a comprehensive evaluation [2], [19].

The RPD scale evaluates discomfort using the modified Borg CR10 scale, which captures relative preferences related to discomfort levels [35]. Additionally, the Perception of Steadiness scale measures subjective stability on a 0–10 scale, providing insights into participants' perceived steadiness [36].

Procedure

Each participant completed one experimental session consisting of four phases: preparation, practice, calibration, and data collection. During the preparation phase, 26 reflective markers were attached to the upper and lower body (Figure 2). Participants then performed practice trials to familiarize themselves with the load-carrying tasks (Figure 3).

Figure 2. Markers Attached to a Participant

Figure 3. A Participant Carries Load of 4 kg, 6 kg, 8 kg, and 10 kg Using a Box with and without Handle

Calibration involved establishing a reference range for two subjective measures. For the Rate of Perceived Discomfort (RPD) scale, participants stood upright with eyes open and extended one arm forward to a 90° angle from the body. A 4 kg load was placed on the palm, and participants were instructed to hold the load until they could no longer maintain the position, following the protocol by [37]. A score of 0 indicated no discomfort or pain, while 10 represented extreme discomfort. For the steadiness scale, participants experienced two controlled reference conditions. For the most stable condition (score = 10), they stood with feet shoulder-width apart, eyes open, and held onto a vertical support pole. For the least stable condition (score = 0), they closed their eyes and stood unsupported on the prosthetic leg only until balance could no longer be maintained.

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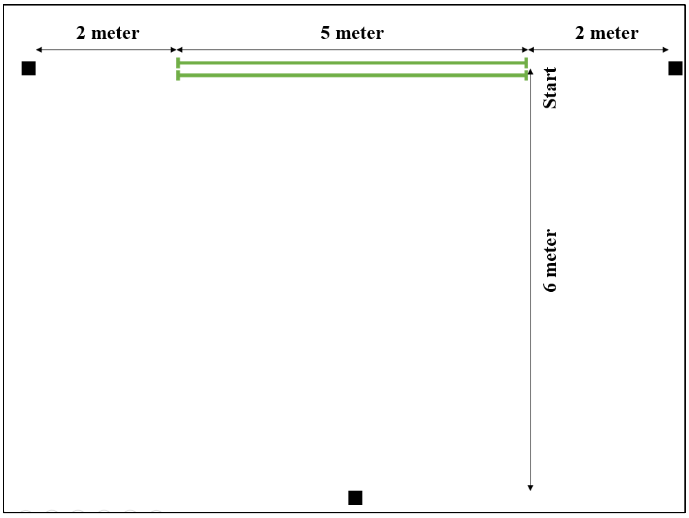

Figure 4. Placement of the Camera; the black box represents the camera

In the data collection phase, participants walked at a self-selected speed along a 5-meter walkway [33], while carrying different load and container combinations. Each treatment was repeated three times, resulting in a total of 24 trials per participant. A helper placed and retrieved the load at each end of the walkway. After each trial, participants rated perceived discomfort and stability. Participants were given a minimum rest period of 60 seconds between trials, with the flexibility to rest longer if needed. This rest protocol was based on pilot testing and is consistent with prior studies involving gait and manual material handling in lower-limb prosthesis users (e.g., [27], 38]). Before each subsequent trial, participants were verbally asked to confirm that they felt adequately rested to continue, thereby reducing potential fatigue-related bias. Three GoPro cameras positioned at the front, side, and rear of the walkway (Figure 4) recorded gait performance.

Data Analysis

The ordinal POGS data were transformed using the Aligned Rank Transform (ART) method in ARTool software to facilitate analysis with ANOVA in IBM SPSS Statistics 24 [39]. Interaction effects and individual factor significance were tested with a threshold of $P < 0.05$. Post-hoc analyses utilized Paired Sample t-tests when appropriate. Subjective data on discomfort and stability were analyzed using two-way repeated-measures ANOVA, with the significance level set at $\alpha < 0.05$. Post-hoc pairwise comparisons were performed using the Least Significant Difference (LSD) test where necessary. Trend analysis was also incorporated to enhance the interpretation of prosthesis performance during manual material handling tasks.

RESULT AND DISCUSSION

This section addresses the impact of load weight, container type, and their interaction on the performance of transfemoral prostheses during MMH tasks. It integrates findings from objective gait assessments and subjective evaluations to identify key challenges and propose targeted design improvements for enhancing prosthetic functionality and user experience.

Effect of Load Mass and Container Type on POGS

As shown in Table 3, the POGS revealed a significant interaction effect of container type and load mass on lateral trunk lean/side flexion in stance ($F(3,56) = 3.389$, $p < 0.05$, $\eta_p^2 = 0.154$). The moderate effect size indicates that the

Table 3. Results of ART-ANOVA test for POGS scores

Criteria	**Effect**					
	Type of Container		**Load**		**Type of Container × load**	
	$F_{(1,56)}$	η_p^2	$F_{(3,56)}$	η_p^2	$F_{(3,56)}$	η_p^2
1. Arm swing	-		-		-	
2. Vaulting in stance	0.145	0.003	0.279	0.015	0.031	0.002
3. Lateral trunk lean/side flexion in stance	4.677	0.077	1.700	0.083	3.389*	0.154
4. Peak sagittal position	1.112	0.019	0.597	0.031	1.330	0.067
5. Peak hip extension in stance	0.195	0.003	2.555	0.120	0.720	0.037
6. Peak hip flexion in swing	0.023	0.000	0.564	0.029	0.032	0.002
7. Peak knee extension in stance	1.621	0.028	0.178	0.009	0.851	0.044
8. Knee flexion in terminal stance	0.196	0.003	0.455	0.024	0.493	0.026
9. Peak knee flexion/heel rise in swing	0.360	0.006	0.158	0.008	0.188	0.010
10. Knee in terminal swing and at initial contact	0.346	0.006	0.040	0.002	0.078	0.004
11. Step symmetry	0.000	0.000	0.714	0.037	0.564	0.029
12. 1st ankle rocker	0.026	0.000	0.009	0.000	0.009	0.000
13. Foot rotation at initial contact	1.798	0.031	0.886	0.045	1.857	0.090
14. Width of base/Lateral thrust	3.054	0.052	1.231	0.062	0.494	0.026
15. Circumduction in swing	0.020	0.018	0.238	0.013	0.282	0.015
16. Swing phase whip	0.500	0.000	0.214	0.011	0.428	0.022

Note. *Significant at $p<0.05$

combined influence of load mass and container design had a meaningful and consistent impact on trunk stability during MMH tasks, particularly on the prosthetic side. However, no significant main effects of load or container type were found for each POGS criterion. One plausible explanation is that the differences between containers (with vs. without handles) were not functionally distinct enough to alter gait mechanics or weight distribution, especially over short-duration tasks. Participants may have compensated through habitual movement strategies that maintained overall performance, masking subtle variations. Moreover, the POGS instrument, which focuses on observable gait deviations, may lack sensitivity to detect finer biomechanical changes related to upper-body loading. The limited sample size may have further reduced the ability to detect small interaction effects.

Further post hoc tests indicated that this effect was driven by the weight of the load, particularly with the 10 kg load for the box without a handle, which exhibited a higher average value compared to loads of 4 kg, 6 kg, and 8 kg (see Figure 5). Conversely, 4 kg, 6 kg, and 8 kg for the box with a handle had a higher average value than the 10 kg load.

Above-knee prosthetic users tend to lean their bodies toward the prosthetic side, especially when carrying loads. Specifically, at 6 kg and 8 kg, participants exhibited a higher average POGS value when carrying a handled box than a box without a handle. Conversely, when carrying a 10 kg load, the average POGS value was higher for the box without a handle than for the handled box. Loads under 8 kg are better carried using a handled box to reduce gait errors, while a 10 kg load is better carried using a box without a handle.

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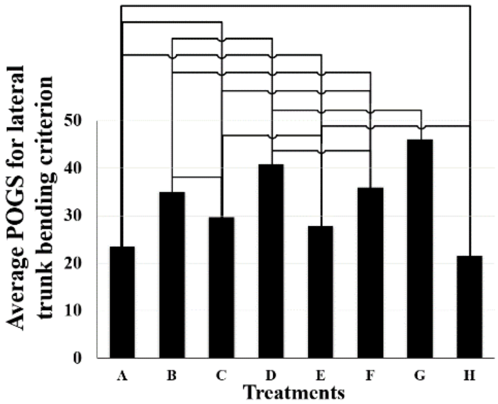

Figure 5. Average POGS scores for the lateral trunk bending criterion across treatments. Treatments (A–H) involve carrying tasks along a 5-meter walkway with varying load weights (4 kg, 6 kg, 8 kg, 10 kg) and container types (box with or without a handle). Error bars represent standard deviations.

Lateral trunk bending is a well-documented compensatory mechanism in above-knee prosthesis users, particularly among individuals with short residual limbs or socket fit challenges 40], [41]. This strategy is often a response to issues such as inadequate femoral stabilization, weak hip abductors, and discomfort caused by misaligned sockets or excessive abduction angles [33]. Improper alignment, especially at the socket or foot, can shift the center of gravity outward, increasing lateral lean as a means to maintain balance [15]. In addition, the mechanical demands of heavier loads can intensify discomfort and fatigue, making such compensatory postures more pronounced [2], [19].

Pain or discomfort, especially at the lateral distal aspect of the residual limb, and an abducted gait further contribute to this issue [33]. [19] increased load mass places additional mechanical strain on the residual limb and prosthetic components, heightening fatigue and instability. This result aligns with the present study's findings that heavier loads and specific container designs interact to influence lateral trunk bending, with load mass being the primary driver of gait deviations.

Furthermore, as shown in Table 4, trend data for each POGS criterion across treatments is summarized using the mode and its corresponding frequency (i.e., the percentage of participants who received the modal score). A mode value of 2 indicates the highest level of gait error for each criterion. In addition to criterion 1 (arm swing), two other criteria consistently show the maximum mode value: criterion 10 (knee position at the end of the swing phase and initial ground contact) and criterion 12 (heel initial contact). We focus on these two criteria due to their critical role in gait stability and comfort.

For the knee criterion, terminal swing impact occurs due to the high speed of the stump when the knee joint reaches maximum extension, causing discomfort and an audible impact sound. Kohler et al. [42] highlights that improper knee joint alignment or settings can exacerbate this issue, resulting in asymmetrical steps and discomfort. Mohamed and Appling [19] emphasizes the importance of gradual speed reduction during terminal swing to ensure smoother transitions and reduce strain on the stump. This issue also arises from suboptimal knee joint settings, the need for adjustments, or the user's habit of ensuring maximum extension by relying on the sound of the impact [33], 40]. A well-designed prosthesis should gradually reduce speed to produce symmetrical steps without impact.

For criterion 12 (heel initial contact), proper gait mechanics require the heel to touch the ground first, then gradually lowering the foot to maintain stability. Deviations can occur if the motion is too fast (foot slap) or too slow (excessive heel compression), leading to excessive pressure on the heel and compromised stability [15]. Contributing factors include incorrect anteroposterior positioning of the prosthetic foot relative to the socket, unsuitable foot selection, insufficient socket flexibility, or excessive reliance on knee extensors [40].

Table 4. Mode and Frequency of POGS Scores Across Treatments (A–H)

Criteria	**Treatment**															
	A		**B**		**C**		D		E		F		G		H	
	Mode	**Freq** (%)	**Mode**	**Freq** (%)	**Mode**	**Freq** (%)	**Mode**	**Freq** (%)	**Mode**	**Freq** (%)	**Mode**	**Freq** (%)	**Mode**	**Freq** (%)	**Mode**	**Freq** (%)
1. Arm swing	2	100	2	100	2	100	2	100	2	100	2	100	2	100	2	100
2. Vaulting in stance	0	75	0	62.5	0	50	0	62.5	0	50	0	62.5	0	62.5	0	75
3. Lateral trunk lean/side flexion instance	1	62.5	1 & 2	50	1	62.5	1	50	1	62.5	1	50	1	50	2	50
4. Peak sagittal position	1	87.5	1	62.5	1	75	1	62.5	1	50	1	87.5	1	50	1	87.5
5. Peak hip extension in stance	0	75	0	62.5	0	75	0	62.5	0	100	0	100	0	87.5	0	87.5
6. Peak hip flexion in swing	1	62.5	1 & 2	50	1	75	1	75	1	62.5	1 & 2	50	1	62.5	1	62.5
7. Peak knee extension in stance	0	62.5	0	62.5	0	75	0	62.5	0	87.5	1	62.5	0	75	0	50
8. Knee flexion in terminal stance	1 & 2	50	1	62.5	2	62.5	2	62.5	2	75	1 & 2	50	1 & 2	50	2	62.5
9. Peak knee flexion/heel rise in swing	0 & 1	37.5	1	50	0 & 2	37.5	0 & 1	62.5	0 & 1	37.5	1	50	1	62.5	0	37.5
10. Knee in terminal swing and at initial contact	2	62.5	2	62.5	2	62.5	2	62.5	2	62.5	2	50	2	62.5	2	50
11. Step symmetry	2	75	1 & 2	50	1	75	2	50	2	62.5	2	62.5	1 & 2	50	2	50
12. 1st ankle rocker	2	87.5	2	100	2	87.5	2	100	2	87.5	2	87.5	2	87.5	2	87.5
13. Foot rotation at initial contact	2	62.5	2	62.5	0 & 2	37.5	2	50	2	62.5	1	62.5	1 & 2	50	2	62.5
14. Width of base/Lateral thrust	1	75	1	87.5	0 & 1	50	1	75	0	62.5	1	75	0 & 1	50	1	75
15.Circumduction in swing	1	62.5	1	75	1	87.5	1	75	1	50	1	87.5	1	62.5	1	75
16. Swing phase whip	1	75	1 & 2	37.5	1	50	1	75	1	62.5	1	62.5	1	75	1	62.5

Note. Treatments (A–H) involve carrying tasks along a 5-meter walkway with varying load weights (4 kg, 6 kg, 8 kg, 10 kg) and container types (box with or without a handle). Mode: Most frequent POGS score. Freq: Percentage of participants with the modal score. POGS score levels (0, 1, 2) are based on the assessment criteria described in the POGS evaluation sheet by [33]

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Criterion 11 (step symmetry) also revealed consistent mode scores of 2 across most treatment conditions, suggesting persistent bilateral gait asymmetry. This finding aligns with prior reports indicating that step asymmetry reflects poor balance and compensatory movement in lower-limb prosthesis users [43]. Additionally, frequent deviations in Criterion 13 (foot rotation at initial contact) may indicate rotational misalignment or instability during heel strike [44]. Deviations in Criterion 8 (knee flexion in terminal stance) could also point to propulsion inefficiencies and suboptimal knee damping during push-off [16]. Collectively, these variables further support the need for holistic evaluation in prosthetic gait optimization.

Effect of Load Mass and Container on RPD

RPD scores revealed significant effects of load mass on several body regions, including the prosthesis side of the back, arm, hand, waist, pelvis, and stump, as well as the normal side of the arm and waist (see Table 5). The effect sizes associated with these findings ranged from partial eta squared values of 0.391 to 0.513, which indicate moderate to large effects. These results suggest that increasing load weight consistently and meaningfully elevates discomfort in both the prosthetic and non-prosthetic sides of the body.

Subsequent post hoc tests clarified that as the load weight increased, participants experienced higher levels of discomfort when utilizing the transfemoral prosthesis for MMH tasks. As illustrated in Figure 6, the prosthesis side

Table 5. Results of Repeated Measures ANOVA test for RPD

Criteria	Effect					
	Type of Container		Load		Type of Container × load	
	$F_{(1,7)}$	η_p^2	$F_{(3,21)}$	η_p^2	$F_{(3,21)}$	η_p^2
Neck	2.215	0.240	1.083	0.134	1.317	0.158
Prosthesis side of shoulder	0.118	0.017	2.779	0.284	0.846	0.108
Normal side of shoulder	0.263	0.036	2.697	0.278	1.068	0.132
Prosthesis side of back	1.039	0.129	4.489*	0.391	1.632	0.189
Normal side of back	0.689	0.090	3.978	0.362	1.000	0.125
Prosthesis side of arm	4.000	0.364	4.667*	0.400	2.100	0.231
Normal side of arm	2.333	0.250	4.972**	0.415	0.950	0.119
Prosthesis side of hand	0.051	0.007	5.175**	0.425	0.037	0.005
Normal side of hand	0.003	0.000	2.933	0.295	0.869	0.110
Prosthesis side of waist	0.590	0.078	5.523**	0.441	1.155	0.142
Normal side of waist	0.646	0.084	4.578*	0.395	0.542	0.072
Prosthesis side of pelvic	0.059	0.008	4.584*	0.396	0.902	0.114
Normal side of pelvic	1.217	0.148	2.745	0.282	0.000	0.000
Normal side of thigh	0.378	0.051	3.501	0.333	0.463	0.062
Prosthesis side of stump	0.010	0.001	7.373**	0.513	1.257	0.154
Knee	0.052	0.007	2.110	0.232	1.278	0.154
Ankle	0.197	0.027	1.640	0.190	0.167	0.023
Foot	0.717	0.093	2.706	0.279	0.956	0.120

Note. *Significant at p<0.05, **<0.01

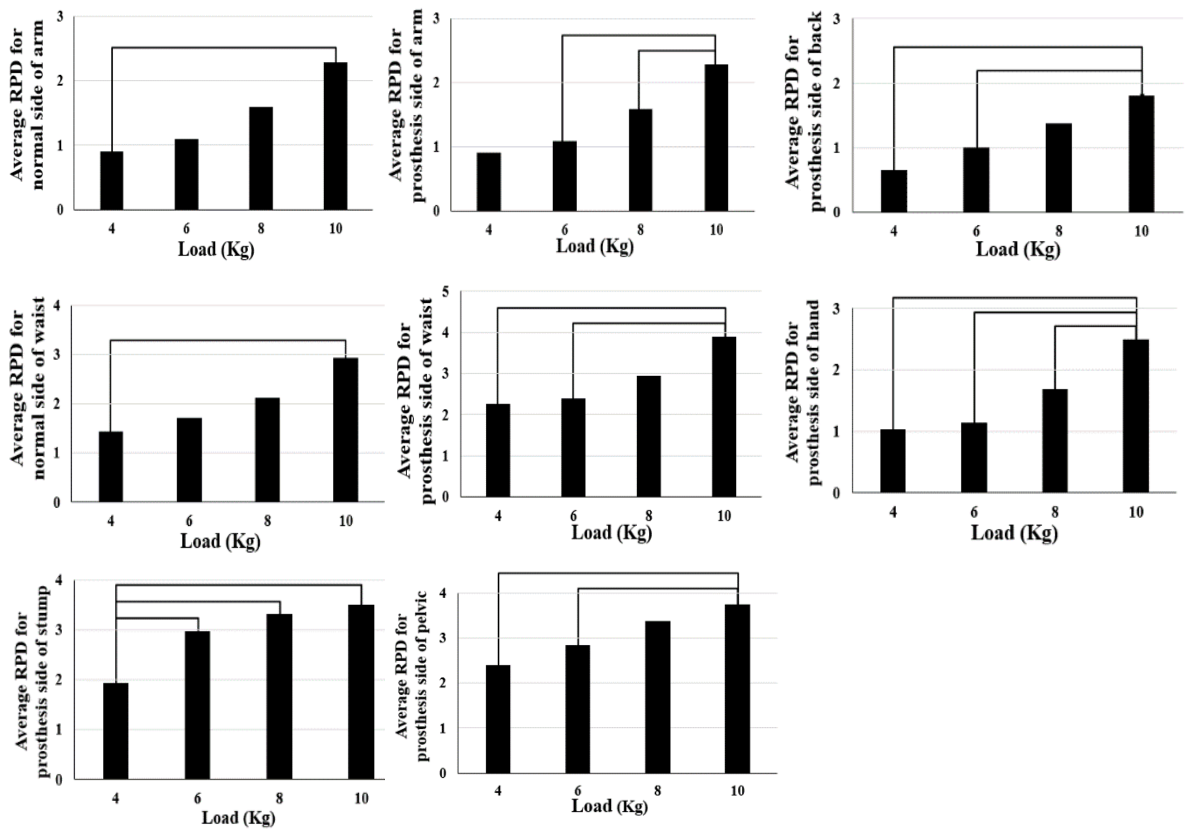

Figure 6. Average RPD for Prosthesis Side of the Back, Prosthesis Side of the Arm, Normal Side of the Arm, Prosthesis Side of the Hand, Prosthesis Side of the Waist, Normal Side of the Waist, Prosthesis Side of Pelvic, and Prosthesis of the Stump.

of the back and waist exhibited the highest average RPD values, followed closely by the prosthesis side of the arm and stump. In contrast, discomfort levels on the normal side of the body—particularly in the waist and arm—were lower. This pattern highlights a dominant discomfort concentration on the prosthetic side, especially in the posterior chain (back, waist, pelvic, stump). The upper body was more affected than the lower extremities, suggesting that compensatory upper-body movements play a key role in managing load imbalance during MMH tasks.

The presence of large effects in multiple regions, particularly on the prosthesis side, implies that the mechanical strain and compensatory movements required to manage heavier loads are not localized but distributed across the upper and lower body. These findings reinforce the need to consider the whole-body impact of MMH tasks for prosthetic users. Discomfort experienced in non-prosthetic regions also highlights the asymmetrical physical demands imposed by prosthesis use, underscoring the importance of ergonomically-informed prosthetic design and task adaptation to mitigate musculoskeletal strain during occupational activities.

Heavier loads intensify compensatory strategies such as lateral leaning, increasing mechanical strain and discomfort on the prosthetic side [15], [41]. These effects are particularly pronounced in the back, pelvis, and stump due to altered posture under load. Additional mechanical demands also contribute to muscle fatigue and joint discomfort in regions responsible for maintaining balance and stability [19]. Ko et al. [45] support these observations, noting that poor socket fit and uneven pressure distribution amplify discomfort and instability, particularly during dynamic tasks. This finding also aligns with Orlando [2], who emphasize that user discomfort reflects the functionality and usability of lower extremity devices, often arising when prosthetic systems fail to meet ergonomic demands.

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Table 6. Results of Repeated Measures ANOVA test for Perception of Steadiness

Criteria	**Effect**					
	Type of Container		**Load**		**Type of Container × load**	
	$F_{(1,7)}$	η_p^2	$F_{(3,21)}$	η_p^2	$F_{(3,21)}$	η_p^2
Stability	0.000	0.000	17.846***	0.719	0.304	0.042

Note. ***Significant at p<0.001

A significantly lower load-bearing capacity was observed on the prosthetic side compared to the normal side. Interestingly, neither container type nor the interaction between container type and load weight significantly influenced load-bearing capacity. This suggests that using boxes with or without handles resulted in minimal differences in load distribution. One possible explanation is the limited ergonomic variation between container types, which likely enabled participants to adjust their grip or posture without substantially affecting perceived discomfort. Furthermore, the short duration of tasks and the inclusion of rest intervals may have mitigated fatigue accumulation, potentially masking any subtle ergonomic effects. Participants may also have employed compensatory strategies to distribute the load more evenly, thereby minimizing discomfort. In body regions less directly involved in load handling—such as the lower limbs or neck—the influence of container ergonomics was likely even less pronounced.

Crucially, the significant impact of load weight on the prosthetic side of the back and arm appears to be linked to the outward-leaning posture adopted by participants. This compensatory strategy, commonly used to maintain balance during dynamic tasks, increases mechanical load on the prosthetic side, particularly in the back, pelvis, and stump [15], [33]. Such altered postures intensify discomfort by placing additional strain on supporting muscles and joints [19].

Effect of Load Mass and Container Type on Perceived Steadiness

Regarding steadiness, the weight of the load carried during MMH significantly affected the participants' perception of steadiness with a large effect size (F (3,21) = 17.846, p <0.001, η_p^2 = 0.719), as shown in Table 6. This substantial effect size indicates the practical relevance that load mass variations have a strong and consistent impact on users' perceived balance while using the transfemoral prosthesis. Post-hoc analysis (Figure 7) revealed a clear trend: as the load weight increased, participants reported lower levels of stability while using the transfemoral prosthesis for

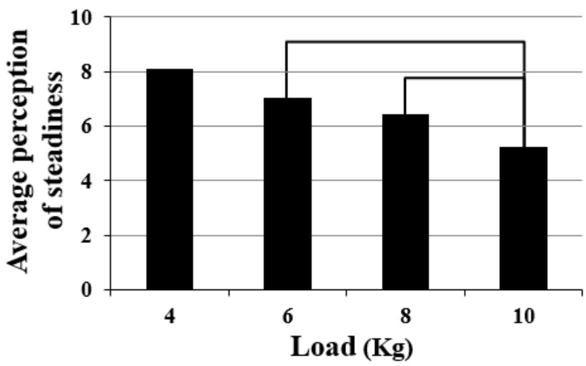

Figure 7. Average Perception of Steadiness

MMH tasks. A notable decline in stability was observed with loads exceeding 4 kg. This decrease in stability can be attributed to the complex coordination required among the ankles, knees, and hips to maintain balance.

The absence of natural ankle and knee functionality in individuals with above-knee amputations, as emphasized by [46], significantly compounds the challenge of maintaining balance during dynamic activities like MMH. Prior studies [15], [19] also emphasize how prosthetic design limitations lead to increased reliance on compensatory strategies, such as lateral trunk bending and hip engagement, which intensify the difficulty in maintaining steadiness during load-carrying tasks.

Implications for Design

The evaluation results, encompassing POGS criteria and two subjective assessments, provided valuable insights for refining the design of the transfemoral prosthesis. These suggestions, tailored to the impact of container type and load weight and their interaction during MMH tasks, focus on enhancing four components of the transfemoral prosthesis: the socket, knee joint, foot, and belt [26]. To improve socket performance, proper alignment is essential for minimizing lateral trunk bending and stabilizing the femur, particularly in users with shorter stumps. An ischial containment (IC) socket with surface curvature and an adduction angle of 7–10° is recommended to prevent excessive outward leaning on the prosthetic side [47], [48]. Additionally, ensuring adequate lateral support within the socket can help reduce discomfort and improve load distribution during dynamic tasks such as MMH [19]. For improved durability, the current iron block connector at the socket base should be replaced with a stainless-steel pyramid connector with four fastening holes.

Regarding the knee joint, terminal swing impact, caused by excessive knee straightening during the swing phase, remains a critical issue [15]. Adjustments are necessary to mitigate shocks caused by excessive knee straightening during the swing phase, which results in terminal swing impact and instability. Adding friction to the knee joint can increase surface roughness, preventing such shocks and improving stability. Reducing the gap between the two connector bars, leaving approximately ±2 mm on the right and left sides of the knee joint, is also suggested to enhance joint stability.

For the foot component, dual-keel soles are recommended for their ability to provide consistent support and comfort during both walking and load-bearing activities [49]. Proper foot alignment is critical for achieving smooth heel contact and avoiding gait deviations such as foot slap or heel compression, which can compromise comfort and stability during MMH tasks. Highsmith et al. [41] emphasize that component-specific gait training and appropriate foot prescription are essential for improving joint kinematics and minimizing biomechanical inefficiencies during dynamic activities.

Regarding the belt, reinforcing the stitching on the existing belt is suggested to enhance its strength. Alternatively, switching to a total elastic suspension (TES) belt is recommended for its ability to securely attach the prosthesis while providing long-term comfort during extended use. A secure and comfortable suspension mechanism is essential for maintaining prosthetic stability and reducing strain during prolonged activity [19].

Based on the results of the gait and discomfort assessments, we propose a prioritization of these four design components. The socket is ranked highest due to its significant influence on pelvic stability and discomfort in the back and stump. The knee joint follows, being closely associated with terminal swing impacts and gait symmetry. The foot is placed third, primarily for its influence on heel strike and foot rollover. Lastly, the belt system, though important for suspension and alignment, is ranked fourth in terms of direct biomechanical impact. This prioritization aims to guide designers in targeting the most impactful and feasible modifications, particularly for individuals engaged in MMH tasks.

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Limitations and Future Direction

This study provides valuable insights into transfemoral prostheses during manual material handling (MMH) tasks but has several limitations. The small, all-male sample limits the generalizability of the findings, particularly with respect to potential gender differences in prosthetic gait patterns. As this study was exploratory, no a priori power analysis was conducted. Future research should include a more diverse participant group and consider formal power calculations based on preliminary effect sizes. Additionally, the experimental conditions may not fully replicate real-world scenarios, such as uneven terrain, workplace hazards, or environmental variables. The findings are also specific to the 4-bar linkage D2 knee joint prosthesis and may not extend to other prosthetic designs, including advanced models. While subjective measures like perceived discomfort and stability provide useful user insights, they may introduce bias and could be complemented with objective tools such as motion analysis or EMG for a more comprehensive evaluation. Future research should aim to expand both the sample size and demographic diversity, including female participants, to better assess variability in prosthetic performance. Studies simulating real-world environments and comparing a wider range of prosthetic designs—including advanced and customizable components—are also recommended. Longitudinal studies incorporating advanced biomechanical tools such as 3D motion capture, electromyography (EMG), and wearable sensors could provide deeper insights into user adaptation, gait dynamics, and device durability over time [27], [45].

CONCLUSION

This study evaluated the performance of a transfemoral prosthesis in manual material handling (MMH) tasks. Gait analysis revealed that lateral trunk lean (POGS Criterion 3) was significantly affected by the interaction between container type and load mass, indicating compensatory upper-body strategies. Other gait deviations, including knee position at terminal swing and heel strike behavior, were observed but did not reach statistical significance. RPD results revealed elevated discomfort in the back, pelvis, and stump regions particularly on the prosthesis side, highlighting the socket as the most affected component. These findings point to the need for refinement of prosthetic design, particularly the socket, to improve balance, user comfort, and gait adaptability during dynamic tasks. Recommendations were derived through combined assessment of POGS, RPD, and perceived stability, focusing on four key components: the socket, knee joint, foot, and belt.

While the findings offer valuable insights, their generalizability is limited by the small, all-male sample and constrained experimental setting. Future research should incorporate more diverse users, simulate real-world tasks, and employ objective tools such as EMG or 3D motion capture to complement user-reported outcomes. Enhancing prosthetic designs for functional MMH use has important implications for workplace inclusion, potentially supporting job retention and equitable access to physically demanding occupations for individuals with lower-limb loss.

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CONFLICT OF INTEREST

The authors declare no conflict of interest regarding the publication of this paper.

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AUTHOR(S) BIOGRAPHY

Khoirul Muslim completed his bachelor's degree in industrial engineering at Bandung Institute of Technology (ITB) in 2007. He pursued further studies at Virginia Polytechnic Institute and State University, earning a master's degree in 2011 and a Ph.D. in 2013. He is an Assistant Professor in the Industrial Engineering Department at ITB, Indonesia. He is also the Technical Committee Coordinator for the Indonesian Ergonomic Association (PEI). His research interests include industrial ergonomics and occupational biomechanics.

Adinda Shafira earned her bachelor's degree in Industrial Engineering from Bandung Institute of Technology (ITB) in 2020. During her undergraduate years, she actively contributed as an assistant at the Laboratory for Work Engineering and Ergonomics, where she developed a strong foundation in the principles and applications of ergonomics in industrial settings. Adinda's research interests lie in industrial ergonomics, particularly in optimizing human performance, safety, and well-being within work environments.

Rifko Rahmat Kurnianto graduated with both a bachelor's and a master's degree in Mechanical Engineering from Bandung Institute of Technology (ITB). He is pursuing his Ph.D. at the University of Groningen, focusing on advancing knowledge in biomechanics. Rifko's research interests center on biomechanics, optimizing human movement and performance in various environments. He aims to contribute to developing technologies and methodologies that improve human health, safety, and efficiency through a deeper understanding of mechanical principles in biological systems.

Sandro Mihradi completed his bachelor's degree in mechanical engineering at the Bandung Institute of Technology (ITB) in 2001. He continued his studies at Toyohashi University of Technology, Japan, earning a master's degree in 2003 and a doctoral degree in 2006. He is currently a Senior Lecturer in the Faculty of Mechanical and Aerospace Engineering at ITB, specializing in the Dynamics and Control Research Group. His research interests span mechanical design, biomedical engineering, and renewable energy systems, with recent works focusing on robotic prosthetics, energy storage systems, and safety in transportation design.

Auditya Purwandini Sutarto is a lecturer at Department of Industrial Engineering,Universitas Qomaruddin, Gresik, Indonesia. She received her B.S. in Industrial Engineering from Institut Teknologi Bandung, a Master of Science in Mathematics from Universitas Gadjah Mada, and PhD in Technology Management from Universiti Malaysia Pahang, Malaysia. Her research contributions span ergonomics, psychophysiology, and artificial intelligence, reflecting a deep commitment to interdisciplinary study.