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Neuromuscular Control of Overground Walking in Transtibial Amputees: Endoskeletal vs. Exoskeletal Prostheses

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Abstract

Background: Transtibial prostheses are commonly classified as endoskeletal or exoskeletal and differ in weight, adaptability, and mechanical response, which may influence gait performance. This study examined whether prosthesis type affects overground walking movement structure and neuromuscular control and assessed the relationship between walking speed and neuromuscular control. **Methods:** Principal component analysis (PCA) was applied to kinematic marker data from 20 unilateral transtibial amputees using either endoskeletal (n = 10; 54.7 ± 6.1 years) or exoskeletal prostheses (n = 10; 57.9 ± 8.7 years) during self-selected overground walking. Principal movements (PMs) were extracted to represent functionally meaningful gait components. Movement structure was evaluated using the relative explained variance of PM positions (rVAR), whereas neuromuscular control was quantified using the root mean square of PM accelerations (RMS; acceleration magnitude) and the number of zero crossings (N; regularity/predictability). Group differences were examined using covariate-adjusted analyses, controlling for preferred walking speed. **Results:** No significant differences in walking movement structure were found between prosthetic types. Unadjusted analyses suggested greater swing-phase acceleration (PM_2) and lower neuromuscular variability across PM_1–PM_4 in the endoskeletal group; however, these effects were no longer significant after adjusting for BMI and walking speed. Walking speed showed strong associations with neuromuscular control ($p \leq 0.003$), with faster speeds linked to greater swing-phase acceleration and reduced variability. **Conclusions:** Walking movement structure and neuromuscular control were comparable between prosthetic types, while walking speed emerged as a key factor in gait evaluation among transtibial amputees.

Keywords: transtibial prosthesis; endoskeletal prosthesis; exoskeletal prosthesis; gait; movement synergies; neuromuscular control; principal component analysis (PCA)

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1. Introduction

A prosthesis is a medical device designed to replace a missing body part and restore functional capability following loss due to injury or disease. Transtibial prostheses replace the limb segment below the knee and are commonly recommended to help individuals regain mobility and resume daily activities after amputation [1–3]. They typically consist of four key components [4]: a socket that interfaces with the residual limb [5], a suspension system that secures the prosthesis [6], a pylon that provides structural support [7], and a prosthetic foot that facilitates balance, shock absorption, and energy return [8]. Transtibial prostheses are generally classified as endoskeletal or exoskeletal [4]. Endoskeletal designs

use a modular internal pylon with interchangeable components, allowing precise alignment adjustment, reduced weight, and easier maintenance, often in combination with Pe-lite or silicone liners for comfort [9,10]. Exoskeletal prostheses, by contrast, have a rigid outer shell that offers greater durability and impact resistance but increases weight and limits adjustability [9,11]. Despite these design differences, no universally optimal prosthetic configuration has been identified, and no single component has demonstrated consistent superiority across all transtibial amputees [6,12].

Differences in prosthetic design influence walking patterns and may contribute to gait asymmetry and misalignment [13–16]. Asymmetry commonly arises from mechanical differences and neuromuscular adaptation [17,18], with gait strategies further influenced by walking speed, limb laterality, and prosthetic foot design [14,19]. These factors lead to variations in stride duration, stance time, balance, stride length, and joint motion, while the absence of plantar flexors increases mechanical demand at the hip, resulting in compensatory movement patterns [14,19]. Although dynamic-response feet improve propulsion, they remain less efficient than the biological limb [14,19]. However, most previous research has primarily examined joint kinematics, kinetics, and spatiotemporal parameters, providing limited insight into how prosthetic design influences whole-body coordination and neuromuscular control [13,17,19]. Consequently, little is known about how different transtibial prosthetic types affect movement synergies and neuromuscular stability across gait phases, despite mechanical differences that may shape neural coordination and gait variability. Strengthening neuromechanical evidence is therefore essential to support more informed prosthetic prescription and rehabilitation planning.

Movement synergies refer to the coordinated activation of multiple muscles to produce efficient and stable movement, reducing redundancy by allowing muscles to operate as functional units [16,20]. Neuromuscular control within these synergies reflects the integration of sensory input, motor planning, and execution, and is therefore essential for evaluating gait stability and balance [20,21]. During locomotion, these synergies coordinate motion across joints, facilitate smooth transitions between stance and swing phases, and interact with passive body dynamics to minimize continuous active muscular control [22]. Passive dynamics arise from the body's intrinsic mechanical properties—such as gravity, inertia, and elasticity—and help reduce neuromuscular demand while enhancing movement efficiency [23]. To characterize these coordinated motor strategies, principal component analysis (PCA) can be applied to decompose complex kinematic data into orthogonal principal movements (PMs), each representing a functionally meaningful gait component—such as stance or swing—while explaining most of the total variance [22]. PCA therefore offers an alternative and complementary approach to conventional gait analysis, enabling detailed evaluation of motor coordination and neuromuscular control through PM positions and accelerations [24,24]. In this context, movement structure refers to the composition and relative contribution of each PM to overall gait coordination, whereas neuromuscular control reflects the nervous system's ability to generate consistent and well-regulated acceleration patterns to maintain stable and efficient movement [22,24]. Thus, applying PCA to transtibial amputees may provide deeper insight into how prosthetic design influences movement synergies, neuromuscular stability, and overall gait performance.

In summary, this study aimed to investigate whether different transtibial prosthesis designs influence overground walking performance by examining both movement composition/structure and the neuromuscular control of individual principal movements (PMs). Given the lighter weight, modularity, and potentially more dynamic mechanical response of endoskeletal designs [25], it was hypothesized that users of these prostheses

would demonstrate more efficient neuromuscular control, characterized by greater acceleration in functionally relevant gait phases and more consistent, predictable movement patterns, compared with users of exoskeletal prostheses. In addition, the relationship between preferred walking speed and neuromuscular control was examined, with the expectation that walking speed would be associated with specific PMs that play a key role in functional gait performance [19]. Clarifying these neuromechanical mechanisms may provide an alternative perspective for guiding prosthetic selection and refining rehabilitation strategies aimed at improving gait efficiency, stability, and mobility in individuals with transtibial amputation.

2. Materials and Methods

2.1. Participants and Experimental Procedures

This study represents a secondary analysis of an openly available dataset published by Samala et al. [26]. No additional data collection was performed, and no new participants were recruited for the present investigation. The raw kinematic marker trajectories were reanalyzed using a PCA framework to address a distinct research question focused on movement synergies and neuromuscular control.

The dataset included 10 individuals using endoskeletal prostheses (54.7 $\pm$ 6.1 years) and 10 using exoskeletal prostheses (57.9 $\pm$ 8.7 years), all recruited from the Sirindhorn School of Prosthetics and Orthotics Clinic, Thailand. Inclusion criteria required unilateral transtibial amputation for $\geq$6 months, medium-to-long residual limb length, Medicare Functional Classification Level (MFCL) K1–K4, and independent ambulation of at least 10 m without gait aids. Participants were required to have normal lower-limb range of motion (American Academy of Orthopaedic Surgeons (AAOS) criteria), muscle strength graded 4–5 on the Oxford scale, and the ability to understand written and spoken Thai. Exclusion criteria included residual limb pain or neuroma; open wounds, congestion, or major limb volume fluctuation; uncontrolled diabetes or hypertension; severe systemic illness; contralateral limb disorders affecting gait; joint contractures; or balance impairments such as dizziness or vertigo. The original study complied with the Declaration of Helsinki and was approved by the Institutional Review Board of the Faculty of Medicine, Siriraj Hospital, Mahidol University, Thailand (protocol code 892/2564 (IRB3), approval date: 10 January 2022). Written informed consent was obtained from all participants in the original investigation.

The present analysis did not combine this dataset with any additional data sources. Only the kinematic trajectories required for the PCA-based analysis were extracted. Trial selection and preprocessing procedures were conducted independently in accordance with the methodological requirements of the current study. In the original protocol [26], data were collected during a single session while participants wore their own prostheses without modification of components such as the foot, suspension system, or liner. A certified prosthetist inspected each prosthesis prior to testing to ensure proper function and safety. Cluster markers were placed bilaterally at key anatomical landmarks. On the sound limb, markers were positioned at the anterior superior iliac spine (ASIS), posterior superior iliac spine (PSIS), pelvis, thigh, shank, knee, ankle, and foot. On the prosthetic limb, markers were placed at corresponding socket and prosthetic foot locations, with adjustments made as necessary to accommodate component geometry.

Participants completed five walking trials along a 10 m indoor walkway at a self-selected speed, initiated by an audio cue. Walking was performed in both directions along the walkway; however, for the present analysis, three trials performed in the same walking direction were selected to ensure directional consistency within the PCA framework. A

3 m mid-section of each trial was captured using eight Raptor infrared cameras (200 Hz; Motion Analysis Corporation, Santa Rosa, CA, USA). Data were processed in Cortex and analyzed in Visual 3D (C-Motion Inc., Germantown, MD, USA). From the five trials, the three most complete trials (without missing marker data) in the selected direction were retained for analysis. Participant and prosthetic characteristics are summarized in Table 1.

Table 1. Characteristics of participants including endoskeletal and exoskeletal prostheses (mean ± SD, * $p < 0.05$).

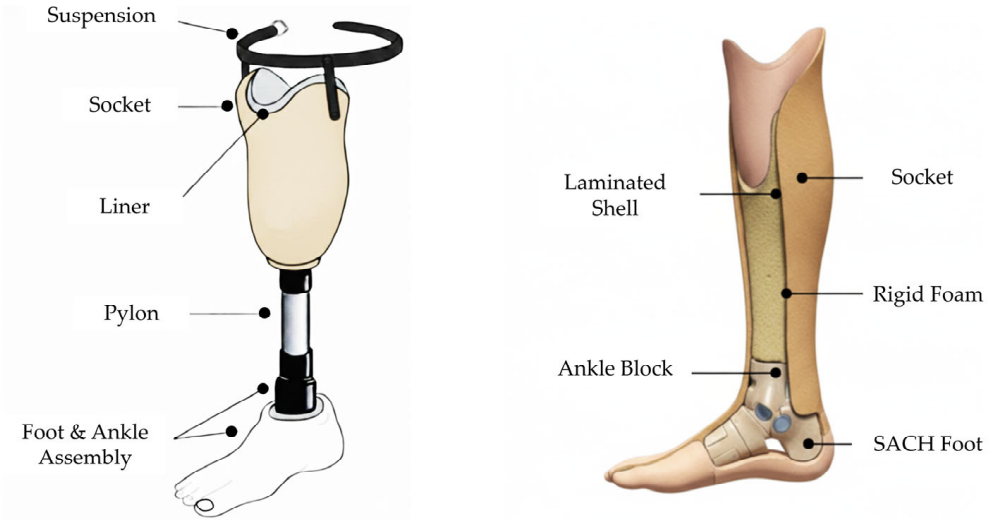
Variable	Endoskeletal (n = 10)	Exoskeletal (n = 10)	*p*-Value
Age (yrs)	54.7 ± 6.1	57.9 ± 8.7	0.425
Body mass (kg)	70.4 ± 13.9	64.8 ± 12.3	0.466
Height (m)	1.65 ± 0.1	1.65 ± 0.1	0.825
Body mass index (kg/m^2)	25.7 ± 3.8	23.7 ± 3.1	0.339
Amputated side (Left/Right)	5/5	3/7	-
Preferred walking speed (m/s)	1.7 ± 0.2	1.4 ± 0.2	0.020 *
Prosthetic characteristics			
- Socket type (PTB-SC/TSB-SC/PTB/TSB)	7/1/2/0	2/2/3/3	-
- Liner type (Pe-lite/Silicone/Aero liner)	10/0	7/2/1	-
- Suspension system (Cuff suspension/Self suspension/Sleeve)	4/6/0	5/4/1	-
- Foot type (SACH/Single axis/sPace)	10/0/0	8/1/1	-

2.2. Prosthetic Characteristics

As shown in Table 1, endoskeletal prostheses demonstrated relatively uniform structural and functional characteristics. Most participants used Patellar Tendon Bearing–Supracondylar (PTB-SC) sockets (n = 7), which enhance suspension by engaging the femoral condyles. One participant used a TSB-SC socket (n = 1), and two used PTB sockets (n = 2). All endoskeletal users wore Pe-lite liners (n = 10), a closed-cell foam material that provides cushioning but does not offer intrinsic suspension. Self-suspension was the most common suspension method (n = 6), relying on anatomical contouring to secure the prosthesis without external sleeves or straps, whereas the remaining participants used cuff suspension (n = 4). All individuals in this group wore Solid Ankle Cushion Heel (SACH) feet (n = 10), which are durable and simple in design but do not provide energy return.

In contrast, exoskeletal prostheses demonstrated greater variability in component configuration. Socket designs included Patellar Tendon Bearing (PTB) (n = 3), Total Surface Bearing (TSB) (n = 3), and TSB-Supracondylar (TSB-SC) (n = 2), reflecting differences in load distribution and suspension characteristics. Liner materials varied and included Pe-lite (n = 7), silicone liners (n = 2), which enhance skin adherence and reduce shear, and one Aero liner (n = 1), designed for lightweight and breathable comfort. Cuff suspension was the most common suspension system (n = 5), followed by self-suspension (n = 4), while one participant used sleeve suspension (n = 1). Most participants wore Solid Ankle Cushion Heel (SACH) feet (n = 8), whereas two used more dynamic alternatives: a single-axis foot (n = 1), improving ankle articulation and ground clearance, and a sPace foot (n = 1), designed to provide greater energy return and shock absorption. The inclusion of these dynamic feet may reflect higher functional demands within the exoskeletal group.

Figure 1 illustrates the general structural components of both transtibial prosthetic types; however, it does not represent the exact component configurations used by participants in the present study.



A: Endoskeletal Transtibial Prosthesis **B:** Exoskeletal Transtibial Prosthesis

Figure 1. Schematic illustration of the basic structural components of transtibial prostheses. (**A**) Endoskeletal transtibial prosthesis showing modular internal pylon construction. (**B**) Exoskeletal transtibial prosthesis showing rigid laminated shell with foam core and integrated ankle block. This illustration presents the general structural concepts of both prosthetic types and does not represent the exact component configurations used by participants in the present study.

2.3. Decomposing Movement Synergies

The quality of all kinematic data was first visually inspected from the C3D files using Mokka [27] to ensure marker visibility, trajectory continuity, and absence of artifacts. Only trials with complete and reliable marker trajectories were retained and exported in CSV format for PCA. During preprocessing, cluster markers on the pelvis, thighs, and lower legs were excluded due to redundancy and inconsistent signal quality. Twelve anatomical markers were retained (Figure 2), including the anterior superior iliac spines (ASIS), posterior superior iliac spines (PSIS), lateral knees, lateral malleoli, heels, and first metatarsal heads, which sufficiently defined the lower-limb kinematic configuration. Each dataset was centered by subtracting the mean posture vector and normalized to the mean Euclidean distance [28]. Three walking trials per participant were then concatenated, generating a shared input matrix (3 trials × 2 prosthesis types × 20 participants) that enabled direct between-group comparison within a common principal movement space.

PCA was performed using PManalyzer software [28] in MATLAB™ R2024a (MathWorks Inc., Natick, MA, USA). The analysis decomposed the 3D kinematic data into orthogonal eigenvectors (PC_k), each representing a principal movement component (PM_k), where k denotes the order of movement components. Each PM_k was visualized as an animated stick figure illustrating its corresponding movement pattern. The associated eigenvalues reflected the variance explained by each PM_k, and the time-dependent PC scores ($PP_k(t)$) described how each PM evolved across time. Principal accelerations ($PA_k(t)$) were obtained via double differentiation of $PP_k(t)$, providing a proxy for neuromuscular control due to their established association with myoelectric activity [29]. To reduce noise amplification, all PCA-derived time series were processed using a 3rd-order zero-phase 10 Hz low-pass Butterworth filter prior to differentiation. Leave-one-out cross-validation was employed to evaluate the robustness of individual PM_k and PCA-derived variables against input data variation [28]. The first five PCs, which together accounted for 97.7% of the total variance and demonstrated stability under cross-validation, were retained for hypothesis testing.

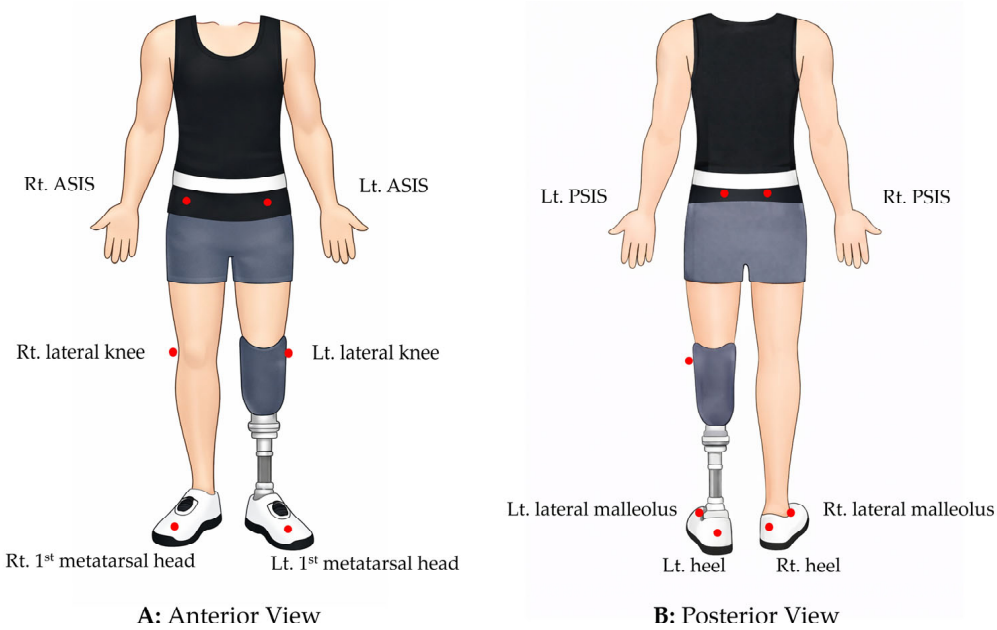


Figure 2. Marker placement configuration used for PCA. Twelve reflective markers (red dots) were positioned bilaterally at key anatomical landmarks defining the lower-limb kinematic configuration, shown in (**A**) anterior and (**B**) posterior views.

Movement composition (synergy structure) was quantified using participant-specific relative explained variance (PP_k_rVAR), derived from the time series of principal positions $PP_k(t)$. PP_k_rVAR indicates the contribution of each PM_k to the total postural variance [28] and reflects its functional role in locomotor coordination [22,24]. For statistical analysis, the mean PP_k_rVAR across three walking trials per participant was used.

Neuromuscular control was assessed using two acceleration-based metrics per trial: the root mean square of principal accelerations (PA_RMS), representing acceleration magnitude [24,30], and the number of zero-crossings (PA_N), reflecting the regularity and predictability of acceleration patterns [24]. Both variables were normalized to walking speed [22,24,30]. Acceleration integrates muscular effort, gravitational forces, and postural adjustments, making it a key mechanical variable regulated by the sensorimotor system and an indicator of neuromuscular control [29].

From a motor-control perspective, body motion changes only when muscle-generated forces interact with gravity and passive body dynamics to generate acceleration [31,32]. Thus, regulating acceleration is a fundamental task of the neuromuscular system during posture and locomotion [33]. Within the PCA framework, whole-body kinematics are decomposed into principal movements, and their corresponding principal accelerations (PAs) characterize the dynamic control of specific movement components [23]. Previous work has demonstrated measurable relationships between PA time series and muscle activation patterns, supporting their interpretation as physiologically meaningful indicators of neuromuscular regulation rather than purely mechanical descriptors [29]. Accordingly, PA_RMS reflects the magnitude of neuromuscular output required to generate movement, whereas PA_N captures the temporal regularity of corrective adjustments, indicating how consistently the nervous system regulates acceleration patterns over time. In this context, neuromuscular control refers to the nervous system's ability to generate stable and well-regulated acceleration patterns to maintain dynamic stability during walking. Higher PA_N values indicate more irregular or unstable control, whereas lower PA_N values reflect smoother and more predictable regulation [28,34]. As with PP_k_rVAR, mean values across three trials per participant were used for statistical comparison.

2.4. Statistical Analysis

Statistical analyses were performed using SPSS software version 26.0 (IBM SPSS Statistics, SPSS Inc., Chicago, IL, USA), with the level of statistical significance set at $\alpha = 0.05$. Data normality was assessed using the Shapiro–Wilk test. Group comparisons between endoskeletal and exoskeletal users were conducted separately for each variable. Normally distributed movement structure variables ($PP_{1–5}$_rVAR) were analyzed using independent-samples *t*-tests, whereas non-normally distributed neuromuscular control variables ($PA_{1–5}$_RMS and $PA_{1–5}$_N) were analyzed using Mann–Whitney U tests. Effect sizes were expressed as Cohen's d for parametric tests and rank biserial correlation (*r*) for non-parametric tests.

To account for potential confounding influences of participant characteristics, covariate-adjusted analyses were performed using a general linear model/ANCOVA framework, with BMI included as the primary covariate and prosthesis type entered as a fixed factor. Because walking trials were performed at a self-selected speed rather than a standardized velocity, preferred walking speed differed between participants and is biomechanically relevant to gait control; therefore, it was included as an additional covariate in separate exploratory models to account for its potential confounding influence. Estimated marginal means and partial eta-squared (η_p^2) were reported for all adjusted analyses. For multiple comparisons across $PM_{1–5}$, the Holm–Bonferroni procedure was applied to control the family-wise error rate [35]. Associations between walking speed and neuromuscular control variables were examined using Spearman's rank correlation (due to non-normality), and Holm–Bonferroni correction was applied within RMS and N variable sets to account for multiple testing.

3. Results

3.1. Walking Movement Synergies

As shown in Figure 3, the first principal movement (PM_1) accounted for the largest proportion of postural variance (PP_1_rVAR: 98.31 ± 0.30) and primarily represented coordinated lower-limb motion in the forward direction of walking, indicating its dominant role in maintaining overall gait progression. The second principal movement (PM_2) reflected the swing phase of gait, characterized by alternating anteroposterior limb motion; although functionally important, it contributed comparatively less variance (PP_2_rVAR: 1.46 ± 0.27). The third principal movement (PM_3) captured single-leg support during mid-stance, involving lateral weight transfer toward the stance limb, but explained only a small portion of postural variance (PP_3_rVAR: 0.08 ± 0.02). The fourth principal movement (PM_4) represented a double-leg support component marked by synchronized ankle and knee flexion–extension in the vertical plane, again contributing minimally to total variance (PP_4_rVAR: 0.07 ± 0.01). Finally, the fifth principal movement (PM_5) described another double-support strategy combining vertical ankle motion with lateral ankle displacement, accounting for the smallest proportion of variance (PP_5_rVAR: 0.05 ± 0.02). Visualizations of PM_1–PM_5 are presented in Figure 3, with animated stick-figure representations provided in Supplementary Video File S1.

3.2. Structures of Walking Movement Synergies

When comparing movement structures between endoskeletal and exoskeletal prostheses (Figure 4), no significant group differences were found. PP_1_rVAR ($p = 0.250$, $d = 0.530$), PP_2_rVAR ($p = 0.155$, $d = 0.660$), and PP_3_rVAR ($p = 0.150$, $d = 0.670$) showed moderate to large effect sizes, while PP_4_rVAR ($p = 0.688$, $d = 0.180$) and PP_5_rVAR ($p = 0.685$, $d = 0.180$) indicated negligible differences.

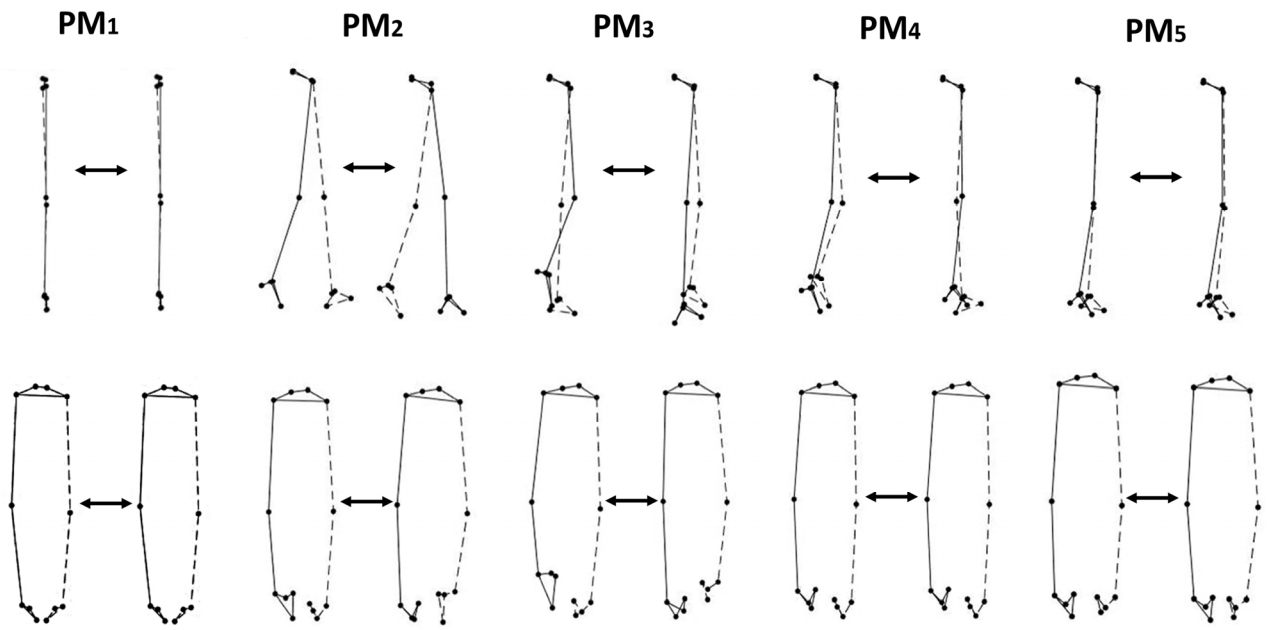


Figure 3. Visualization of the first to fifth principal movements ($PM_{1–5}$; columns) during overground walking, shown in the sagittal plane (**top row**) and frontal plane (**bottom row**). The sagittal view allows clearer observation of forward progression and swing-phase dynamics, whereas the frontal view highlights lateral weight shifting and stance stability. Images were generated from pooled trials of endoskeletal and exoskeletal transtibial prosthesis users. The dashed line represents the left lower limb. Each PM representation is accompanied by its relative explained variance of principal position (PP_k_rVAR, %).

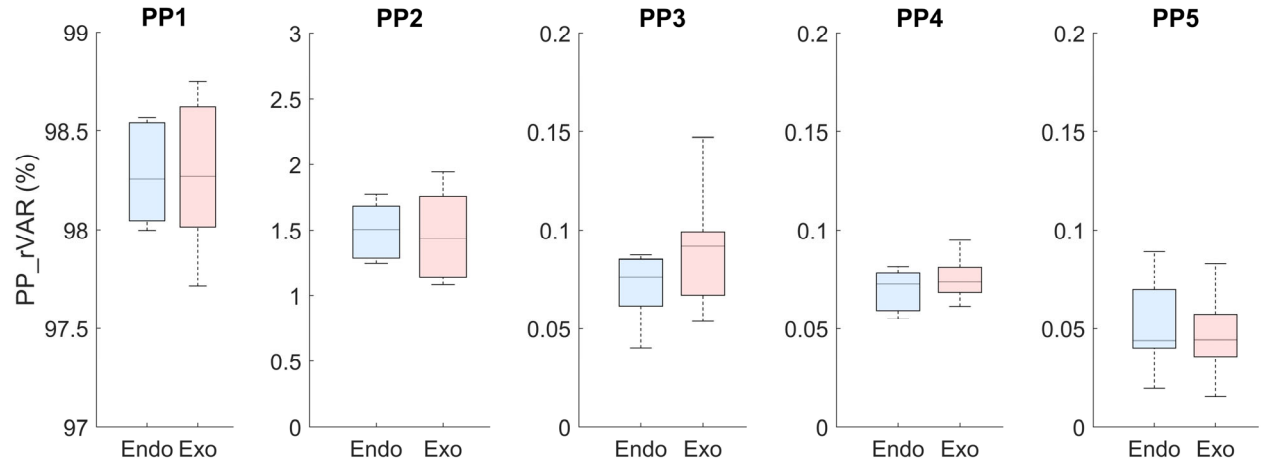


Figure 4. Comparison of the relative explained variance of principal positions ($PP_{1–5}$; PP_rVAR) between users of endoskeletal (Endo; blue) and exoskeletal (Exo; pink) transtibial prostheses. Box plots illustrate the median, interquartile range, and data distribution. Although PP_3 initially showed a significant between-group difference after adjusting for walking speed, this effect did not remain significant after applying Holm–Bonferroni correction across $PP_{1–5}$. Overall, no significant differences in movement structure were observed between groups.

After adjusting for walking speed using a GLM/ANCOVA framework, pairwise comparisons of the estimated marginal means confirmed the absence of significant between-group differences across $PP_{1–5}$ (all Bonferroni-adjusted $p \geq 0.167$). Although PP_3_rVAR showed a small trend toward a group effect, it did not reach statistical significance. Overall, these findings indicate that the structure of walking movement synergies was comparable between users of endoskeletal and exoskeletal transtibial prostheses.

3.3. Neuromuscular Control of Movement Synergies

As shown in Figure 5A, a significant group difference in acceleration magnitude was observed only for PA_2_RMS (p = 0.028, r = 0.490), indicating a medium effect size. In contrast, PA_1_RMS (p = 0.151, r = 0.320), PA_3_RMS (p = 0.290, r = 0.240), PA_4_RMS (p = 0.174, r = 0.300), and PA_5_RMS (p = 0.762, r = 0.070) did not reach statistical significance. However, when adjusting for preferred walking speed using a GLM/ANCOVA framework, none of the RMS components ($PA_{1–5}$) remained significantly different between groups (all Bonferroni-adjusted $p \geq 0.126$). These results suggest that the initially observed PA_2_RMS difference is likely influenced by differences in self-selected walking speed rather than reflecting an intrinsic effect of prosthesis type.

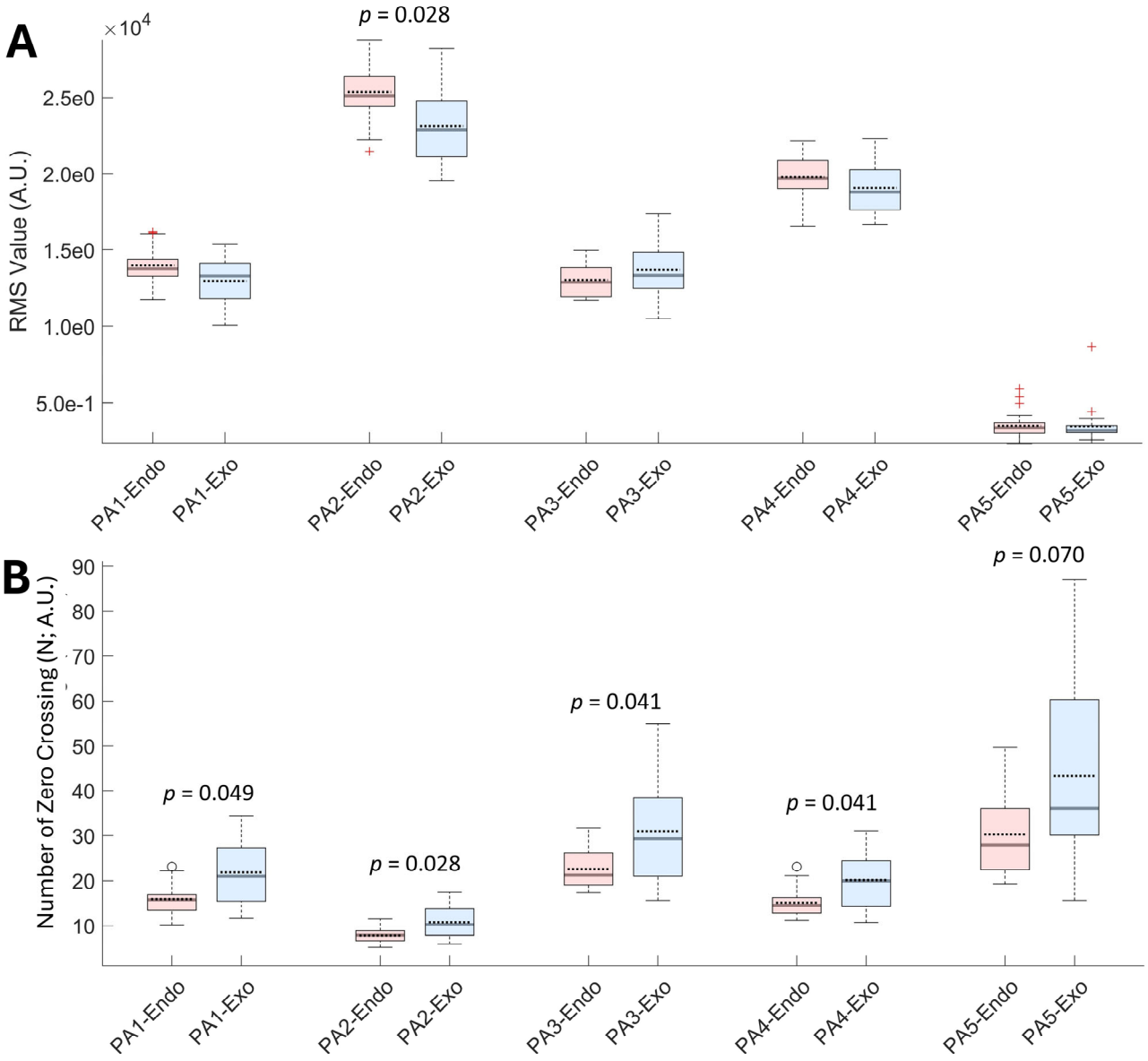


Figure 5. Box plots comparing neuromuscular control variables between transtibial amputees using endoskeletal (Endo; blue) and exoskeletal (Exo; pink) prostheses. (**A**) Normalized root mean square of principal accelerations ($PA_{1–5}$; RMS, A.U.), representing acceleration magnitude. (**B**) Normalized number of zero crossings of principal accelerations ($PA_{1–5}$; N, dimensionless), reflecting the regularity or predictability of acceleration patterns. p-Values indicate results from the unadjusted Mann–Whitney U tests. Although several differences reached statistical significance in the unadjusted comparisons, none remained significant after adjusting for preferred walking speed.

As shown in Figure 5B, significant group differences in neuromuscular variability (PA_k_N) were observed for PA_1_N (p = 0.049, r = 0.440), PA_2_N (p = 0.028, r = 0.490), PA_3_N (p = 0.041, r = 0.460), and PA_4_N (p = 0.041, r = 0.460), reflecting medium effects, whereas PA_5_N did not reach statistical significance (p = 0.070, r = 0.410). However, when adjusting for preferred walking speed using a GLM/ANCOVA framework, none of the NZC components ($PA_{1–5}$) remained significantly different between groups (all Bonferroni-adjusted $p \geq 0.396$), and all 95% confidence intervals included zero. These adjusted findings indicate that the initially observed differences in neuromuscular variability are likely influenced by differences in self-selected walking speed rather than prosthesis type, and that neuromuscular control consistency was ultimately comparable between endoskeletal and exoskeletal prosthesis users.

3.4. Correlation Between Neuromuscular Control of Movement Synergies and Walking Speed

As shown in Table 2, correlation analysis revealed significant relationships between walking speed and both the acceleration magnitude (PA_k_RMS) and neuromuscular variability (PA_k_N) of individual principal accelerations. Among the PA_k_RMS variables, PA_2_RMS showed a strong positive correlation with walking speed (ρ = 0.639, p = 0.002) and PA_3_RMS showed a strong negative correlation (ρ = −0.626, p = 0.003). Both correlations remained significant after Holm–Bonferroni correction (p_adj = 0.010 and 0.012, respectively), whereas no other RMS variables were significant.

Table 2. Spearman's rank correlation coefficients (ρ) and p-values for the relationship between walking speed and (A) normalized root mean square acceleration ($PA_{1–5}$_RMS) and (B) normalized number of zero crossings ($PA_{1–5}$_N). p-Values were adjusted for multiple comparisons using the Holm–Bonferroni procedure within each variable set (RMS and N) (* $p < 0.05$, ** $p \leq 0.001$).

Variable		A: RMS					B: N				
		PA_1	**PA_2**	**PA_3**	**PA_4**	**PA_5**	**PA_1**	**PA_2**	**PA_3**	**PA_4**	**PA_5**
Walking speed	ρ	−0.033	0.639	−0.626	0.310	−0.208	−0.940	−0.955	−0.952	−0.944	−0.890
	p	0.890	0.002 *	0.003 *	0.184	0.380	<0.001 **	<0.001 **	<0.001 **	<0.001 **	<0.001 **

In contrast, walking speed demonstrated very strong negative correlations with all PA_k_N variables (ρ between −0.890 and −0.955, all $p < 0.001$), and all associations remained highly significant after Holm–Bonferroni correction. These findings indicate that faster walking speeds are consistently associated with reduced neuromuscular variability, reflecting smoother, more predictable, and potentially more efficient control of movement components across gait phases.

4. Discussion

This study investigated walking movement synergies in transtibial amputees using endoskeletal and exoskeletal prostheses through PCA. Movement structure was examined using relative explained variance (PP_rVAR), while neuromuscular control was assessed using acceleration magnitude (PA_RMS) and neuromuscular variability (PA_N). The main finding is that the overall structure of walking movement synergies did not differ between prosthetic types. Moreover, although several unadjusted comparisons initially indicated group differences in acceleration magnitude and neuromuscular variability, these effects disappeared after adjusting for preferred walking speed. Thus, once walking speed was controlled, neuromuscular control characteristics were comparable between endoskeletal and exoskeletal users.

Although movement structures were similar between groups, the initial unadjusted analyses suggested differences in neuromuscular control between prosthesis users. How-

ever, these differences were no longer present following adjustment for preferred walking speed, indicating that the observed variations in acceleration magnitude and neuromuscular regularity were more strongly related to gait velocity than to prosthetic construction itself. The comparable movement structure between groups likely reflects the ability of both prosthetic systems to provide adequate functional support and alignment to maintain coordinated stance–swing organization despite differences in mechanical design. The faster preferred walking speed observed in the endoskeletal group may therefore represent user-related or behavioral influences rather than an inherent mechanical advantage of one prosthetic type.

Considering walking speed, three main observations can be highlighted. First, the strong positive correlation between walking speed and PM_2 acceleration magnitude (PA_2_RMS) indicates that faster walking is associated with greater swing-phase acceleration, reflecting enhanced limb advancement, ground clearance, and overall walking efficiency [36]. Second, the strong negative correlation between walking speed and PM_3 acceleration magnitude (PA_3_RMS), representing single-limb support, suggests that faster walking is accompanied by more controlled acceleration behavior during stance, indicating improved stability in this critical support phase [22]. Third, the very strong negative correlations between walking speed and PA_k_N across $PM_{1–5}$ demonstrate that higher walking speeds are consistently associated with more regular and predictable movement patterns, reflecting more efficient neuromuscular coordination. These findings reinforce walking speed as a sensitive indicator of both gait performance and neuromuscular control in transtibial amputees, consistent with previous evidence linking higher walking speeds to increased biomechanical demand and coordinated control strategies [37]. Importantly, these strong relationships suggest that individual functional capacity (e.g., preferred walking speed) may play a greater role in shaping neuromuscular control strategies than prosthesis type alone, which is particularly relevant given the heterogeneity in prosthetic design across participants.

From a clinical perspective, these findings suggest that prosthesis type alone may not be the dominant determinant of neuromuscular control during steady-state walking. Instead, gait performance appears closely associated with preferred walking speed [38], which itself may be influenced by multiple interacting factors including user confidence, prosthesis familiarity, comfort, and physical capacity. Rehabilitation should therefore continue to emphasize strategies that enhance safe walking capacity and coordination while progressively encouraging the development of more efficient self-selected walking speeds. In practice, this may include strengthening of key lower-limb musculature, task-specific gait re-education, and balance training to improve postural stability and reduce variability in movement execution. Functional task practice, step training, and repetitive walking exposure may further help individuals develop consistent neuromuscular control patterns [39]. In addition, because gait variability typically increases under more challenging environmental conditions [40–42], incorporating real-world walking tasks such as uneven terrain, directional changes, or variable walking speeds into rehabilitation and assessment may provide more ecologically valid insights into functional gait control and support better transfer to daily-life mobility.

This study has several limitations. First, variations in suspension methods and socket designs may have influenced the findings, as these factors can affect gait asymmetry, joint kinematics, and overall gait parameters [43,44]. The dataset also lacked information on prosthetic weight, which may alter limb inertia and acceleration profiles [26], highlighting the need to examine its effects on proprioception and muscle activation in future research. Second, the absence of a healthy control group limits interpretation of how walking synergies in amputees differ from those in non-amputees. Third, the lack of markers on the

head, trunk, and upper extremities restricted assessment of whole-body contributions to gait and balance [34]. Fourth, the use of participants' own prostheses introduced substantial variability in component type, alignment, and mechanical properties [23], which may have contributed to the difficulty in drawing clearer prosthesis-specific conclusions. Additionally, several clinically relevant parameters were unavailable, including time since amputation, cause of limb loss, prosthesis experience, and detailed MFCL K-level classification. These factors are known to influence gait capacity and neuromuscular control and may help to better characterize the most challenging cases. The relatively small sample size (n = 10 per group) further limits generalizability and the ability to detect subtle differences. Finally, the scope of the present work was limited to binary group comparisons due to dataset constraints. Future studies should incorporate more detailed prosthesis classification (multi-class analysis), introduce subgrouping approaches based on user characteristics, and apply unsupervised techniques such as clustering to better account for heterogeneity in prosthesis design and user adaptation patterns. In addition, although PCA provides an interpretable linear decomposition of coordinated movement patterns [23], alternative dimensionality-reduction approaches such as independent component analysis (ICA) or nonlinear extensions (e.g., kernel-based methods) may offer complementary perspectives by targeting statistical independence or nonlinear manifold structure [45,46]. Exploring such approaches in larger and more diverse datasets may further elucidate the complexity of neuromuscular control in prosthetic gait.

5. Conclusions

This study compared overground walking performance in unilateral transtibial amputees using endoskeletal and exoskeletal prostheses by applying PCA to characterize movement structure and neuromuscular control. The overall organization of walking movement synergies was similar between the two prosthetic groups. Although initial unadjusted analyses suggested differences in acceleration magnitude and neuromuscular variability, these effects were no longer present after adjusting for preferred walking speed, indicating that gait velocity rather than prosthesis construction was the primary factor underlying the observed differences. Walking speed demonstrated strong and systematic associations with neuromuscular performance: faster walking was linked to greater swing-phase acceleration, reduced variability during the stance phase, and more regular, predictable movement patterns across multiple gait components. These findings reinforce preferred walking speed as a clinically meaningful indicator of gait performance and neuromuscular control in transtibial amputees. Overall, prosthesis type alone did not appear to be the dominant determinant of neuromuscular control during steady-state walking in this cohort. Given the variability in prosthetic components and user-related factors, future studies should include larger samples, more standardized prosthetic configurations, and richer clinical information, and may benefit from multi-class and clustering approaches to better understand subgroups of users and prosthetic function.

Supplementary Materials: The following supporting information can be downloaded at: https://www.mdpi.com/article/10.3390/prosthesis8020021/s1, Supplementary Video S1: Principal movements.

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Informed Consent Statement: Patient consent was waived, as no additional participants were recruited and no new data were collected for this study. The analysis is based entirely on a publicly available dataset, for which informed consent had already been obtained by the original research team.

Data Availability Statement: The dataset used in this study is openly available from Samala et al. (2024) [26], Scientific Data, DOI: 10.1038/s41597-024-03677-3.

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