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Biomechanical Effects of Passive Exosuit Assistance on Tibiofemoral Loading and Dynamic Stability during Downhill Walking

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Abstract

Background: Downhill walking demands significant eccentric muscle activity at the knee, which increases tibiofemoral joint contact force (TFCF) and challenges gait stability. However, it remains unclear whether passive exosuits can reduce TFCF during downhill walking without compromising dynamic stability. This study investigated the biomechanical effectiveness of a passive trunk-to-thigh soft exosuit in decreasing TFCF and examining its effects on dynamic stability during downhill walking.

Methods: Fourteen healthy male participants walked on an instrumented treadmill set at a -15° decline at $1.0 \text{ m}\cdot\text{s}^{-1}$ under both unassisted and assisted conditions. Knee joint kinetics, TFCF, and dynamic stability parameters were estimated using a subject-specific full-body musculoskeletal model. Statistical Parametric Mapping (SPM) and paired t-tests were used to analyze continuous waveforms and discrete metrics, respectively.

Results: The exosuit decreased biological stance phase TFCF impulse by $\sim 7\%$ (axial) and $\sim 6\%$ (resultant). Notably, participants adopted shorter, higher-cadence, and wider steps with the exosuit, suggesting that the reduced loading was driven concurrently by both the device's direct support and user-driven gait adaptations. SPM identified a significant decrease in knee joint power generation during mid-stance (23–29% gait cycle (GC)). Additionally, the ML margin of stability (MoS) significantly increased by 7.4%, while the anteroposterior (AP) MoS remained unchanged.

Conclusions: These results indicate that the current passive assistance can reduce cumulative tibiofemoral loading during downhill walking.

Concurrently, the increased lateral margin of stability reflects a more cautious, wide-based gait strategy. These outcomes demonstrate that the device reduced joint loading through an intertwined combination of direct mechanical resistance and the induction of a joint-protective, shorter-step gait strategy. Consequently, passive wearable devices represent a promising approach for joint load management during downhill walking.

Keywords: Downhill walking, passive exosuit, tibiofemoral loading, dynamic stability, joint contact force.

1. Introduction

Walking on uneven terrain is a common part of daily life, but it presents unique biomechanical challenges compared to level walking. Specifically, while both propulsive (positive) and collisional (negative) work occur during level gait, downhill walking alters lower limb mechanics by substantially increasing the requirement to absorb negative energy. This increased energy dissipation is necessary to counteract gravity and safely descend the body's center of mass (COM) [1]. This energy dissipation is mainly managed by the knee joint, where the extensor muscles must contract eccentrically to control flexion during the weight acceptance phase of stance [2]. While effective at regulating descent speed, this mechanism generates substantial internal muscle forces, increasing tibiofemoral joint contact force (TFCF) to levels well above those during level walking [2, 3]. These repetitive, high-magnitude compressive loads are closely associated with cartilage degeneration and increased risk of knee osteoarthritis (OA) progression in individuals with existing pathology [4, 5].

Beyond the challenge of joint loading, downhill walking places significant demands on postural control [6]. The destabilizing force of gravity pulls the COM forward and downward, increasing the risk of slips and falls [7]. To maintain stability, individuals often adopt conservative or cautious gait strategies, such as reducing step length and increasing step width [6, 8] or increasing muscle co-contraction [3]. While muscle co-contraction has been proposed as a neuromuscular strategy to increase knee joint stiffness and enhance joint stability during walking [9], it may paradoxically increase compressive load on the tibiofemoral joint and accelerate OA progression [3], creating a complex trade-off between gait safety and long-term joint health. Consequently, there is a crucial clinical and engineering need for interventions that can both lower tibiofemoral joint loading and support gait stability during descent.

Given these biomechanical demands, passive exosuits may provide a solution for assisting downhill walking. Unlike powered devices, passive exosuits depend on springs, dampers, or other flexible components to store or release energy generated by the user's movement. This makes them

especially suitable for downhill gait, in which the lower limbs, particularly the knee extensors, must absorb significant negative energy to counteract gravity [1]. By resisting hip flexion during early stance, a passive exosuit has the potential to decrease eccentric muscle effort and reduce TFCFs without requiring motors or batteries [10]. Essentially, the device functions as a biomechanical brake. It leverages the same gravitational forces that normally challenge stability and joint health to provide low-barrier, energy-efficient support during descent.

Previous research using passive exosuits and exoskeletons during downhill walking has shown promising results. Several studies have demonstrated that devices directly crossing the knee joint can use elastic elements or dampers to dissipate negative joint power, thereby reducing TFCF and quadriceps activation during descent [11-14]. While effective, devices that directly actuate the knee are often bulky or mechanically restrictive. Alternatively, passive assistance can be applied proximally, without crossing the knee, to indirectly offload the joint through dynamic coupling and mechanical energy redistribution [15]. Because the lower limb acts as a coupled kinetic chain during downhill walking, absorbing a portion of the body's kinetic energy at the hip or trunk decreases the total biological braking requirement, meaning the knee does not have to dissipate as much energy [1, 2]. Supporting this concept, a recent pilot study examined various passive exosuit configurations and found that connecting the trunk to the posterior thigh with an elastic exoband produced the greatest reduction in TFCF (-8.8%) [10]. However, this prior pilot study was limited to preliminary peak force assessments.

Building on this knowledge, the objective of the current study was to examine the biomechanical effects of a passive trunk-to-thigh elastic exosuit on downhill walking. Specifically, we investigated how this exosuit affected knee joint kinetics, kinematics, and overall dynamic stability during downhill walking in healthy individuals. We hypothesized that a passive elastic exosuit connecting the trunk to the posterior thigh would decrease the TFCF during downhill walking. In particular, we expected to observe a reduction in TFCFs with the exosuit compared to unassisted downhill walking, without compromising dynamic stability.

2. Methods

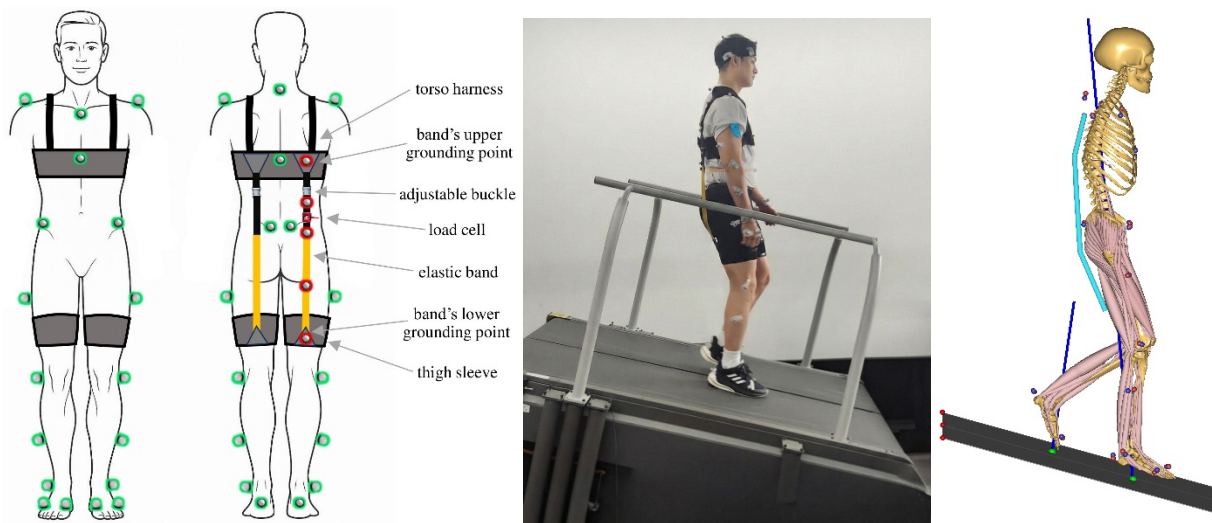


Figure 1. (Left) Illustration of the passive exosuit. The device features a torso harness and two thigh sleeves connected at the back by two passive elastic exobands. These exobands were anchored between the torso harness and the posterior thigh and include an adjustable buckle to modify their length. A load cell was integrated in-line on the right-side exoband to measure tension during gait. Reflective markers for anatomical landmarks are shown in green, while those on the exoband are shown in red. **(Middle)** A participant wearing the passive exosuit during downhill walking. **(Right)** Musculoskeletal model with the exoband represented as a string anchored to the trunk and thigh, corresponding to the upper and lower grounding points marked by the exoband markers.

2.1. Exosuit design

The exosuit was fabricated using textiles and passive elastic bands, as shown in Fig. 1. The mechanical properties of the exosuit's elastic element, including stiffness and force-elongation behavior, were reported elsewhere [16]. To prevent deformation of the suit under tension from the band, a lightweight, high-toughness composite textile (CT5K.18/wov4, Dyneema, USA) was chosen as the primary material. The exosuit consisted of five components: a torso harness with shoulder straps, two thigh sleeves, and two exobands connecting the torso harness to the thigh sleeves. The torso harness and thigh sleeves were adjustable to fit a range of body shapes and sizes. Additionally, the length of the exobands was modified by a buckle to allow individual customization. A load cell (LSB205, FUTEK Technology, Inc., CA), synchronized with the motion capture system, was integrated into the right-side exoband to measure its tension. The exosuit weighed 209 g. Pretension was defined as the initial tensile force in the exoband when the participant stood upright in the anatomical position, prior to walking. The exoband length was adjusted so that the load cell measured a standing tensile force equal to 5% of the participant's body weight (BW).

Mechanistically, the exobands were anchored to the posterior thigh so that band tension created a hip extension moment (i.e., resisted hip flexion) and countered trunk forward flexion. During downhill walking, trunk forward inclination and hip flexion increase mainly early in stance and again near

push-off, stretching the exobands and generating elastic tension. This passive tension was expected to lessen the biological effort needed by the hip extensors (e.g., gluteus maximus and hamstrings) during periods of hip flexion control. Furthermore, we hypothesized that this elastic resistance to hip flexion during stance would indirectly reduce the biological demand on the knee extensors, thereby reducing tibiofemoral joint contact forces.

2.2. Participants

Fourteen healthy male participants (age: 22.15 ± 1.95 years; height: 1.75 ± 0.06 m; weight: 75.26 ± 8.77 kg) were recruited through an advertisement for this study. All participants reported no musculoskeletal injuries in the past six months. Informed written consent was obtained from all participants, and the Chung-Ang University Institutional Review Board approved the study.

2.3. Experimental protocol

Participants performed downhill walking ($1.0 \text{ m} \cdot \text{s}^{-1}$) trials on a dual-belt instrumented treadmill (ITC-21-20; Bertec, Columbus, OH) at a -15° inclination. This speed was selected based on our previous pilot study [10], as a 15° decline is steep and naturally necessitates a slower, more cautious pace for safety and stability compared to normal level walking speeds ($1.1\text{--}1.3 \text{ m} \cdot \text{s}^{-1}$). To ensure participants were adequately acclimated to the exosuit, and to minimize transient learning effects during data collection, a structured training protocol was implemented across three sessions before data collection. Each session included 5 minutes of unassisted (i.e., without the exosuit) downhill walking, followed by 5 minutes of walking with the exosuit, with a 5-minute rest in between. A 24-hour rest period was given between sessions. During the final data collection, participants completed two 1-minute downhill walking trials (unassisted and assisted) in random order, with a 5-minute rest interval between trials. Given the extensive prior training, 1 minute was deemed sufficient to reach steady-state gait.

Twenty-four reflective markers were attached following the Plug-in Gait model (excluding arms), along with five additional markers on the exoband to identify its origin, insertion, and alignment (see Fig. 1). Marker trajectories in 3D were recorded at 200 Hz using an 8-camera Oqus 7 motion capture system (Qualisys, Gothenburg, Sweden), synchronized with a split-belt treadmill (Bertec Corporation, Columbus, OH) sampling at 1000 Hz for ground reaction forces. The exoband force was measured at 1000 Hz with an LSB205 load cell (FUTEK Advanced Sensor Technology, Inc., Irvine, CA, USA), synchronized with the motion capture system.

2.4. Musculoskeletal modeling

Data were analyzed using the AnyBody Modeling System (version 7.4.3; AnyBody Technology, Aalborg, Denmark) with a full-body musculoskeletal model (excluding the hands) adapted from the AnyBody Managed Model Repository (AMMR) [17]. Marker trajectories were low-pass filtered with a

7 Hz 4th-order Butterworth filter, while ground reaction force data were filtered at 20 Hz. Each model was scaled individually to match the participant's specific anthropometry. Joint kinematics were optimized by minimizing the positional error between experimental and virtual markers on the skeletal model. Joint kinetics were then calculated via inverse dynamics using a third-order muscle recruitment criterion, which aims to minimize the maximum muscle stress [18]. Subsequently, the continuous tibiofemoral joint contact force (TFCF) was computed within AnyBody as the net compressive force transmitted across the joint surface. This was derived from the inverse dynamics solution that combines the optimized muscle forces, passive structural forces, and the joint reaction forces required to satisfy dynamic equilibrium at the knee. All joint moments, joint powers, and TFCFs were normalized to BW to facilitate inter-subject comparisons. All degrees of freedom of the cervical spine were constrained, resulting in the neck motion being rigidly coupled to the torso segment. The exoband was modeled as a string element using the AnyForce function, anchored to the skeleton at the upper and lower grounding points corresponding to the attached markers, as shown in Fig. 1. Load cell data were used as the exoband force after being filtered with a 10 Hz Butterworth low-pass filter. This modeling represents the exoband as an ideal tensile element with fixed skeletal attachment points and does not account for interface compliance, soft-tissue deformation, or strap migration/slip. Therefore, the estimated exoband-induced load transfer should be interpreted as an upper bound of effective assistance.

Following a previously described musculoskeletal modeling approach [16], each trial was processed using two inverse-dynamics runs. In the first run, the exoband force was omitted to compute the total joint moment required to reproduce the measured motion. In the second run, the measured exoband tension was applied while keeping the same kinematics, yielding the biological moment generated by muscles and passive tissues. The exoband moment was then obtained as the difference between runs (exoband moment = total joint moment – biological moment). Accordingly, biological TFCF refers to TFCF generated by the musculoskeletal system, including muscle forces and passive joint structures, excluding any force contribution provided by the exosuit. When exosuit assistance was present, the total modeled TFCF was decomposed into biological and exosuit components. To ensure the model produced physiologically feasible solutions in accordance with recommended validation practices [19], simulated muscle activations and knee joint kinetics (moment and power) were validated against experimental downhill walking data from Lay et al. [20], following established methodologies for verifying predicted muscle recruitment patterns [21, 22]. Full details of this validation framework are available in the Supplementary Material.

2.5. Data analysis

To verify that participants reached and maintained a steady-state gait pattern within the 1-minute collection period, we analyzed 10 consecutive right-leg gait cycles (GCs) from the last 20 seconds of each trial. Steady-state was confirmed by the absence of monotonic trends in step kinematics across this extraction window and by the exceptionally low intra-subject variability in temporal parameters (e.g., the group mean coefficient of variation for stance time was $< 3\%$ for both conditions).

Kinematic and kinetic data were processed using a custom MATLAB® (MathWorks, Natick, MA, USA) script. Gait events were detected using the vertical component of the ground reaction force (vGRF), with heel strike occurring at the first moment when vGRF surpassed 30 N, and toe-off was identified at the first moment when vGRF dropped below 30 N during the stance phase. Data from heel-strike to the subsequent heel-strike of the same leg (0% to 100% of stride time) were time-normalized to 101 data points using cubic spline interpolation. From these events and the marker data, spatiotemporal parameters were calculated using standard definitions for the right leg, including step frequency, step length, step width, and stance time. Step length and step width were defined as the anteroposterior (AP) and mediolateral (ML) distances between the heel markers at consecutive contralateral heel strikes (left-to-right), and stance time was the duration from foot contact to toe-off for that foot. The variability of each parameter was quantified as the standard deviation (SD) across all steps for each participant. Dynamic stability was assessed using MoS in both ML and AP directions, calculated with the extrapolated center of mass (XCoM) method, as described by Hof et al. [23]. Briefly, at each step, the margin of stability (MoS) was defined as the minimum distance between the XCoM (the COM position adjusted forward by its velocity) and the boundary of the base of support in the respective direction. Finally, the TFCF impulse was computed by integrating the absolute value of the contact force over time during the stance phase of downhill walking.

2.6. Statistical analysis

All outcome measures were summarized as the mean $\pm$ SD. Variables were averaged across 10 selected gait cycles per participant to reduce within-subject variability, and these participant-level means were used for statistical analyses. All continuous biomechanical waveforms (e.g., joint angle and moment vs. % gait cycle) were compared between conditions using one-dimensional Statistical Parametric Mapping (SPM) tests. SPM allows us to test for differences across the entire time series without predefining discrete points, identifying any significant temporal regions of difference. Before the statistical testing, all continuous waveforms and discrete peak variables were confirmed to be normally distributed using the Shapiro-Wilk test for each discrete parameter and the built-in 'spm1d.stats.normality.ttest' function for SPM data. All statistical tests

applied a significance level of $\alpha = 0.05$. The discrete analyses were performed using SPSS (version 25; IBM Corp., Chicago, IL). The SPM-based time-series analyses used the open-source SPM1D toolkit (v0.4, www.spm1d.org) in MATLAB (R2024a, The MathWorks).

3. Results

In both unassisted and assisted conditions, toe-off occurred at approximately 68% of the GC. Fig. 2 illustrates the force in the passive exoband during the normalized GC. The exoband remained engaged throughout the GC, with tension fluctuating between approximately 33 and 42 N. The force showed a slight increase in early stance, dipped to a local minimum around mid-stance (45–55%), then increased sharply to reach the peak at toe-off (65–70%). After toe-off, the force decreased slightly but remained present throughout swing, confirming that the band stayed engaged for the entire GC.

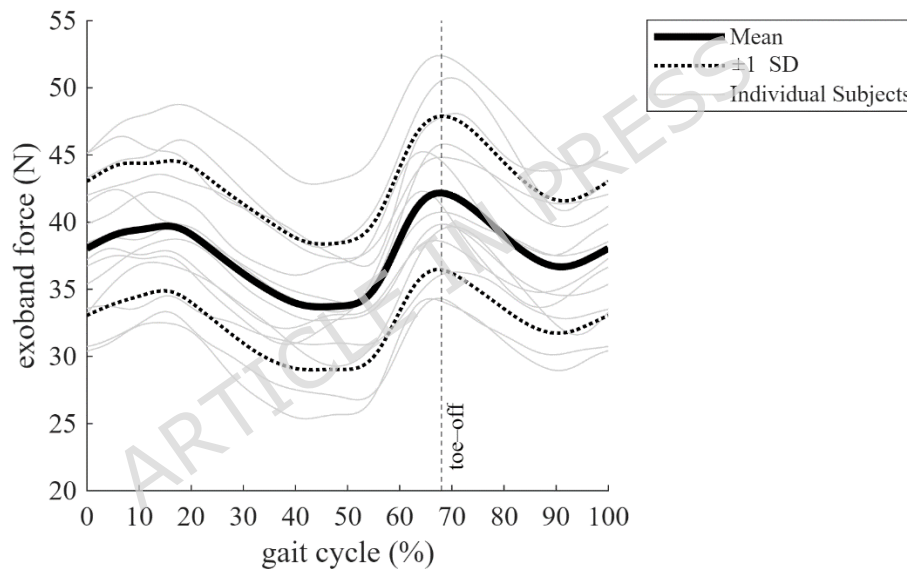


Figure 2. Exoband force (N) during downhill walking ($n=14$). Thick solid line: mean across 14 participants; dotted lines: ± 1 SD; light gray lines: individual participant averages; vertical dashed line: toe-off ($\sim 68\%$ of GC). The exoband remained engaged throughout the GC, with mean tension fluctuating between approximately 33 and 42 N.

Fig. 3 compares knee joint biomechanics between unassisted and assisted downhill walking. Note that the knee moments and powers reported are the biological (muscle-generated) values, as the exosuit does not span the knee and thus adds no direct moment or power to the knee joint. The SPM analysis showed no differences in knee angle or moment plots between unassisted and assisted conditions during the stance phase, as the SPM curves for these variables did not exceed the significance threshold. However, a statistically significant difference in knee power was observed during mid-stance. During 23%–29% of the GC, the assisted condition

exhibited significantly lower knee power than in unassisted walking ($p < 0.05$).

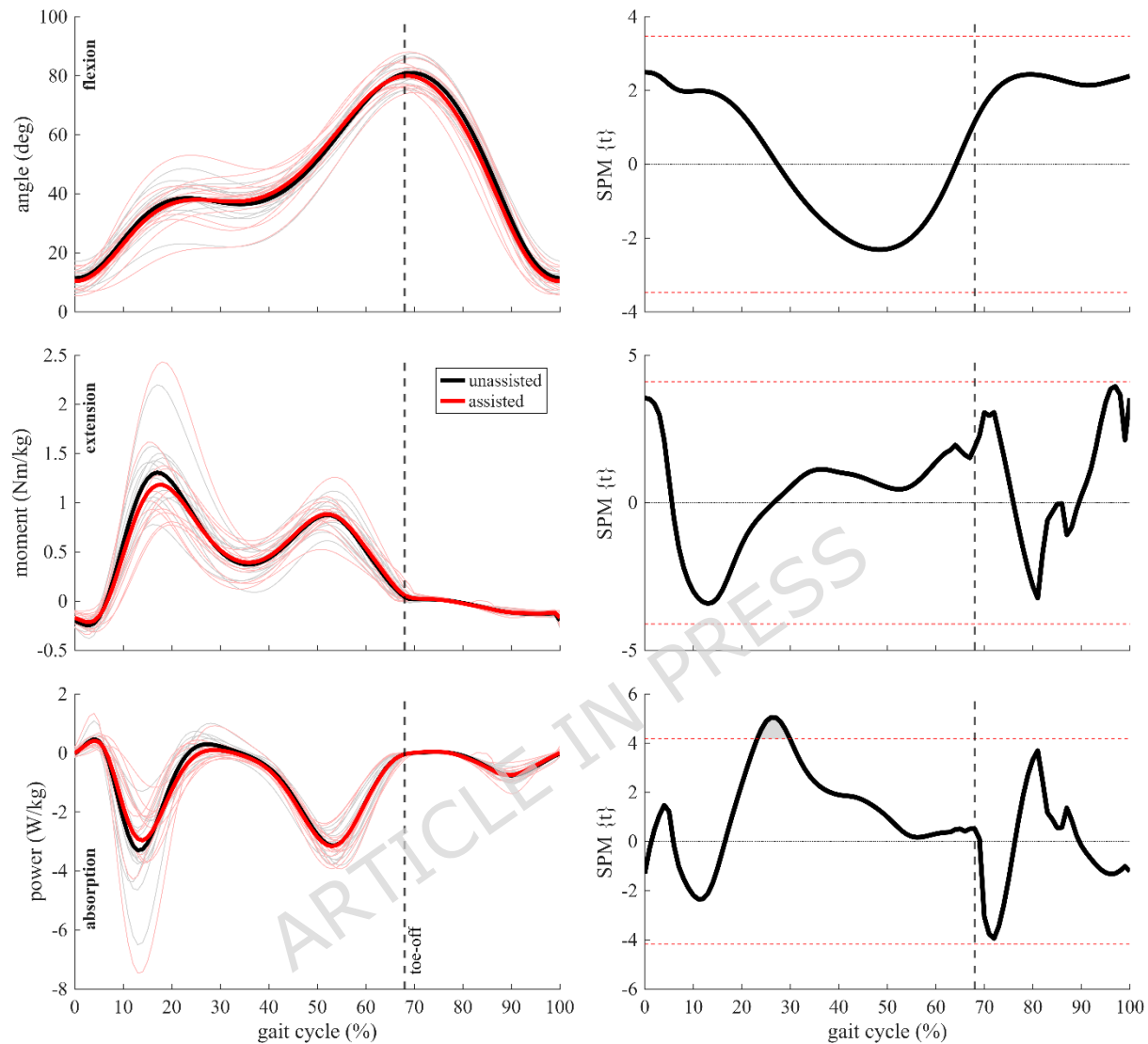


Figure 3. Knee joint biomechanics during downhill walking under unassisted and assisted conditions ($n=14$). (Left) Group means (thick lines) and individual subject trajectories (thin lines) of knee joint angle, biological knee moment, and biological knee power over normalized GC for unassisted (black line) and assisted (red line) downhill walking. The vertical dashed line indicates toe-off ($\sim 68\%$ GC). (Right) SPM results from 1D paired t-tests comparing the two conditions for each metric. No significant differences were observed in knee angle or knee moment during stance, as the SPM curves did not pass the threshold. A significant difference in knee power was found between 23–29% GC ($p < 0.05$), highlighted in gray.

Table 1. (mean $\pm$ SD) Spatiotemporal and dynamic stability parameters for downhill walking during unassisted and assisted conditions ($n=14$).

Variable	unassisted	assisted	p-value
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step frequency (steps/min)	122.30 ± 4.30	126.46 ± 6.25	0.020 *
step length (cm)	49.50 ± 2.97	47.93 ± 2.89	0.036 *
step width (cm)	20.24 ± 3.05	21.80 ± 3.72	0.004 *
stance time (s)	0.66 ± 0.02	0.64 ± 0.03	0.011 *
step length variability (cm)	3.21 ± 1.00	3.17 ± 0.97	0.908
step width variability (cm)	1.23 ± 0.39	1.23 ± 0.48	0.997
stance time variability (s)	0.015 ± 0.004	0.016 ± 0.005	0.598
mediolateral margin of stability (cm)	9.42 ± 0.79	10.10 ± 0.95	0.005 *
anteroposterior margin of stability(cm)	25.13 ± 2.94	23.26 ± 3.30	0.057

* indicates significant comparisons (paired-sample t-test, $p < 0.05$).

As shown in Table 1, exosuit assistance significantly increased step frequency and reduced step length and stance time during assisted downhill walking compared to the unassisted condition. Step width was significantly wider with exosuit assistance. Variability in step length, step width, and stance time was similar across conditions. Additionally, ML MoS was significantly larger with exosuit assistance, while AP MoS showed no difference.

During downhill walking, the biological TFCF profiles appeared similar between conditions (Fig. 4, left column), and SPM (paired 1D t-tests of the biological forces) detected no significant between-condition differences in the biological TFCF waveforms across stance (Fig. 4, right column). Additionally, the exosuit contribution was greatest in the axial direction during stance (Fig. 4, middle column).

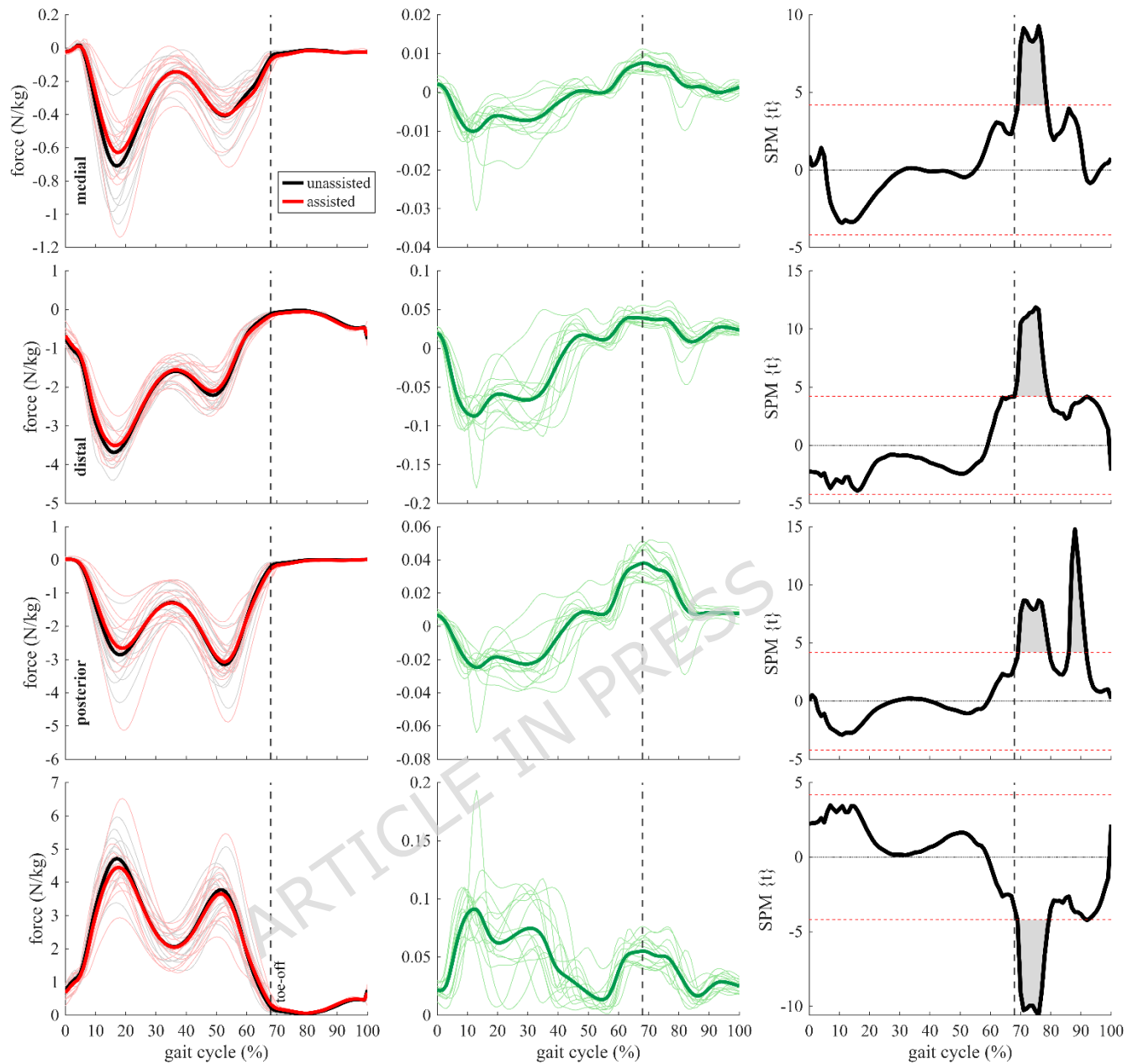


Figure 4. Tibiofemoral joint contact forces during downhill walking in unassisted (black lines) and assisted (red lines) conditions (n=14). Four rows correspond to the ML, axial, AP, and resultant force components, respectively. Each row's left panel shows the group means (thick lines) and individual subject trajectories (thin lines) of time-normalized biological contribution, the middle panel displays the exosuit contribution, and the right panel presents the SPM results from 1D paired t-tests comparing biological contributions between conditions. The vertical dashed line marks toe-off (~68% of GC). No differences in biological contributions were found with SPM during the stance phase. Notably, the exosuit exerted its greatest assistive force in the axial direction.

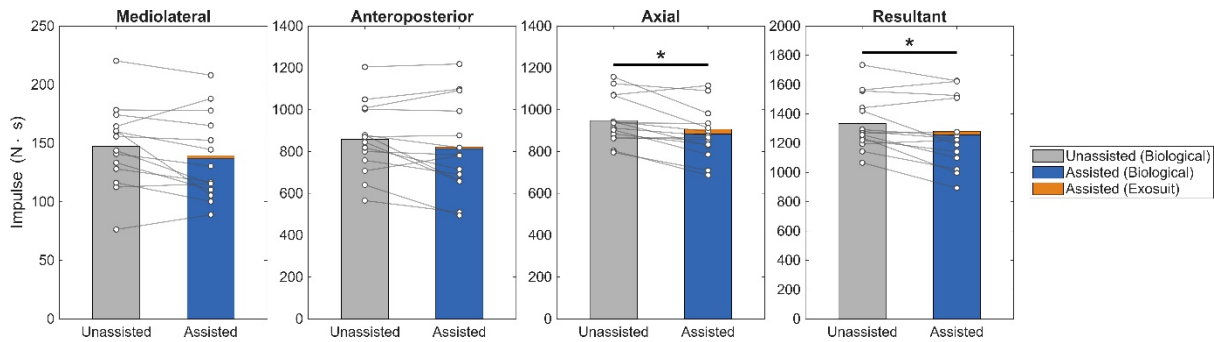


Figure 5. (mean $\pm$ SD) Tibiofemoral joint contact force impulses (N·s) during the stance phase of downhill walking for unassisted and assisted conditions ($n = 14$). Bar charts display the impulse for the mediolateral, anteroposterior, axial, and resultant forces. Assisted condition bars divided into biological (blue) and exosuit (orange) contributions. Gray lines represent individual within-subject responses, connecting the unassisted condition to the biological component of the assisted condition. Asterisks indicate a significant decrease in the biological impulse for the axial and resultant forces while wearing the exosuit (paired t-tests, $p < 0.05$). Detailed values are provided in Supplementary Table S1.

During downhill walking, the stance phase TFCF impulse, computed from the biological force contribution, was significantly lower with the exosuit compared to the unassisted condition (Fig. 5). In paired two-tailed t-tests comparing the biological contributions of unassisted and assisted conditions, wearing the exosuit resulted in a lower axial impulse and a lower resultant impulse (both $p < 0.01$). Specifically, the biological axial impulse decreased by $\sim 7\%$ with the exosuit (from 947.9 ± 110.2 N·s in unassisted condition to 884.3 ± 118.0 N·s in assisted condition; see supplementary table S1), and the biological resultant impulse by $\sim 6\%$ (from 1334.3 ± 178.3 N·s to 1257.4 ± 224.3 N·s). In contrast, the biological ML and AP impulse components showed no significant differences between unassisted and assisted conditions. These findings indicate that wearing the exosuit resulted in significant reductions in axial and resultant tibiofemoral joint loading during stance, without notable changes in ML or AP loading.

Analysis of within-subject responses indicated that the overall reductions in impulse were generally consistent across the cohort, although individual variations were observed (Figure 5). Specifically, for the axial impulse, 12 of the 14 participants demonstrated a reduction in the biological component during the assisted condition, whereas two participants exhibited an increase. Similarly, 11 of the 14 participants showed a reduction in the resultant impulse. Furthermore, exploratory correlation analysis between the change in step length and the change in axial TFCF impulse yielded a strong positive correlation ($r = 0.83$, $p < 0.001$). This indicates that approximately 69.5% of the variance in individual impulse reduction was shared with modifications in step length.

4. Discussion

The present study aimed to evaluate the biomechanical effectiveness of a passive trunk-to-thigh soft exosuit on tibiofemoral joint loading and dynamic gait stability during downhill walking. Our results showed that the exosuit effectively decreased the cumulative load (impulse) on the tibiofemoral joint during the stance phase by about 7% in the axial direction and 6% in the resultant TFCF. Additionally, the exosuit significantly reduced positive knee joint power during mid-stance and resulted in a more cautious gait pattern, characterized by higher step frequency, shorter step length, and wider step width. Furthermore, ML MoS increased significantly, whereas AP MoS remained unchanged.

The significant decrease in TFCF impulse underscores the device's potential for long-term joint protection. While peak joint loads are critical markers of acute mechanical stress, research on OA has progressively broadened to include the "cumulative load" hypothesis. Long-term studies indicate that the cumulative impulse of joint loading serves as a highly robust predictor of cartilage volume loss, radiographic progression, and symptom worsening over time [4, 5]. By reducing total impulse by approximately 7%, the passive exosuit reduced mechanical stress on the tibiofemoral joint during downhill walking. A potential confounding factor is that the assisted condition resulted in a modest ~3% reduction in step length, which could account for part of the ~7% decrease in TFCF impulse. Shortening step length has been shown to directly reduce knee joint loading. For example, Milner et al. [24] found that a 15% decrease in step length significantly reduced cumulative TFCF impulse in adults with and without obesity (12-19% and ~28% respectively). Daniel Syrett et al. [25] reported a ~16% reduction in medial knee contact force impulse with a 15% shorter step length. Based on these findings, the ~3% shorter steps observed here are likely the primary driver of the ~7% reduction in impulse under assistance. Our correlation analysis strongly supports this, demonstrating that 69.5% of the variance in impulse reduction was directly shared with step length changes ($r = 0.83$). The remaining 30.5% of unexplained variance likely reflects a complex combination of the exosuit's direct elastic offloading, altered trunk kinematics, shifts in the ground reaction force vector, and inherent neuromuscular adaptations. Because our study design did not utilize controlled-kinematic conditions (e.g., walking to a metronome or fixed foot-placement targets), the true isolated contribution of the elastic element cannot be definitively quantified, representing a limitation of the current protocol. Nevertheless, this data indicated that the device's greatest benefit for cumulative joint protection stems from its ability to continuously and safely induce a protective, shorter-step gait strategy.

While the direct translation from a ~7% per-step reduction in TFCF impulse to clinically meaningful protection against OA progression cannot be definitively established from the current acute study, this magnitude of load reduction is highly comparable to that achieved by established conservative

clinical interventions (e.g., lateral wedge insoles and knee braces, which typically yield a 4–14% reduction in medial compartment load [26]). Therefore, while this magnitude of offloading may not completely arrest advanced OA progression on its own [4, 5], it represents a highly clinically relevant buffer against acute cartilage overloading and cumulative joint fatigue for healthy individuals and those with early-stage joint degeneration.

Notably, the assisted condition exhibited a significant reduction in knee joint power generation during 23–29% of the gait cycle (Fig. 3), indicating a smaller positive power burst at mid-stance. This phase corresponds to late mid-stance (~28% GC), when the quadriceps act concentrically to initiate knee extension and generate a positive power burst. The reduced power generation in the assisted condition indicates a lower muscular demand for active energy generation at this phase. Because the exoband does not span the knee joint, this change likely reflects an indirect effect of exosuit assistance on knee energetics rather than a direct knee-level power contribution.

The mechanism by which proximal hip-extension assistance reduces distal knee kinetics cannot be resolved from the present data alone. The exoband remained continuously engaged and provided sustained tension across the gait cycle (Fig. 2), meaning the device generated a persistent hip extension moment throughout stance, a condition that would plausibly reduce the demand on the knee extensors to resist forward trunk progression. The observed reduction in mid-stance knee power generation (23–29% GC) is consistent with reduced quadriceps demand at this phase. However, the lack of any significant difference in the knee moment and power waveforms during early stance (SPM) indicates that the primary energy dissipation task at the knee was largely unchanged. This pattern suggests the exosuit modestly reduced the active knee extensor contribution during mid-stance without altering the dominant eccentric braking strategy that defines downhill walking. Whether this reflects a redistribution of energy across joints, a reduction in co-contraction, or a shift in load-sharing between passive and active structures cannot be determined without hip joint power analysis and trunk kinematics, neither of which were considered in the present study. Future work incorporating multi-joint power analysis and segment kinematics is required to identify the operative load-transfer pathway.

Despite the current mechanistic ambiguity, the observed reduction in mid-stance knee power indicates that lightweight, proximal hip assistance successfully reduces quadriceps effort. This achieves an offloading effect similar to that of prior passive and quasi-passive devices designed for downhill or descent-like tasks [13, 14], notably accomplishing this without the need to directly actuate or restrict the knee joint.

In parallel with these joint-level effects, the exosuit induced several spatiotemporal gait adaptations. Participants adopted a gait pattern characterized by a higher step frequency (+3.4%), a shorter step length (−3.0%), and a wider step width (+7.7%). In biomechanical literature, this pattern is often referred to as cautious gait and is commonly seen among older adults and individuals walking downhill to reduce slip risk [6, 8]. The exosuit likely promoted this gait strategy via both mechanical and behavioral pathways. First, we speculate that the tension of the exobands resists hip flexion, which may discourage long steps. Second, users may intuitively shorten their steps and increase cadence to synchronize with the exosuit's assistive force on each step. Importantly, this modified gait is biomechanically beneficial for the knee, since reducing step length consistently lowers tibiofemoral loads or contact forces during walking [24, 25]. Therefore, beyond its direct support, the device encouraged a protective movement pattern to complement its mechanical assistance.

Dynamic stability is crucial for passive assistive devices because added stiffness or constraints to the joints can potentially hinder compensatory balance reactions [27]. In this study, the ML MoS increased by 7.4% with the exosuit, whereas AP MoS remained unchanged. However, it is important to note that a larger MoS does not necessarily equate to enhanced intrinsic dynamic stability; rather, it often reflects a compensatory response [28]. This increase in ML MoS is consistent with the 7.7% wider step width observed in the assisted condition. Therefore, rather than interpreting this as an absolute improvement in global stability, it is highly likely that participants adopted a more conservative, cautious gait strategy in the frontal plane. This adaptation likely serves to manage perceived instability while adjusting to the novel assistive forces and the added mass of the exosuit. Notably, the variability of step length, step width, and stance time was unchanged between conditions (Table 1), suggesting that exosuit wear did not increase gait variability during steady-state downhill walking. Together, these results indicate that the exosuit altered the mean gait strategy (wider steps) without measurably degrading spatiotemporal consistency. Moreover, unlike rigid exoskeletons that may interfere with interlimb coordination and reduce dynamic stability during walking [27], the lightweight textile design of this exosuit imposes minimal mechanical constraints, thereby likely preserving natural balance control.

5. Limitations

A key limitation is that the exosuit appears to reduce TFCF through an intertwined combination of direct mechanical offloading by the elastic element and implicit gait modifications during exosuit-assisted walking. Given the nature of our data collection and study design, it is not possible to isolate the independent contributions of each pathway. Specifically, the mechanical assistance and the associated kinematic changes occurred

simultaneously under the same experimental conditions. We did not include an experimental manipulation (such as controlled step length/kinematic constraints, matched-speed-matched-kinematics trials, or graded stiffness/engagement settings) or a modeling approach that would enable us to decompose TFCF changes into device-driven versus behavior-driven factors. Therefore, the reported reductions in peak and impulse TFCF should be interpreted as the overall effect of wearing the exosuit, rather than as evidence that either direct offloading or gait adaptation alone is sufficient to explain the observed changes.

These findings reflect acute (i.e., within-session) responses during steady-state treadmill downhill walking after adequate familiarization in healthy young males and should not be interpreted as evidence of long-term clinical benefit or generalizability to older adults or individuals with existing knee conditions (e.g., knee OA). Although we observed decreases in cumulative load, the long-term clinical effectiveness of this impulse reduction in preventing cartilage degeneration remains to be investigated.

A methodological limitation of this study is the absence of experimental EMG data to validate subject-specific muscle excitations. Furthermore, while we successfully validated the temporal phasing of our model against published in vivo data, the lack of a perfectly matched reference dataset (-15° decline at 1.0 m/s) precluded direct validation of absolute TFCF magnitudes. Consequently, while relative within-subject comparisons remain robust, the absolute TFCF values should be interpreted cautiously. Future studies incorporating simultaneous EMG collection and matched boundary conditions are needed to strengthen the fidelity of these estimates further.

6. Conclusion

This study demonstrated that a passive trunk-to-thigh exosuit altered biomechanics during downhill walking, reducing cumulative tibiofemoral load ($\sim 7\%$ reduction in stance-phase impulse) and positive knee power, and inducing a more cautious gait pattern accompanied by increased ML stability ($\sim 7.4\%$ increase in ML MoS). The passive exosuit achieved its protective effect through an intertwined combination of direct mechanical resistance, evidenced by reduced knee joint power generation, and induced spatiotemporal gait changes (shorter, higher-cadence, wider steps). However, the present study cannot isolate the independent contribution of each pathway. Together, these findings highlight the potential of lightweight, passive wearable technologies as accessible interventions for managing tibiofemoral contact loading during downhill walking.

Abbreviations

AMMR	AnyBody Managed Model Repository
AP	anteroposterior
BW	body weight
COM	Center of Mass
GC	Gait Cycle
HAT	Head-Arms-Trunk
ML	mediolateral
MoS	Margin of Stability
OA	Osteoarthritis
SD	standard deviation
SPM	Statistical Parametric Mapping
TFCF	Tibiofemoral contact force
vGRF	vertical ground reaction force
XCoM	Extrapolated Center of Mass

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Authors' contributions

AN and WK contributed to conceptualization, data curation, methodology, formal analysis, visualization, software, writing – original draft, and writing – review & editing. JY and HGD contributed to conceptualization, methodology, formal analysis, visualization, writing – original draft, and writing – review & editing. GL contributed to conceptualization, data curation, methodology, formal analysis, visualization, writing – original draft, writing – review & editing, funding acquisition, supervision, and project administration.

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Data availability

The datasets used and/or analyzed during the current study are available from the first author on reasonable request.

Declarations

Ethics approval and consent to participate

The experimental design and protocol were approved by the Chung-Ang University Institutional Review Board (1041078-202107-HR-214-0). All participants provided informed consent before the experiments.

Consent for publication

All authors consent to the publication of this manuscript.

Competing interests

GL is an employee of HUROTICS Inc., which has a commercial interest in the passive exosuit evaluated in this study. However, the authors conducted the research and analysis entirely independently. HUROTICS Inc. did not provide specific funding for this trial and had no role in the study design, data collection, data analysis, interpretation of results, or writing of the manuscript; nor did the company review or approve the manuscript before submission. The findings and conclusions presented in this paper are solely those of the authors and do not necessarily reflect the views or positions of HUROTICS Inc.

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