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Multi-Plane Biomechanical Analysis of Lower Limb Joints: Implications for Rehabilitation and Fall Prevention

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Abstract This study examines the biomechanical responses of the ankle, knee, and hip joints during walking on varying slopes to understand how different inclinations affect joint loading and movement mechanics. While previous research has explored slope walking, many studies lack detailed multi-plane analyses of joint moments and accelerations, limiting their applicability in rehabilitation and injury prevention. To address these gaps, we employed advanced motion capture and force plate measurements to quantify joint moments and accelerations at inclinations of 0°, 5°, 7.5°, and 10°. Our results indicate that steeper slopes significantly increase joint moments and accelerations, particularly in the knee and hip during incline walking and in the ankle during decline walking. These findings highlight the increased biomechanical demands on lower limb joints, emphasizing the need for tailored rehabilitation programs, training strategies, and ergonomic interventions. By providing a more comprehensive understanding of slope-related mechanical stresses, this study contributes valuable insights for injury prevention, rehabilitation, and performance optimization in both clinical and athletic settings. The findings suggest that to decrease the risk of falling and manage the demands of inclined walking, appropriate walking strategies and improved safety measures should be implemented, especially during decline and anterior-posterior orientations. This study also offers additional understanding of optimal incline walking techniques for secure and practical locomotion.

Keywords: Slope walking, Gait analysis, Motion capture, Locomotion mechanics, Ergonomic interventions.

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Introduction

Balance is one of the fundamental capabilities of human movement, essential for stability and fall prevention during various locomotor tasks. Maintaining balance requires that the vertical projection of the body's center of mass (COM) remains within the base of support [1]. When this condition is compromised, individuals are at a higher risk of losing balance, which can lead to falls and potentially severe injuries [2]. Walking on inclined or declined surfaces poses a significant challenge to the postural control system, as it disrupts stability and places increased demands on musculoskeletal coordination [3,4]. Moreover, gait inherently involves a dynamic state of imbalance, further emphasizing the importance of effective balance control during locomotion [5,6].

Globally, falls are a critical public health concern, responsible for approximately 684,000 deaths annually, making them the second leading cause of unintentional injury-related fatalities [1]. These outcomes are especially pronounced in low- and middle-income regions, such as the Western Pacific and South-East Asia, which account for around 60% of global fall-related deaths [2,3]. Locomotor difficulties particularly when walking on uneven surfaces, stairs, or slopes—increase the risk of falls and challenge both balance and joint stability [4]. Among these, slope walking is especially hazardous; it is less stable than stair ascent and imposes considerable joint loading, particularly during downhill locomotion, which may lead to pain or injury over time.

A broad body of research has investigated gait stability under various conditions, including age-related changes [7,8], methodological approaches [9–11], walking strategies [12–16], pathological gait patterns [17–20], and surface characteristics [12,21–23]. Studies such as those by Wang *et al*. [24] and Blair *et al*. [25] have explored stability on irregular terrains and at varying walking speeds. Similarly, Schut *et al*. [26] examined dynamic stability and loading during outdoor gait. Despite these efforts, the understanding of postural control and stability during slope walking remains limited, especially in publicly accessible environments.

Many existing studies focus exclusively on kinematic parameters or employ treadmills to simulate incline walking, which may not accurately reflect the mechanics of overground locomotion. For instance, Vierra *et al*. [25] analyzed postural stability on inclined surfaces using only kinematic data, omitting kinetic insights that are crucial for understanding joint loading. Park *et al*. [26] noted that treadmill-based incline walking may differ biomechanically from overground walking, suggesting that custom-built incline platforms offer a more ecologically valid alternative. Furthermore, studies assessing COM trajectories often rely on full-body marker sets [27,28], which are expensive and time-consuming to implement. Thus, there is a pressing need for comprehensive, accessible methodologies that integrate both kinematic and kinetic assessments.

Notably, little is known about how specific slope angles influence dynamic stability and joint loading during incline and decline walking. Most previous research has emphasized energy expenditure or generalized gait adaptations without providing a detailed, multi-plane analysis of joint mechanics [29–32]. This study addresses these gaps by evaluating the biomechanical responses of the ankle, knee, and hip joints during walking on slopes of 0°, 5°, 7.5°, and 10°, using advanced motion capture and force plate technologies. By quantifying joint moments and accelerations across multiple planes, we aim to elucidate how varying slope gradients affect joint loading, balance control, and injury risk.

The outcomes of this research offer data-driven insights for optimizing rehabilitation protocols, developing safer and more effective training strategies, and improving the design of assistive devices. Ultimately, this study contributes to a more comprehensive understanding of slope-induced mechanical demands, with implications for both clinical applications and performance optimization in real-world walking environments.

Materials and Methods

Participants

Twenty healthy volunteers (mean age: 24 ± 1.2 years) with a normal body mass index (BMI) participated in this study. To minimize variability due to mobility impairments, individuals with known musculoskeletal injuries, orthopedic conditions, or difficulties performing walking tasks were excluded. All participants provided written informed consent prior to participation. The study protocol was reviewed and approved by the Ethics Committee of Universiti Malaysia Perlis (UniMAP), ensuring compliance with ethical standards for human research.

All participants were given time to familiarize themselves with a custom-designed inclined wooden platform, shown in Figure 1. The platform measured 5 meters in length and 1 meter in width, with an adjustable inclination that could be set to 0°, 5°, 7.5°, or 10°. The selection of the three specific slope angles (5°, 7.5°, and 10°) was driven by architectural relevance and environmental thresholds. The 5° slope represents the maximum allowable incline for accessible ramps under Malaysian Standard MS 1184:2014 and the Uniform Building By-Laws (UBBL), as well as international guidelines like the Americans with Disabilities Act (ADA). The 7.5° and 10° angles were selected to represent steeper, non-standard gradients often encountered in natural outdoor terrains or non-compliant environments, allowing for a systematic assessment of biomechanical adaptations across both standard accessible and challenging surfaces. The structure was specifically engineered to allow two embedded force plates (Bertec Corporation, Columbus, OH, USA), each measuring 60 × 40 × 10 cm, to lie flush with the walking surface for accurate force measurements. Ground reaction forces (GRFs) were recorded at a sampling frequency of 200 Hz. Simultaneously, a three-dimensional (3D) motion capture system composed of five Oqus cameras (Qualisys, Gothenburg, Sweden) captured dynamic and static movement data at the same frequency (200 Hz).

It is acknowledged that the 5 m walkway distance utilized in this study is relatively short, which may limit the observation of fully established steady-state gait over a prolonged duration. This distance was necessary due to laboratory spatial constraints and the engineering requirements of the custom-built adjustable platform to safely house the embedded force plates. To mitigate this limitation, participants were given ample familiarization trials prior to data collection, and kinetic extractions were focused on the central steps to best approximate steady-state locomotion.

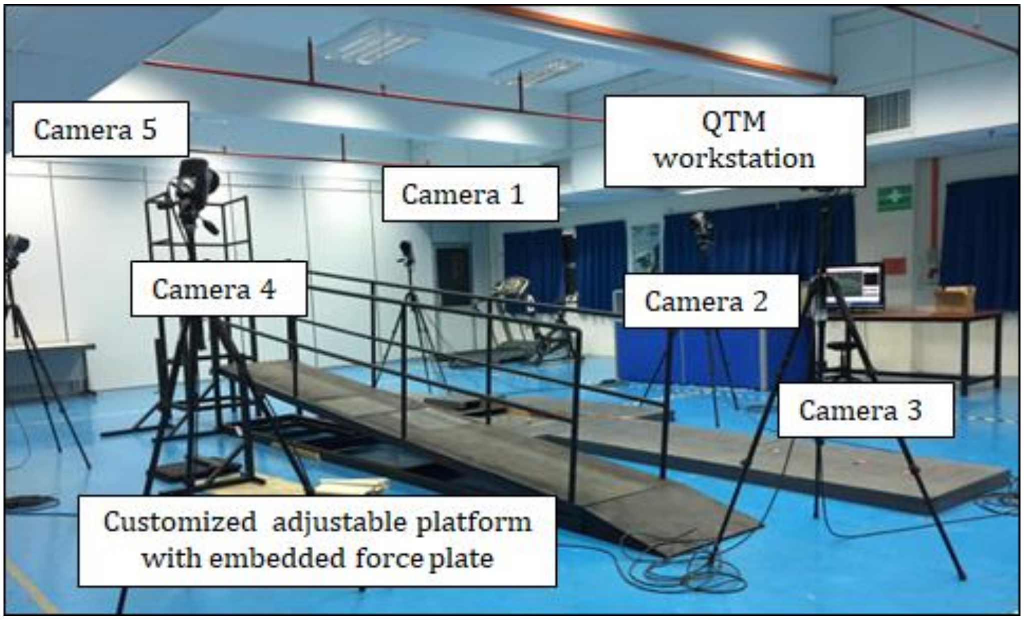

Figure 1. Experimental setup for slope walking analysis, showing the inclined platform with embedded force plates and the motion capture system with five cameras positioned to capture three-dimensional gait data

Marker Placement and Data Acquisition

Reflective markers with a diameter of 20 mm were placed on anatomical landmarks of the right lower limb, following a modified version of the "Plug-in Gait" lower limb marker set. The marker placements are illustrated in Figure 2, showing both anterior (Figure 2a) and posterior (Figure 2b) views. A total of ten reflective markers were affixed directly to the skin or modified footwear to ensure unobstructed visibility of the foot markers. Each subject completed a static calibration trial prior to beginning the walking trials.

Participants were instructed to walk at a self-selected, natural pace along the platform under level (0°) and inclined conditions (5°, 7.5°, and 10°), in both uphill and downhill directions. All walking trials were conducted while the subjects wore minimal shoes to preserve natural foot mechanics. The modified shoes were adapted to allow direct placement of markers on the foot, rather than on the shoe surface.

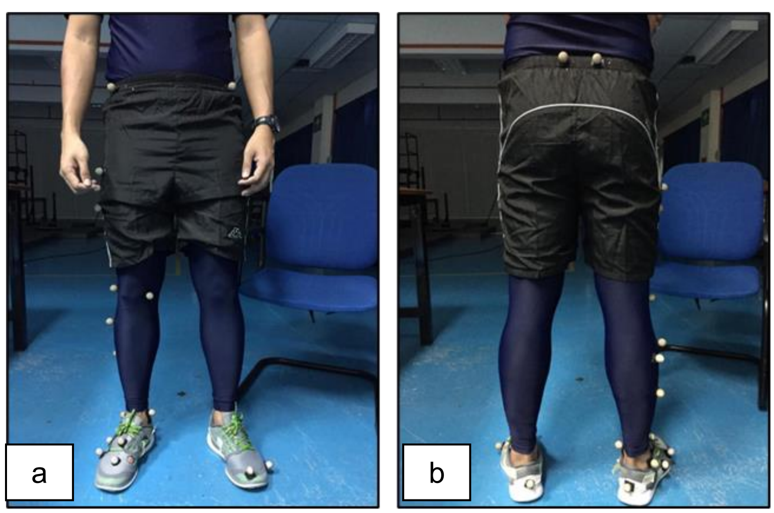

Figure 2. Marker placement on the subject's body (a) front view and (b) rare view

Data Processing

Marker trajectory data captured by the Qualisys motion capture system were initially processed using Visual3D software (C-Motion Inc., Germantown, MD, USA) to construct biomechanical models and compute joint kinematic and kinetic parameters. The dynamic and static trials were exported in .3cd format for further analysis. A fourth-order low-pass Butterworth filter with a cut-off frequency of 6 Hz was applied to the marker and force plate data to minimize high-frequency noise and signal artifacts.

A hybrid Visual3D model was constructed for each subject, comprising four lower limb segments: pelvis, thigh, shank, and foot. Each subject's anthropometric data (height and weight) were entered into the model to scale segment lengths and normalize joint kinetics to body mass. The joint reaction forces (JRFs) at the hip, knee, and ankle were calculated relative to the anatomical coordinate system planes, providing insight into intersegmental loading during slope and level walking.
The following parameters were extracted for analysis:

- Ground Reaction Force (GRF)
- Joint Reaction Forces (JRFs) of the hip, knee, and ankle
- Joint Moments
- Joint Accelerations

All computed parameters were exported from Visual3D and further analyzed in MATLAB (2019) for custom calculations, including the refinement of joint accelerations and kinetic measures.

Statistical analysis was performed using the Statistical Package for the Social Sciences (SPSS) version 17.0 (IBM Corp., Armonk, NY, USA). Normality of the data was assessed using the Shapiro-Wilk test. Due to non-normal distribution, non-parametric Kruskal-Wallis tests were employed to determine the significance of differences in joint moments and accelerations between level and slope walking conditions. Statistical significance was established at $P < 0.05$.

Results

Joint Moment on Ankle, Knee and Hip during Slope Walking

Figure 3 displays six graphs showing the joint moments of the (a) ankle, (b) knee, and (c) hip in the sagittal (X) plane under both declining and inclined conditions. Each graph plots the joint moment (Nm) against the percentage of stance phase (%). The graphs were distinguished by different line styles and colors representing four different angles: 10°, 7.5°, 5°, and 0°. The trends in the data indicate variations in the joint moments at different angles and phases of stance. For the ankle joint in Figure 3(a), both inclined and declined walking in the sagittal plane showed an almost similar pattern, with pronounced peaks at 70% to 80% of the stance phase. The frontal plane pattern of the joint moment during inclined walking is almost similar to that during level walking. Instead, during decline walking, a 5° inclination is almost similar to level walking with a lower magnitude, reaching a peak at 70%–80% of the stance phase. The transverse plane for both inclined and declined walking showed almost similar patterns with different variabilities and magnitudes. The 10° inclination showed obvious variability with the highest magnitude for both walking conditions.

Figure 3(b) knee joint demonstrated distinct patterns and magnitudes across all planes, both for incline and decline walking, suggesting that the knee joint responds more variably to changes in slope than the ankle joint does. These observations indicate that the knee may require additional adjustments in joint moments to accommodate the demands of walking on inclined or declined surfaces. Similarly, Figure 3(c) shows that the hip joint displayed unique patterns for all planes during both incline and decline walking at almost all inclinations, showing different patterns with different variabilities and magnitudes. However, during inclined walking, the 5° and 10° inclinations show similar styles, with the 10° inclination having a higher magnitude.

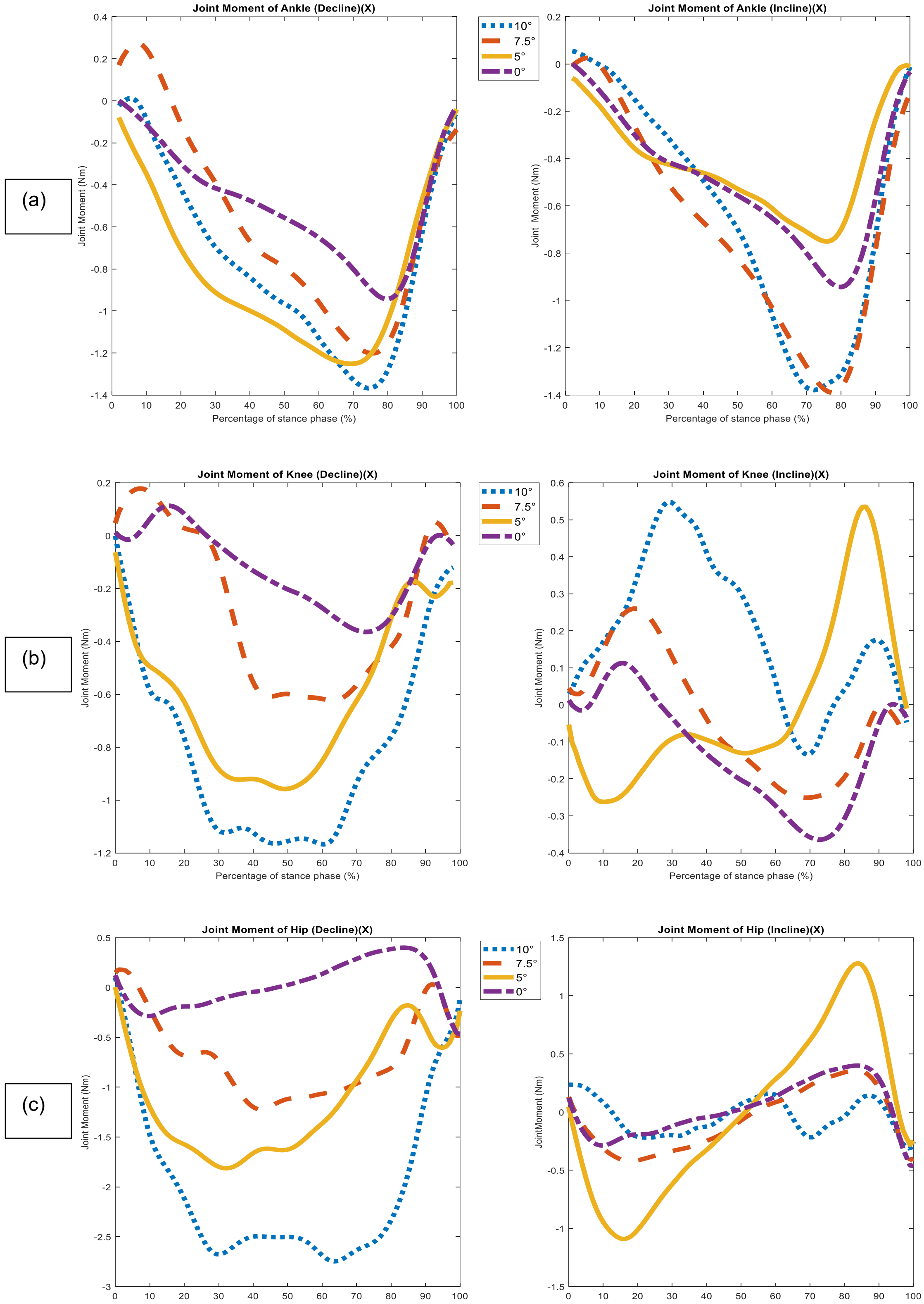

Figure 3. Joint moments of the (a) ankle, (b) knee, and (c) hip in the sagittal (X) plane during decline walking at incline angles of 0°, 5°, 7.5°, and 10°

During inclined walking, peak joint moments at the ankle, knee, and hip were generally higher than in level walking, particularly in the sagittal plane where forward and backward motions are dominant. This suggests that ascending slopes require greater joint moments to counter gravity and ensure stability. However, certain incline angles did not produce significantly different peak joint moments from level walking, implying that moderate inclines may not significantly increase the stress on all joints. For instance, the knee joint in the sagittal plane at a 7.5° incline did not show a significant difference in peak joint moment compared with level walking, suggesting that moderate inclines may be manageable without excessive joint stress. Additionally, in the frontal plane, which relates to lateral stability, the ankle joint at a 5° incline and the hip joint at a 7.5° incline showed no significant increase in peak joint moments, indicating that mild inclines may not demand excessive lateral stabilization.

In decline walking, peak joint moments also increased compared to level walking, highlighting the challenge of controlling descent. The need to absorb impact forces and decelerate the body's momentum, particularly in the sagittal plane, results in elevated joint moments that counteract gravitational pull. However, exceptions were noted, such as the ankle joint in the frontal plane at a 5° decline, where peak joint moments did not differ significantly from level walking. This suggests that at lower decline angles, the lateral stabilization demands may be similar to those on level ground.

Incline and decline in walking generally increase peak joint moments, reflecting higher biomechanical demands on the lower limbs. However, the extent of these increases varies by slope angle, movement plane, and joint, offering insights into the dynamic adjustments required for stability and control on sloped surfaces. Detailed findings for peak joint moments across various conditions are summarized in Table 1 and Table 2, which highlight significant increases in peak joint moments for most inclined and declining scenarios. The results indicated a clear trend of increased biomechanical demands with steeper slopes. Table 1 demonstrates that during inclined walking, the ankle joint exhibited a significant increase in peak joint moments in the sagittal (JMAP(x)) and transverse (JMPD(z)) planes, particularly at 10° inclination (mean: 1.637 ± 0.204 Nm, $p = 0.011$ for JMAP(x); mean: 0.620 ± 0.205 Nm, $p = 0.003$ for JMPD(z)). The knee joint exhibited significant increases in the frontal (JMML(Y)) and transverse planes, with the highest values observed at 10° (mean: 1.810 ± 0.458 Nm, $p < 0.001$ for JMML(Y); mean: 0.987 ± 0.248 Nm, $p < 0.001$ for JMPD(z)). The hip joint demonstrated significant increases in all planes at 10°, particularly in the frontal plane (mean: 2.233 ± 0.568 Nm, $p < 0.001$). Table 2 shows that, during walking, the ankle joint demonstrated significant increases in both the sagittal and transverse planes, with peak moments recorded at 5° ($p = 0.043$ for JMAP(x); $p = 0.019$ for JMPD(z)) and 10° ($p = 0.003$ for JMPD(z)). The knee joint displayed pronounced increases across all planes, with the sagittal plane reaching 1.436 ± 0.806 Nm at 10° ($P = 0.031$). The hip joint showed a substantial increase in the sagittal plane at 10° (mean: 3.295 ± 1.200 Nm, $p < 0.001$), while the transverse plane values were also significantly elevated at steeper inclines ($p = 0.022$).

Table 1. Peak joint moment of ankle, knee and hip during incline walking

	Degree of Inclination (°)	$JM_{AP(x)}$		$JM_{ML(Y)}$		$JM_{PD(z)}$	
		Mean	P-value	Mean	P-value	Mean	P-value
ANKLE	0.0	1.065 ± 0.502	-	0.972 ± 0.731	-	0.231 ± 0.150	-
	5.0	1.171 ± 0.512	0.631	1.087 ± 0.749	0.481	0.380 ± 0.172	0.011*
	7.5	1.417 ± 0.222	0.143	0.506 ± 0.318	0.280	0.297 ± 0.201	0.529
	10.0	1.637 ± 0.204	0.011*	0.944 ± 0.292	1.000	0.620 ± 0.205	0.003*
KNEE	0.0	0.620 ± 0.215	-	0.451 ± 0.134	-	0.168 ± 0.058	-
	5.0	1.142 ± 1.218	0.796	1.167 ± 0.545	0.002*	0.400 ± 0.195	0.001*
	7.5	0.545 ± 0.223	0.481	0.520 ± 0.230	0.631	0.249 ± 0.123	0.280
	10.0	0.796 ± 0.249	0.417	1.810 ± 0.458	0.000*	0.987 ± 0.248	0.000*
HIP	0.0	0.595 ± 0.178	-	0.870 ± 0.204	-	0.206 ± 0.099	-
	5.0	1.721 ± 2.332	0.105	1.198 ± 0.399	0.165	0.618 ± 0.396	0.002*
	7.5	0.679 ± 0.289	0.579	0.763 ± 0.197	0.218	0.290 ± 0.227	0.529
	10.0	1.372 ± 0.818	0.003*	2.233 ± 0.568	0.000*	1.110 ± 0.689	0.002*

Table 2. Peak joint moment of ankle, knee and hip during decline walking

	Degree of Inclination (°)	$JM_{AP(x)}$		$JM_{ML(Y)}$		$JM_{PD(z)}$	
		Mean	P-value	Mean	P-value	Mean	P-value
ANKLE	0.0	1.065 ± 0.502	-	0.972 ± 0.731	-	0.231 ± 0.150	-
	5.0	1.504 ± 0.253	0.043*	0.999 ± 0.761	0.796	0.461 ± 0.532	0.019*
	7.5	1.242 ± 0.371	0.481	0.450 ± 0.177	0.315	0.468 ± 0.361	0.011*
	10.0	1.417 ± 0.371	0.181	0.641 ± 0.337	0.792	0.952 ± 1.276	0.003*
KNEE	0.0	0.620 ± 0.215	-	0.451 ± 0.134	-	0.168 ± 0.058	-
	5.0	1.540 ± 0.747	0.000*	0.987 ± 0.532	0.002*	0.289 ± 0.131	0.003*
	7.5	1.090 ± 0.430	0.011*	1.035 ± 0.387	0.002*	0.417 ± 0.182	0.001*
	10.0	1.436 ± 0.806	0.031*	0.744 ± 0.215	0.007*	0.303 ± 0.096	0.011*
HIP	0.0	0.595 ± 0.178	-	0.870 ± 0.204	-	0.206 ± 0.099	-
	5.0	2.735 ± 1.400	0.000*	0.910 ± 0.296	0.529	0.379 ± 0.204	0.009*
	7.5	1.621 ± 0.780	0.000*	1.812 ± 0.701	0.001*	0.645 ± 0.339	0.000*
	10.0	3.295 ± 1.200	0.000*	1.072 ± 0.295	0.263	0.426 ± 0.224	0.022*

Joint Acceleration on Ankle Knee and Hip during Slope Walking

The study also examined joint accelerations, which reflect the dynamic responses of the joints to inclined and declined walking conditions. The joint accelerations varied significantly depending on the slope angle. For the ankle joint, as shown in Figure 4(a), the acceleration patterns were largely consistent across both the incline and decline conditions, although the magnitude of acceleration varied significantly, especially at the 10° incline. This suggests that steeper slopes require a higher dynamic response from the ankle joint to maintain control. In Figure 4(b), the knee joint and declining walking conditions induced higher accelerations, indicating an additional demand on the knee for control during descent. This increased acceleration reflects the knee's need to stabilize and decelerate the body on a downward slope, where the gravitational pull intensifies the load on the joint. The hip joint in Figure 4(c) also experienced elevated accelerations, particularly during declining walking, which emphasizes the need for increased control to prevent falls. These findings indicate that both inclined and declined walking demand considerable adjustments from the hip joint, especially in response to the body's forward and downward momentum on slopes.

Higher peak joint accelerations were generally observed for inclined conditions than for level walking, although certain angles showed exceptions. For example, at a 7.5° incline, ankle joint acceleration in certain planes did not differ significantly from level walking, suggesting that moderate inclines might not place as much additional demand on the dynamic response of the ankle joint as steeper inclines do. Incline-slope walking has a higher joint acceleration than level walking. Except for ankle joint acceleration in the transverse and frontal planes, and ankle joint acceleration in the sagittal plane at 7.5° inclination. Decline-slope walking has higher knee and hip joint accelerations than level-slope walking. However, the 7.5° inclination showed a different result. Slope walking has a lower ankle joint moment in the sagittal plane but a lower moment in the transverse and frontal planes. The exception was the 10° and 5° inclinations in the sagittal and transverse planes, respectively.

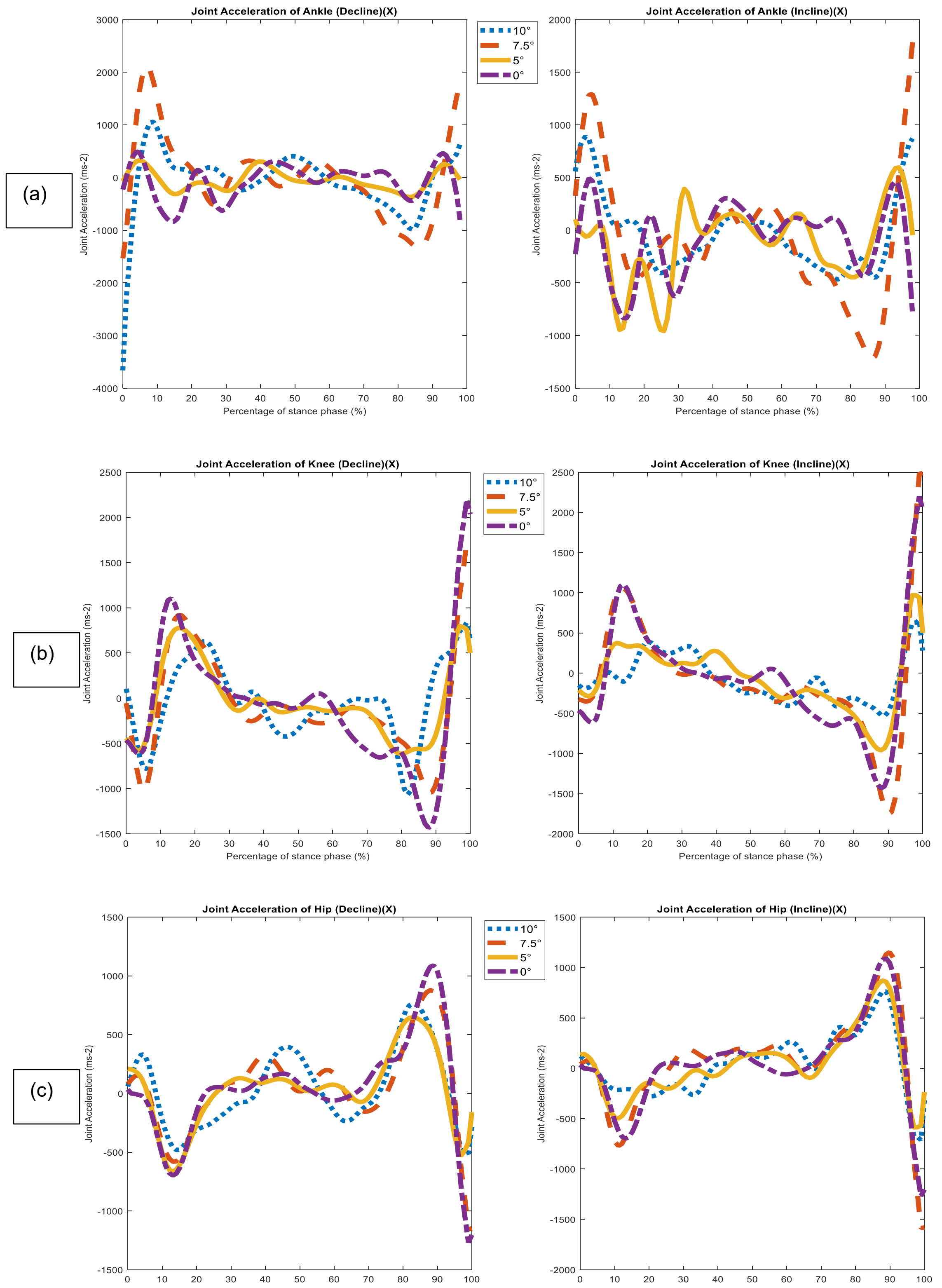

Figure 4. Joint accelerations of the (a) ankle, (b) knee, and (c) hip in the sagittal plane (X) during decline walking at incline angles of 0°, 5°, 7.5°, and 10°

Table 3 demonstrates that during inclined walking, the ankle joint exhibited significant reductions in sagittal acceleration (JAAP(x)) as the slope increased, with values decreasing from 3625.13 ± 1256.77 at 0° to 2133.43 ± 778.22 at 10° ($p < 0.05$). Biomechanically, this reduction reflects a necessary compensatory strategy rather than reduced joint effort. As the incline steepens, individuals naturally adopt a more cautious, deliberate gait characterized by shorter step lengths and reduced walking velocity to maintain balance and prevent backward falls. This slower, controlled progression minimizes rapid dynamic impacts and segmental momentum at the ankle, resulting in lower peak accelerations even though the required net joint moments to propel the body upward against gravity are significantly increased. Conversely, the transverse acceleration (JAPD(z)) showed an increase at 7.5° (3105.17 ± 1226.59, $p > 0.05$) before declining at 10°. The knee joint exhibited a notable increase in sagittal acceleration at 7.5° (2571.43 ± 399.37, $p > 0.05$), whereas transverse acceleration decreased significantly at 10° (1601.23 ± 1090.30, $p = 0.022$). The hip joint displayed significant decreases in transverse acceleration with increasing slopes, from 2579.07 ± 622.02 at 0° to 980.99 ± 297.67 at 10° ($p < 0.001$), while sagittal and frontal accelerations remained relatively stable. Table 4 shows that, during walking, the ankle joint exhibited a biphasic trend in sagittal acceleration, peaking at 10° (4696.32 ± 6012.01, $p > 0.05$). The knee joint showed reduced sagittal acceleration across the slopes, with the highest acceleration observed at 7.5° (2461.18 ± 475.19, $p > 0.05$). The hip joint revealed a substantial increase in transverse acceleration at 5° (1724.85 ± 1276.35, $p = 0.002$), which then stabilized at higher slopes.

Table 3. Peak joint acceleration of ankle, knee and hip during incline walking

	Degree of Inclination (°)	$JA_{AP(x)}$		$JA_{ML(Y)}$		$JA_{PD(z)}$	
		Mean	P-value	Mean	P-value	Mean	P-value
ANKLE	0.0	3625.13 ± 1256.77	-	3041.36 ± 1924.90	-	2597.86 ± 1598.20	-
	5.0	2605.12 ± 2080.10	0.015*	2641.71 ± 1021.93	0.739	2272.04 ± 1524.10	0.529
	7.5	2197.21 ± 632.32	0.005*	3410.12 ± 1762.91	0.529	3105.17 ± 1226.59	0.631
	10.0	2133.43 ± 778.22	0.016*	1719.08 ± 797.04	0.313	2062.01 ± 1140.16	0.562
KNEE	0.0	2294.40 ± 1108.67	-	1921.09 ± 590.41	-	2778.72 ± 1056.83	-
	5.0	1480.26 ± 574.04	0.075	1563.96 ± 635.49	0.123	2214.46 ± 2022.62	0.043*
	7.5	2571.43 ± 399.37	1.000	1041.97 ± 406.27	0.002*	2509.18 ± 1326.40	0.393
	10.0	1502.23 ± 496.27	0.263	1429.87 ± 564.36	0.118	1601.23 ± 1090.30	0.022*
HIP	0.0	1436.47 ± 529.88	-	1047.85 ± 314.99	-	2579.07 ± 622.02	-
	5.0	1147.72 ± 329.52	0.218	759.35 ± 295.73	0.043*	1654.81 ± 757.98	0.009*
	7.5	1635.63 ± 248.22	0.529	929.38 ± 362.17	0.529	1685.71 ± 553.47	0.004*
	10.0	1312.45 ± 551.23	0.792	795.83 ± 245.51	0.118	980.99 ± 297.67	0.000*

Table *4.* Peak joint acceleration of ankle, knee and hip during decline walking

	Degree of Inclination (°)	$JM_{AP(x)}$		$JM_{ML(Y)}$		$JM_{PD(z)}$	
		Mean	P-value	Mean	P-value	Mean	P-value
ANKLE	0.0	3625.13 ± 1256.77	-	3041.36 ± 1924.90	-	2597.86 ± 1598.20	-
	5.0	1992.59 ± 819.30	0.002*	3132.16 ± 839.30	0.684	2428.63 ± 587.35	0.684
	7.5	2931.49 ± 789.94	0.435	3312.88 ± 1551.30	0.631	4338.23 ± 3005.89	0.123
	10.0	4696.32 ± 6012.01	0.313	15508.82 ± 32536.72	0.792	4114.17 ± 4423.56	0.875
KNEE	0.0	2294.40 ± 1108.67	-	1921.09 ± 590.41	-	2778.72 ± 1056.83	-
	5.0	1675.64 ± 427.78	0.315	1400.87 ± 589.90	0.123	2138.90 ± 456.81	0.280
	7.5	2461.18 ± 475.19	0.971	1790.77 ± 1731.72	0.165	3810.86 ± 2656.70	0.436
	10.0	1746.30 ± 906.54	0.562	1621.22 ± 544.76	0.263	2104.65 ± 1158.86	0.263
HIP	0.0	1436.47 ± 529.88	-	1047.85 ± 314.99	-	2579.07 ± 622.02	-
	5.0	1068.49 ± 274.59	0.089	685.24 ± 176.38	0.009	1724.85 ± 1276.35	0.002*
	7.5	1459.76 ± 264.89	0.912	1147.35 ± 523.29	0.579	2051.76 ± 1223.23	0.023*
	10.0	1197.74 ± 578.11	0.492	940.65 ± 374.95	0.428	2384.52 ± 1481.66	0.313

Discussion

In decline walking, kinematic data derived from stride parameters indicate that the knee and hip joints are primarily involved in deceleration and stability control during the walking session along the steepest slope in the sagittal plane, where joint moments gradually increase with slope steepness. Increased transverse moments in the ankle joint during moderate decline underscore the importance of rotational stability in avoiding abrupt torques that could lead to injury. This complements the findings of Lin and Thomas [21], who suggested that even moderate gradients can impose notable biomechanical adaptations. Although this is similar to the findings of previous studies [22,23], some of the results require further research. For instance, moderate slopes do not always result in increased joint moments, suggesting that the body may employ energy-efficient strategies or protective mechanisms under these conditions. Furthermore, the variability in frontal and transverse plane responses, compared to Xie's findings [27], highlights the need to examine factors such as population demographics, walking speeds, and incline types. These findings have important implications for injury prevention and rehabilitation. Increased joint loading on steeper slopes indicates the necessity for targeted strengthening programs, particularly for individuals who frequently navigate inclined environments. Strengthening the knee and ankle muscles to enhance their stiffness and stability may prevent overuse injuries and improve performance. Additionally, these biomechanical insights can inform the design of assistive devices such as exoskeletons and orthotics tailored to mitigate joint stress during slope walking.

Because EMG measurements were not included in this study, the estimation of muscle activity relies on the observed kinetic demands. The increased transverse moments in the ankle joint during moderate decline underscore the importance of rotational stability. Biomechanically, these elevated moments dictate a higher internal demand on the ankle invertor and evertor muscles to co-activate and generate counter-torques, thereby preventing abrupt rotations that could lead to ligamentous injury. Furthermore, higher joint accelerations necessitate rapid force generation to decelerate the body. Therefore, strengthening the knee and ankle muscles to enhance their active stiffness and dynamic stability is highly recommended. Improved muscular conditioning allows the joints to safely absorb the increased kinetic energy and impact forces of slope walking, which may prevent overuse injuries and improve overall functional performance.

Although this study provides valuable biomechanical insights, it must be acknowledged that the participants were healthy young adults (mean age 24 ± 1.2 years). Consequently, the findings do not fully reflect the gait mechanics and physiological limitations of older adults, who are the primary target demographic for fall prevention in an aging society. Therefore, this paper serves as a preliminary study to establish foundational baseline kinetic and kinematic data for slope walking, which will be critical for future comparative studies involving elderly subjects.

Conclusions

This research demonstrates that the biomechanical demands on the lower limbs are significantly greater during slope walking compared to level walking. Inclined walking requires higher joint moments and forces, as the body must be lifted and stabilized through increased activity at the knee and hip joints. Conversely, declined walking places greater demand on the ankle and knee joints, which are essential for controlling motion and absorbing forces at the joints.Statistical analysis reveals that mild slopes (between 5° and 7.5°) do not consistently produce significant increases in peak joint moments across all movement planes compared to level walking (e.g., ankle JMAP(x) at 7.5° yielded $p = 0.143$). This suggests that these moderate inclines induce mechanical demands that remain relatively close to baseline locomotion. However, inclined slopes at 10° significantly elevate biomechanical demands, demonstrating statistically significant increases in peak joint moments ($p < 0.05$) particularly at the knee and hip in the frontal and sagittal planes. This indicates that steeper gradients transfer substantial mechanical strain to the joints, posing a higher risk for musculoskeletal fatigue or injury. Lower limb strengthening exercises should thus be personalized, particularly due to the effect of slope on lower limbs, and more so, hikers, and workers in inclined surroundings. This is one of the main reasons why exercises to increase muscle strength around the ankle, knee, and hip joints must be encouraged to help build joint stability and to prevent such injuries. To further advance the understanding of slope walking biomechanics, it is suggested that future programs increase slope angles and walking speeds as well as consider inter-individual differences.

Conflicts of Interest

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

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References

[1] Deng, R., Li, B., Qin, M., Yu, X., Sun, J., Jiao, F., & Huang, Y. (2024). The characteristics and risk factors of fatal falls among adults aged 60 and above in Southwest China. *Scientific Reports, 14*(1), 7020.

[2] Long, S., Hu, L., Luo, Y., Li, Y., & Ding, F. (2023). Incidence and risk factors of falls in older adults after discharge: A prospective study. *International Journal of Nursing Sciences, 10*(1), 23–29.

[3] Suhendra, N., Gustiono, D., & Masmui, G. N. (2015). Anatomical data analysis and normal femur bone reconstruction for designing titanium alloys total hip joint femoral stem. *Malaysian Journal of Fundamental and Applied Sciences, 11*(4), 152–158.

[4] Yang, Z., Qu, F., Liu, H., Jiang, L., Cui, C., & Rietdyk, S. (2019). The relative contributions of sagittal, frontal, and transverse joint works to self-paced incline and decline slope walking. *Journal of Biomechanics, 92*, 35–44.

[5] Kafetzakis, I., Konstantinou, I., & Mandalidis, D. (2024). Effects of hiking-dependent walking speeds and slopes on spatiotemporal gait parameters and ground reaction forces: A treadmill-based analysis in healthy young adults. *Applied Sciences, 14*(11), 4383.

[6] Nuckols, R. W., Takahashi, K. Z., Farris, D. J., Mizrachi, S., Riemer, R., & Sawicki, G. S. (2020). Mechanics of walking and running up and downhill: A joint-level perspective to guide design of lower-limb exoskeletons. *PLOS ONE, 15*(8), e0231996.

[7] Vernillo, G., Giandolini, M., Edwards, W. B., Morin, J. B., Samozino, P., Horvais, N., & Millet, G. Y. (2017). Biomechanics and physiology of uphill and downhill running. *Sports Medicine, 47*, 615–629.

[8] Veerman, Q. W., Ten Heggeler, R. M., Tuijthof, G. J., de Graaff, F., Fluit, R., & Hoogeslag, R. A. (2024). High variability exists in 3D leg alignment analysis, but underlying principles that might lead to agreement on a

universal framework could be identified: A systematic review. *Knee Surgery, Sports Traumatology, Arthroscopy*. https://doi.org/10.1002/ksa.12512

[9] Zhuang, H., Wang, J., Wang, N., Li, W., Li, N., Li, B., & Dong, L. (2024). A review of foot–terrain interaction mechanics for heavy-duty legged robots. *Applied Sciences, 14*(15), 6541.

[10] Yamin, N. A. A. A., Basaruddin, K. S., Ijaz, M. F., Mat Som, M. H., Shahrol Aman, M. N. S., & Takemura, H. (2024). Stress analysis on the ankle joint during incline and decline standing. *Applied Sciences, 14*(18), 8168.

[11] Belal, M., Alsheikh, N., Aljarah, A., & Hussain, I. (2024). Deep learning approaches for enhanced lower-limb exoskeleton control: A review. *IEEE Access, 12*, 43884.

[12] Emery, C. A. (2018). Injury prevention in kids' adventure and extreme sports: Future directions. *Research in Sports Medicine, 26*(sup1), 199–211.

[13] Chodkowska, K., Borkowski, R., & Błażkiewicz, M. (2024). Perturbations during gait on a split-belt treadmill: A scoping review. *Applied Sciences, 14*(21), 9852.

[14] Barzyk, P., Zimmermann, P., Stein, M., Keim, D., & Gruber, M. (2024). AI-smartphone markerless motion capturing of hip, knee, and ankle joint kinematics during countermovement jumps. *European Journal of Sport Science, 24*(10), 1452–1462.

[15] Lin, Y., Rankin, J. W., Lamas, L. P., Moazen, M., & Hutchinson, J. R. (2024). Hindlimb kinematics, kinetics, and muscle dynamics during sit-to-stand and sit-to-walk transitions in emus (*Dromaius novaehollandiae*). *Journal of Experimental Biology*, jeb-247519.

[16] Higgins, S., Dickin, D. C., Hankemeier, D., Wells, M. D., & Wang, H. (2024). The effect of incline walking on lower extremity and trunk mechanics in older adults. *Sports Medicine and Health Science, 7*(1), 56–60.

[17] Zhang, Z., Xu, D., Gao, X., Liang, M., Baker, J. S., & Gu, Y. (2024). The effect of different degrees of ankle dorsiflexion restriction on the biomechanics of the lower extremity in stop-jumping. *Applied Bionics and Biomechanics, 2024*(1), 9079982.

[18] Patti, A., Gervasi, M., Giustino, V., Figlioli, F., Canzone, A., Drid, P., Thomas, E., Messina, G., Vicari, D. S. S., Palma, A., & Bianco, A. (2024). The influence of ankle mobility and foot stability on jumping ability and landing mechanics: A cross-sectional study. *Journal of Functional Morphology and Kinesiology, 9*(3), 160.

[19] Jie, T., Li, J., & Teo, E. C. (2024). Limb biomechanical difference between bounced and alternating jumping rope. *International Journal of Biomedical Engineering and Technology, 45*(2), 150–163.

[20] Raitor, M., Ruggles, S., Delp, S. L., Liu, C. K., & Collins, S. H. (2024). Lower-limb exoskeletons appeal to both clinicians and older adults, especially for fall prevention and joint pain reduction. *IEEE Transactions on Neural Systems and Rehabilitation Engineering, 32*, 1577–1585.

[21] Lin, J., Thomas, G. C., Divekar, N. V., Peddinti, V., & Gregg, R. D. (2024). A modular framework for task-agnostic, energy shaping control of lower limb exoskeletons. *IEEE Transactions on Control Systems Technology, 32*(6), 2359–2375.

[22] Teater, R. H., Zelik, K. E., & McDonald, K. A. (2024). Biomechanical effects of adding an articulating toe joint to a passive foot prosthesis for incline and decline walking. *PLOS ONE, 19*(5), e0295465.

[23] Namayeshi, T., Lee, P. V. S., & Ackland, D. (2024). Gait balance recovery during tripping: The influence of walking speed and ground inclination on muscle and joint function. *Journal of Biomechanics, 172*, 112178.

[24] Wang, W. F., Lien, W. C., Liu, C. Y., & Yang, C. Y. (2018). Study on tripping risks in fast walking through cadence-controlled gait analysis. *Journal of Healthcare Engineering, 2018*(1), 2723178.

[25] Leardini, A., Benedetti, M. G., Catani, F., Simoncini, L., & Giannini, S. (1999). An anatomically based protocol for the description of foot segment kinematics during gait. *Clinical Biomechanics, 14*(8), 528–536.

[26] Schut, I. M., Engelhart, D., Pasma, J. H., Aarts, R. G., & Schouten, A. C. (2017). Compliant support surfaces affect sensory reweighting during balance control. *Gait & Posture, 53*, 241–247.

[27] Xie, H., & Chien, J. H. (2024). Walking on different inclines affects gait symmetry differently in the anterior-posterior and vertical directions: Implication for future sensorimotor training. *PeerJ, 12*, e18096.

[28] Ramlee, M. H., Wahab, A. H. A., Wahab, A. A., Latip, H. F. M., Daud, S., & Kadir, M. (2017). The effect of stress distribution and displacement of open subtalar dislocation in using titanium alloy and stainless steel Mitkovic external fixator—A finite element analysis. *Malaysian Journal of Fundamental and Applied Sciences, 13*, 477–482.

[29] Walker, R. L., O'Brien, T. D., Barton, G. J., Carter, B., Wright, D. M., & Foster, R. J. (2025). Are challenging walking environments linked to falls or risk of falling in children with cerebral palsy? A systematic review. *Gait & Posture*.

[30] Li, W., Zhang, Y., & Chien, J. H. (2025). Applying bilateral mastoid vibration changes the margin of stability in the anterior–posterior and medial–lateral directions while walking on different inclines. *European Journal of Medical Research, 30*(1), 108.

[31] Schoenwether, B., Ripic, Z., Nienhuis, M., Signorile, J. F., Best, T. M., & Eltoukhy, M. (2025). Reliability of artificial intelligence-driven markerless motion capture in gait analyses of healthy adults. *PLOS ONE, 20*(1), e0316119.

[32] Saini, S., Shikalgar, A., Tejani, N., Palekar, T. J., & Kumar, P. (2025). Effectiveness of a gait assessment model in enhancing understanding and learning in healthcare education. *Internet Journal of Allied Health Sciences and Practice, 23*(2), 3.