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Accelerations and EMG Differences Between Isocaloric High-Incline Walking and Level-Grade Jogging

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Abstract

International Journal of Exercise Science 18(7): 995-1009, 2025. https://doi.org/10.70252/LKZW7240 High-incline walking is a popular mode of exercise and often serves as an alternative to level-grade running for improving cardiovascular fitness. This study examined the muscle activation and biomechanical differences between high-incline walking (HIW) at a 20% grade and level-grade jogging (LGJ) at matched exercise intensities. Nineteen physically inactive adults (18-31 years) participated. Participants completed two isocaloric exercise trials (HIW and LGJ), each lasting ten minutes. EMG data were obtained from eight muscles: biceps femoris (BF), gluteus maximus (GM), lateral gastrocnemius (LG), tibialis anterior (TA), vastus lateralis (VL), anterior deltoid (AD), erector spinae (ES), and soleus (SOL). Resultant accelerations on the foot and sacrum were measured using inertial measurement units (IMUs). At matched isocaloric intensities between LGJ and HIW, mean peak resultant acceleration was significantly higher in LGJ for the foot (22.14 ± 8.44 m/s^2) and sacrum (27.21 ± 7.92 m/s^2). Peak EMG activation was significantly greater during LGJ in TA (40.9% ± 21.4) and VL (52.6% ± 39.8). The EMG integral was significantly greater in the LGJ for the TA (13.8% ± 5.0) than during HIW. Despite being matched isocalorically, LGJ increased peak and integral muscle activation and produced higher foot and sacrum accelerations. These findings highlight HIW as a lower-impact, yet equally demanding, alternative to LGJ, with implications for exercise prescription and injury prevention.

Keywords: Gait, lunge, lower limb accelerations

Introduction

Human locomotion relies on the lower-body muscles to coordinate limb movements during activities such as walking, jogging, hiking, and running.[1] Beyond transportation, many people engage in these activities for exercise, often using treadmills to adjust variables such as speed and grade to alter workout intensity. The IHRSA Health Club Consumer Report (2019) revealed that treadmills are the most popular equipment in health clubs, with over half of members using them for walking (51%) and more than a quarter for running or jogging (28%). Given the popularity of treadmills as primary or preferred exercise tools, equipment manufacturers have sought to meet ever-evolving fitness trends and consumer demand for safer and more effective exercise options.

To cater to the demands of gym-goers and at-home exercisers, many companies have begun manufacturing and promoting high-incline treadmills, with some models reaching inclines of up to 40%. These treadmills target claims that incline exercise increases muscle activation and decreases impact forces. While studies have examined muscle activation and impact force at inclines of up to ~15%,[2–6] few have investigated the specific advantages of high inclines (>18% grade) and their effects on biomechanical factors, such as impact forces and lower leg accelerations. Research suggests that inclines above ~15% alter coordination, resulting in a leg swing pattern more akin to a walking lunge than level surface walking.[7] However, these findings have not been extended to the higher inclines now available with modern treadmills, nor have the effects of prolonged activity on lower limb accelerations or muscle activation been thoroughly explored. Understanding these effects is essential because impact forces and the repetitive shocks associated with running are considered significant contributors to commonly experienced running injuries.[1,8]

Given the high rates of injuries among runners, numerous studies have explored alternative modes of locomotion aimed at reducing the risk of injury while providing a comparable cardiovascular workout.[9,10] Swain et al compared walking at an 11% grade with level-grade running at a matched intensity and found that ground reaction forces were higher during running. They also noted that increasing the walking speed from moderate to vigorous did not increase loading forces.[11] These findings suggest that subjects can increase speed, thereby amplifying the intensity during incline walking without increasing loading forces, contrary to what occurs when increasing running speed.[11,12] It also suggests that incline walking is a suitable alternative to running for cardiovascular exercise, which may help mitigate biomechanical stresses associated with impact. However, Swain et al. only tested inclines <11%, so it remains unclear whether steeper inclines further reduce injury risk and impact forces while still providing comparable cardiovascular benefits.

Assessing whether steeper inclines reduce impact requires reliable measures of loading, which are traditionally obtained from ground reaction forces (GRFs). GRFs help quantify impact but require specialized lab-based equipment such as instrumented treadmills or walkways. Advances in inertial measurement units (IMUs) now enable the collection of accelerations outside the lab at a lower cost.[13,14] Elevated lower-limb accelerations recorded with IMUs have been associated with overuse injuries, stress fractures, and biomechanical imbalances, making these data valuable for injury detection and prevention.[1,8] Additionally, the peak accelerations from IMUs correlate strongly with GRFs, supporting their use as indicators of biomechanical stress during running and walking.[13,15,16] The position of the IMU sensor plays a crucial role in the accuracy and reliability of the acceleration data.[17,18] Sacrum-mounted sensors provide insights into whole-body accelerations, whereas foot-mounted sensors capture localized impact and impact characteristics directly tied to lower limb stress.[15,16,19–21] Each placement offers a complementary perspective on impact forces and their relationship to muscle activation patterns, thereby helping explain alterations in gait mechanics and other variables at high inclines.[2,4]

This augmentation of gait during incline activities has been shown to increase the activation of specific lower extremity muscles, enabling them to meet the different requirements of uphill locomotion.[2] Research by Lay et al demonstrated an increase in both the magnitude and duration of EMG activity in the gluteus maximus, biceps femoris, soleus, and tibialis anterior during uphill walking at 15% and 39% inclines compared to level-grade walking.[4] However, few studies have compared incline and level-grade activities when matched for intensity, leaving it unclear whether observed differences in muscle activation and impact forces are due to grade itself or simply differences in intensity.

To match intensity, it is essential to understand the metabolic cost of running, which has been attributed to the "cost of generating force" hypothesis. This hypothesis suggests that consumption increases in proportion to the weight supported. Kram and Taylor further explained that, at any given running velocity, the vertical force equals body weight; however, shorter ground contact times at faster speeds increase the required rate of force production. Together with muscle volume, these factors largely account for the metabolic cost of running.[22-24] Matching the metabolic cost provides a valid framework for comparing biomechanical differences independent of energy expenditure.[5,23,24] In theory, matched work requires similar total muscle volume and force production. In practice, however, the way that work is distributed across muscles can vary depending on the mode of locomotion and slope. During incline walking, redistribution becomes more pronounced as gradients increase, and gait adapts to higher inclines.[2] According to Gidley and Bailey, coordination changes emerge at slopes of ~15-20%, producing a lunge-like gait. This altered movement pattern suggests that high-incline walking may alter muscle recruitment in ways not observed during level-grade jogging. These findings highlight how slope influences coordination strategies, but beyond this, little is known about how high-incline walking compares with level-grade activity when matched for energy expenditure, particularly with respect to lower limb accelerations and muscle activation. This gap forms the basis of the present study.

The primary purpose of this study is to investigate the differences in foot and sacrum resultant accelerations and the differences in muscle activation between two isocaloric exercises: level-grade jogging (LGJ) and high-incline walking (HIW, 20% grade). The secondary purpose is to determine whether the resultant accelerations and muscle activation change over time. The results of this study may help inform exercise prescriptions for individuals seeking alternatives to traditional treadmill running.

Methods

Participants

Nineteen subjects (10F/9M) volunteered for this study (age: 25± 4 years; height: 168± 8.5 cm; mass: 71.57± 13.12 kg). All subjects reported performing <150 minutes a week of moderate aerobic exercise or <75 minutes of vigorous exercise per week in the past six months. All met the requirements of having no known neuromuscular, cardiovascular, or orthopedic diseases. Each subject completed a written informed consent form before participating, as per the California State Polytechnic University, Humboldt Institutional Review Board.

Protocol

Participants first completed pre-testing, followed by two isocaloric exercise testing sessions on separate days. They then returned at least 24 hours after the last exercise session to undergo biomechanical testing to mitigate the effects of fatigue.

Participants reported to the lab to review the IRB consent forms, ensure that participation criteria were met, and receive instructions for the upcoming sessions. Age and height were recorded. All participants underwent a body composition analysis using the Cosmed USA Inc. BODPOD.

After pre-testing, subjects returned to the lab and followed the isocaloric exercise testing protocol established by Sato et al. All participants were asked to refrain from eating within two hours prior to the test but had eaten within the last six hours, and to avoid caffeine, alcohol, and nicotine use for at least four hours. Additionally, they were instructed to refrain from vigorous resistance training and aerobic exercise for at least 14 hours before testing. Participants took part in a ~5-minute familiarization session to establish a jogging speed that was estimated to be below the lactate threshold on the Trackmaster TMX425C (Full Vision Inc., Newton, KS) treadmill, which represents a Respiratory Exchange Ratio (RER) of 0.83-0.92 and was verbally affirmed to be comfortably sustained for at least 40 minutes. Jogging was used first to establish a sustainable speed, as some subjects were unable to safely reach a comparable VO_2 during HIW in pilot testing. Breath-by-breath gas exchange analysis was performed every 2 minutes using the ParvoMedics TrueOne 2400 (ParvoMedics, Sandy, UT) metabolic cart. Heart rates were measured using the Polar H1 heart rate monitor (Polar, Kempele, Finland). Once speed was established, participants rested for ~5 minutes. Next, participants began level-grade jogging at the established speed for 20 minutes. The exercise bout lasted 20 minutes; metabolics, speed, RER, and HR were recorded.

Participants returned to the lab at least 24 hours later to complete the HIW protocol. The Lankford equation for oxygen uptake was used as a starting point for establishing an isocaloric speed.[25] Participants then started a ~5-minute familiarization session at the 20% incline. The speed was adjusted to ensure participants reached a respiratory exchange ratio (RER) of 0.83-0.92 and to ensure the Relative Oxygen Consumption (VO_2) in milliliters per kilogram per minute (mL/kg/min) was within ±3.0 of the previous session's (VO_2) measurements. Once speed was established, participants rested for ~5 minutes. Participants then began walking at the established speed and 20% incline; the exercise bout lasted 20 minutes, and metabolic data, speed, RER, and HR were collected. The speed from the trials was recorded and used in the biomechanical protocol.

The total subject pool included in the final analysis consisted of 19 participants (9 males, 10 females) (See Table 1).

Table 1. Subject anthropometric and descriptive information.

Variable	Mean ± *SD*
Age (yr)	25 ± 4.2
Height (cm)	168 ± 8.5
Weight (kg)	71.57 ± 13.12
Body Fat % (BF)	25.19 ± 9.26

Note. Values are represented by mean ± SD. n=19. yr = years. cm = centimeters. kg = kilograms. BF % = body fat percentage.

Subjects returned to the lab to perform two 10-minute activities on the treadmill, one HIW (20%) and one LGJ (0%), at the speeds that were isocalorically matched in pretesting. Bipolar surface electrodes (Ag/AgCl 10 mm IED, Trigno Delsys) were placed on the Tibialis Anterior (TA), Soleus (SOL), Lateral Gastrocnemius (LG), Vastus Lateralis (VL), Biceps Femoris (BF), Gluteus Maximus (GM), Anterior Deltoid (AD), and Erector Spinae (ES) according to SENIAM guidelines.[26] Site locations were shaved, cleaned, and lightly abraded to improve the signal-to-noise ratio before placing the electrodes. Electrode positions and signal quality were visually inspected on each muscle before collection began. Subjects were encouraged by the researchers to contract and hold the contraction for ~3 seconds maximally. Subjects were given a rest period of at least 2 minutes between each MVC. EMG signals were collected at 2000 Hz and pre-amplified with a gain of 1700 (input impedance > 100 MΩ, standard mode rejection ratio > 110 dB at 60 Hz). Following the MVC trials and a ~5-minute rest period, participants were asked to complete both activities, incline walking and level-grade jogging, at the speeds determined by Sato et al. (2024) for ten minutes, with at least a 5-minute break between each test. Participants were randomly assigned which of the two activities they completed first on the treadmill (Trackmaster TMX425C, Full Vision Inc., Newton, KS).

EMG activity was recorded after the first minute for 30 seconds and then again in the last 30 seconds of the 10-minute trial. This was repeated for the second ten-minute trial. Mean peak EMG activation across the stance phase was expressed as a percent of MVC, and the area under the curve or integral for each muscle during LGJ and HIW was expressed as a percent of MVC. Before subject arrival, IMU devices were calibrated using a nine-camera Vicon Motion Capture system. IMUs were placed on the top of the right foot and the sacrum and collected accelerations at 1125 Hz in the x, y, and z directions. Data were analyzed using a custom Visual3D pipeline that utilized IMU accelerations and manually created events from visual identification (video camera footage) of foot strikes and toe-offs to generate stance events. Peak resultant acceleration for the foot and sacrum was calculated for ten strides and averaged to find the mean peak resultant acceleration for LGJ and HIW during the two time periods (Minute 1 and Minute 9.5). AUC and peak activation were also calculated from 10 strides during the 30-second collection.

Statistical Analysis

A mixed model 2x2 ANOVA was used to identify differences in mode (LGJ, HIW), time (min 1, min 9.5), and interactions (mode x time). A priori planned comparisons were used for LGJ min

1 vs. LGJ min 9.5, HIW min 1 vs. HIW min 9.5, LGJ min 1 vs. HIW min 1, and LGJ min 9.5 vs. HIW min 9.5. Post hoc comparisons were analyzed using Bonferroni adjustments. Variables analyzed were EMG AUC, EMG PA, and IMU peak resultant acceleration. Statistical analysis was performed using JASP (JASP Team, 2023), with significance set at $p < 0.05$.

Results

Resultant Accelerations

All pairwise comparisons are found in Table 2. Resultant accelerations of the foot demonstrated significant differences between mode ($p<.001$) and mode x time ($p=.009$). Foot resultant accelerations for LGJ min 1 (22.14±8.44 m/s^2) were significantly higher than HIW min 1 (10.48 ± 3.47 m/s^2, $p<.001$, $d = 1.711$), and resultant accelerations for LGJ min 9.5 (25.37±9.66 m/s^2) were significantly higher than HIW min 9.5 (10.31 ± 2.97 m/s^2, $p<.001$, $d = 2.211$). LGJ min 1 (22.14 ± 8.44 m/s^2) was significantly lower than LGJ min 9.5 (25.37 ± 9.66 m/s^2, $p=.004$, $d = 0.474$). All other foot acceleration resultant comparisons were not significant.

Sacrum accelerations demonstrated significant differences between modes ($p<.001$). Sacrum resultant accelerations for LGJ min 1 (27.21 ± 7.92 m/s^2) were significantly higher than HIW min 1 (12.58 ± 3.08 m/s^2, $p<.001$, $d = 2.062$). Sacrum resultant accelerations for LGJ min 9.5 (30.11 ± 10.55 m/s^2) were significantly higher than HIW min 9.5 (13.39 ± 4.24 m/s^2, $p<.001$, $d = 2.355$). All other sacrum acceleration resultant comparisons were not significant.

Table 2. The mean of peak resultant values of the foot and sacrum IMU sensors

Segment	LGJ MIN 1	HIW MIN 1	LGJ MIN 9.5	HIW MIN 9.5
Foot	22.14 ± 8.44*	10.48 ± 3.47	25.37 ± 9.66*#	10.31 ± 2.97
Sacrum	27.21 ± 7.92*	12.58 ± 3.08	30.11 ± 10.55*	13.39 ± 4.24

All data are expressed in m/s^2. * = Significant difference between LGJ and HIW at the same time point ($p < .001$). # = Significant changes over time within each mode.

Peak EMG Activation

Peak EMG activation data are found in Table 3. Significant differences existed between modes for TA ($p = 0.003$) and AD ($p < 0.001$). Significant interactions between mode and time were present in TA ($p = 0.03$), AD ($p < 0.001$), and SOL ($p = 0.007$). Peak EMG activation for the TA during LGJ min 1 (40.9% ± 21.4) was significantly higher than HIW min 1 (31.5% ± 18.4, $p=.002$, $d = 0.481$). Peak EMG activation was also significantly greater for the TA during LGJ min 1 (40.9% ± 21.4) than LGJ min 9.5 (34.6% ± 17.7, $p<.001$, $d = 0.324$). For the AD, LGJ min 9.5 (6.0% ± 5.7) was significantly greater than HIW min 9.5 (3.9% ± 4.7, $p = .005$, $d = 0.49$). For the SOL, LGJ min 1 (91.8% ± 49.2) was significantly greater than LGJ min 9.5 (79.0% ± 45.1; $p = .002$, $d = 0.273$). Peak activation of the GM was significantly greater in LGJ min 1 (35.8% ± 25.3) when compared to LGJ min 9.5 (30.6% ± 19.9, $p=.015$, $d = 0.234$). The peak activation of the LG was also greater during LGJ min 1 (76.8% ± 30.5) compared to LGJ min 9.5 (64.4% ± 26.8, $p=.005$, $d =$

0.4). Peak activation for the VL was higher in LGJ min 1 (52.6% ± 39.8) compared to HIW min 1 (42.4% ± 46.1, p=.01, d = 0.511). Peak activation of the VL was also higher in LGJ min 9.5 (50.9% ± 45.1) than in HIW min 9.5 (39.8% ± 47.0, p=.004, d = 0.561). All other peak activation comparisons were not significant.

Table 3. *Mean (SD) of peak activation as a percent of MVC*

Muscle	*LGJ MIN 1*	*HIW MIN 1*	*LGJ MIN 9.5*	*HIW MIN 9.5*
AD	5.2% ± 5.5	3.9% ± 5.2	6.0% ± 5.7*	3.9% ± 4.7
BF	27.9% ± 12.5	23.2% ± 14.6	28.6% ± 14.2	22.0% ± 12.7
ES	35.2% ± 58.3	19.7% ± 10.3	40.5% ± 58.5	20.7% ± 11.6
GM	35.8% ± 25.3	33.3% ± 21.4	30.6% ± 19.9#	32.8% ± 22.7
TA	40.9% ± 21.4*	31.5% ± 18.4	34.6% ± 17.7#	29.5% ± 20.3
VL	52.6% ± 39.8*	42.4% ± 46.1	50.9% ± 45.1*	39.8% ± 47.0
LG	76.8% ± 30.5	61.8% ± 33.3	64.4% ± 26.8#	54.8% ± 28.6
SOL	91.8% ± 49.2	73.9% ± 46.1	79.0% ± 45.1#	71.9% ± 47.0

* = Significance between LGJ and HIW (p<.05), # = significance in time-based changes within each mode (LGJ or HIW)

Table 4. *Mean (SD) of integral (area under the curve) as a percent of MVC*

Muscle	*LGJ MIN 1*	*HIW MIN 1*	*LGJ MIN 9.5*	*HIW MIN 9.5*
AD	1.4% ± 0.7	1.4% ± 1.2	1.8% ± 1.1	1.6% ± 0.9
BF	10.4% ± 4.7	11.5% ± 6.1	11.1% ± 6.4	11.9% ± 6.3
ES	18.7% ± 27.9	15.7% ± 12.7	19.6% ± 27.0	16.1% ± 13.9
GM	12.5% ± 8.5	15.5% ± 12.0	13.9% ± 9.8	16.0% ± 11.5
TA	13.8% ± 5.0*	10.1% ± 3.8	11.2% ± 5.0#	9.1% ± 3.2
VL	18.7% ± 7.7	18.8% ± 9.8	18.3% ± 8.6	18.6% ± 8.4
LG	29.4% ± 13.5	31.3% ± 20.7	25.0% ± 11.5	29.1% ± 20.3
SOL	36.7% ± 21.6	37.6% ± 32.0	31.8% ± 19.4#	37.8% ± 29.5

* = Significance between LGJ and HIW (p<.05), # = significance in time-based changes within each mode (LGJ or HIW)

Integral EMG Activation

The area under the curve of the TA, SOL, LG, BF, VL, GM, AD, and ES are shown below in Table 4. Significant differences between modes existed for TA (p=.01). Significant interactions between mode and time were present in SOL (p=.01) and TA (p=.02). AUC for the SOL was significantly higher during the LGJ min 1 (36.7% ± 21.6) than in LGJ min 9.5 (31.8% ± 19.4, p=.03, d = 0.187). AUC for the TA during LGJ min 1 (13.8% ± 5.0) was significantly greater than HIW min 1 (10.1%

± 3.8, p=.019, d = 0.848) and significantly higher than LGJ min 9.5 (9.1% ± 3.2, p<.001, d = 0.602). All other area-under-the-curve comparisons were not significant.

Discussion

Resultant Acceleration

Resultant accelerations gathered by IMUs placed on the foot and sacrum highlighted higher resultant accelerations during LGJ than in the HIW (p<.001). This finding aligns with previous research by Gottschall and Kram, and Swain et al which indicated that loading rates in the lower limb decrease as the incline increases. Moreover, even when activities are matched for intensity, level-grade exercises produce higher loading forces. The heightened resultant accelerations observed in the foot and sacrum during level ground jogging compared to high-incline walking underscore the increased biomechanical stress and potential risk of injury associated with jogging on flat surfaces. This comparison emphasizes the advantages of incline walking as a lower-impact alternative for cardiovascular exercise.

While it cannot be stated with certainty that an increase in resultant accelerations is directly comparable to GRF forces or loading forces, studies demonstrating the accuracy and correlation between skin-mounted accelerometers and force plates have yielded promising results.[13,14,16] Thus, it is logical to assume that the increase in accelerations of the foot and sacrum suggests that there is likely also an increase in the force relationship between the ground and the body. Walking on a steep incline results in less vertical foot descent for each stride because of the increased elevation gain required by each step. This reduces the impact forces at foot strike, as the body's mass does not shift entirely independently onto the landing foot. Other studies have demonstrated that incline walking significantly alters gait mechanics that mitigate shock transmission to the lower extremities.[2,4] Other factors, such as the needed alteration in the body's center of mass, the vertical displacement during each stride, the change in friction demands, and the difference in foot clearance, are also responsible for the reduction in impact forces during incline walking, as seen in other studies when compared to level-grade exercise.[11,12,27]

Consistent with expected gait differences, LGJ involved a higher step frequency than HIW. Additionally, the slower cadence favored during incline walking further reduces the impact loading rate, as slower gait velocities allow for more controlled and gradual force application, decreasing the peak ground reaction forces.[11] Incline walking reduces foot drop, slows cadence, and results in biomechanical adjustments, such as increased knee flexion and plantarflexed foot position, collectively enhancing shock absorption and reducing resultant accelerations.[28] These findings suggest that lesser foot drop, slower cadence, and biomechanical adaptations effectively contribute to the lower resultant accelerations observed during high-incline treadmill walking.

The significant increase in resultant accelerations between time points during LGJ for the foot but not for the sacrum can be explained by the peak activation values of the erector spinae. These values, although not significant, can help reveal how the body responds to the impact of the lower limbs striking the ground. The erector spinae, in particular, is used to stabilize the unilateral hip flexion and extension movements that occur during running.[28] While significant increases in foot resultant accelerations were observed during LGJ over time, sacrum accelerations remained stable, possibly due to the stabilizing role of the erector spinae in mitigating shock transmission.[29] The increase in foot resultant accelerations can also be partially explained by the significant decrease in TA AUC and peak EMG values between time points during the LGJ, as the TA is an essential muscle for ankle control and dorsiflexion. During running, it has a higher sustained level of activity compared to other muscles, which makes it more susceptible to fatigue and related injuries.[6,30] There were no observed differences in the HIW over time, and the resultant accelerations were significantly lower. This suggests that the risk of impact-related injuries during HIW is reduced and that continued submaximal exercises are unlikely to substantially increase the resultant accelerations. Further testing is necessary to confirm the seemingly stable resultant acceleration values during HIW over time, as well as to assess how different speeds and intensities at high inclines may affect them.

Peak Activation

All muscles demonstrated higher average peak activation during the LGJ. This is likely due to the increased contraction speed and step count needed to meet the demands of running when compared to walking. Swanson and Caldwell have similarly highlighted the elevated step count and the associated increase in contraction speed during running compared to walking, providing ample reason for the heightened EMG activity observed during LGJ compared to HIW. This aligns with the widely accepted physiological understanding that faster and more dynamic movements necessitate greater muscle engagement. Further reinforcing this idea, Gazendam and Hof found that EMG profiles during running could be decomposed into basic patterns distinct from walking, with several muscles showing speed-dependent activation that varied significantly between running and jogging. This suggests that as speed increases, so does the demand on the muscles, resulting in higher EMG activation during running compared to walking.

The significantly higher AD activation during the LGJ compared to HIW could result from the increased counterbalance movements required at higher speeds during jogging[2,31]. All the muscles tested demonstrated higher peak activation during LGJ. An explanation for this can be drawn from a study by Kyröläinen et al, which further supports the results by demonstrating that EMG activities of leg muscles increased with running speed, highlighting the dynamic nature of muscle activation in response to changing exercise demands, most notably speed. Guidetti et al noted significant variability in EMG profiles among individuals during running, underscoring the complexity of neuromuscular control from person to person. This variability was especially pronounced at higher speeds, with many subjects exhibiting distinct running styles and strike patterns compared to one another. Participants also demonstrated shifts in running dynamics during the trials, likely to alleviate localized muscle fatigue. This could help explain why the peak EMG activity revealed that the GM, TA, AD, LG, and SOL muscles displayed significant decreases in percent activation levels between minutes 1 and 9.5 during LGJ, but not during HIW. This suggests that while both activities were matched isocalorically, implying that the rate of energy expenditure was the same, the two activities had varying effects on the muscles over time.

Research suggests that muscle fatigue can manifest as either an increase or a decrease in EMG amplitude and is significantly influenced by exercise conditions, differing between maximally fatiguing contractions and submaximal efforts. During maximal efforts, such as sprinting or weightlifting, the body quickly mobilizes a considerable number of motor units to generate force, resulting in increased peak EMG activity over time. This is due to the recruitment of additional motor units and the increased firing rate of already active units, which helps meet the body's requirements. Nummela et al examined EMG activities and ground reaction forces during fatigued and non-fatigued sprinting and observed increased activation near the end of a maximal effort, indicating heightened neuromuscular activity to sustain force production amidst muscular fatigue. Schlink et al provided evidence that muscles can show a decrease in peak activation during submaximal exercise. This decline during running was attributed to the repetitive and consistent load placed on specific muscle groups, leading to localized fatigue and a diminished ability to maintain high muscle activation levels. These nuanced responses underscore the complexity of the neuromuscular system's adaptation to different exercise intensities and fatigue levels, and they help explain why there was a decrease in peak activation over time during the LGJ. This decrease could also be due to individuals increasing their efficiency as they exercise and not activating muscles that are not necessary for meeting the activity's demands. With subjects reporting less than 150 minutes of moderate aerobic activity each week, likely, LGJ and HIW at the selected pace and duration are somewhat of a novel stimulus, despite the familiarization trials.

In contrast, HIW did not display a significant decrease in activation for any of the muscles. This could be due to the unique and dynamic stride associated with HIW, as well as the increased demand for balance and propulsion resulting from the incline. While some subjects might have noted a burning sensation in the lower leg muscles, fatigue could have been offset by the "walking lunge" that occurs in grades above 18%.[7] This altered gait may engage a broader range of muscle fibers and distribute the load more effectively among different muscle groups, thereby preventing the same level of fatigue in any single muscle. The walking mechanics at high gradients become progressively more focused on lifting the center of mass as the slope increases.[5] A shift in coordination at more extreme inclines is attributed to a need for the extension of joints in response to the increased flexion at touchdown.[7] This increased need for extension in response to increased flexion at touchdown increases torque on each joint, likely requiring greater force; however, when paired with a longer contact time, it results in lower peak activation values for many of the lower-body muscles.

These findings reignite the discussion on whether level-grade or high-incline exercises have a more beneficial impact on muscles, and they also spark questions regarding the effects of each mode. Millet and Lepers explored the impact of ultra-endurance running on neuromuscular function, shedding light on the phenomenon where various muscle groups and fiber types exhibit distinct fatigue rates. Although this study did not encompass ultra-endurance activities, the insights from their research can help explain the absence of a consistent decline in peak EMG activity in muscles during HIW, as they further support the idea that a shift in fibers was occurring to mitigate fatigue.

Few studies have compared different forms of locomotion at matched intensities (kcal/min), and it is challenging to address all the potential variables in a single bout. Results from this study revealed a series of novel discoveries regarding changes and differences between and within subjects throughout the testing period. Some subjects utilized a heel-first stride during the HIW, while others opted to pivot more onto the balls of their feet, and some shifted their stance throughout the trial. Factors such as trunk angle and head tilt also varied widely across participants and may play a role in compensating for the intense 20% grade during the HIW. While neither mode seemed more affected by time, some participants reported calf soreness or fatigue following the HIW activity, while others did not. It is possible that the ten-minute duration was not sufficient to develop detectable differences in EMG data. Future research should investigate whether these metrics change over time or play a significant role in performance or enjoyment during submaximal incline exercise. Additionally, it should examine whether, at higher or lower intensities, there is a shift in trends for either EMG or IMU data between the two isocaloric activities.

Area Under the Curve

Since both HIW and LGJ were matched on oxygen uptake, it can be assumed that participants' total activated muscle volume and force generation were similar[22–24]. Almost all the muscles examined during the study exhibited higher peak activation levels during the LGJ. However, the area under the curve, or the integral, is crucial for understanding the differences between the two activities. The evident biomechanical differences between running and walking suggest that high-incline walking would result in more prolonged muscle activation due to its distinct gait pattern and slower strides, as the center of mass is lifted upwards.[27,32] While this prolonged activation did not lead to significant differences, the AUC trended higher during HIW for most lower-body muscles (BF, GM, VL, LG, SOL).

The stability of AUC during HIW suggests that the muscles were able to work efficiently and sub-maximally, as no evident signs of fatigue were observed in the collected data, aside from a few participants mentioning calf soreness during the incline walking. Since both activities were matched based on oxygen uptake, the differences observed between the two activities should be explained by the total activated muscle volume and the rate of force generation. In running, these two factors alone account for 98% of the increase in metabolic rate between different velocities.[23] When this is considered, it becomes more evident that the reduced contact time during LGJ results in a higher force output but a lower amount of total activated muscle. In HIW, there is a longer contact time; therefore, a lower rate of force generation and a higher amount of total activated muscle. This can be seen when comparing the AUC values with the peak EMG values; HIW trends are higher in AUC than in LGJ, but LGJ trends are higher in peak activation than in HIW.

These results align with similar studies. Tokuhiro et al documented an increase in AUC for the lower limbs as the gradient increased from 3° to 12°. Wall-Scheffler et al noted an increase in the duration of lower extremity muscle activation as the incline increased. While these studies support the findings of this study, it is still important to note the research limitations. Despite gathering muscle activation data for eight muscles, six of which were located in the lower body, many factors remain unaccounted for. One example is the muscles not captured due to equipment limitations and functionality restrictions; these other muscles could have illustrated a different story of work distribution between the two activities or helped better explain coordination differences at extreme inclines.

The different biomechanical demands placed on the body during LGJ can help explain the increase in AD and TA activation. Similarly to what was seen in peak activation, counterbalance movements are more extreme during running. Thus, the magnitude and duration of the activation were greater than those of HIW, where there was less sway due to the walking lunge-style strides.[31,33] The lower integral for the tibialis anterior muscle represents a decrease in sustained activation during tasks such as dorsiflexion and ankle control during the stance phase, and it could provide more insights into some of the task-specific demands and biomechanical variations that HIW poses.

Our study aimed to investigate the differences in muscle activation and resultant accelerations during level-grade jogging and high-incline walking locomotion, as well as how these differences changed over time. The findings revealed differences in peak activation and area under the curve for the eight muscles examined (TA, LG, SOL, VL, BF, GM, ES, AD), highlighting the distinct biomechanical demands of each mode. Most notably, the significant decrease over time in peak activation during LGJ for the LG, SOL, TA, GM, and AD, and a lack of decline for these muscles during the HIW.

There was a significant increase in resultant accelerations during LGJ over time, as well as when compared to HIW. This suggests that HIW might result in less orthopedic and musculoskeletal stress than LGJ when matched isocalorically. Longer trials and more kinematic data could help reveal whether these differences persist as submaximal exercise continues, and whether high-incline walking could be a safer alternative to traditional level-grade jogging. This study offers new insights into metrics related to uphill locomotion, emphasizing the biomechanical differences between two matched-intensity activities. The results may help refine exercise prescription and training design.

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