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Role of Occupational Footwear and Prolonged Walking on Lower Extremity Muscle Activation during Maximal Exertions and Postural Stability Tasks

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Abstract: *Background:* Occupational footwear and a prolonged duration of walking have been previously reported to play a role in maintaining postural stability. The purpose of this paper was to analyze the impact of three types of occupational footwear: the steel-toed work boot (ST), the tactical work boot (TB), and the low-top work shoe (LT) on previously unreported lower extremity muscle activity during postural stability tasks. *Methods:* Electromyography (EMG) muscle activity was measured from four lower extremity muscles (vastus medialis (VM), medial hamstrings (MH), tibialis anterior (TA), and medial gastrocnemius (MG) during maximal voluntary isometric contractions (MVIC) and during a sensory organization test (SOT) every 30 min over a 4 h simulated workload while wearing ST, TB, and LT footwear. The mean MVIC and the mean and percentage MVIC during each SOT condition from each muscle was analyzed individually using a repeated measures ANOVA at an alpha level of 0.05. *Results:* Significant differences ($p < 0.05$) were found for maximal exertions, but this was limited to only the time main effect. No significant differences existed for EMG measures during the SOT. *Conclusion:* The findings suggest that occupational footwear type does not influence lower extremity muscle activity during both MVIC and SOT. Significantly lower muscle activity during maximal exertions over the course of the 4 h workload was evident, which can be attributed to localized muscular fatigue, but this was not sufficient to impact muscle activity during postural stability tasks.

Keywords: postural stability; ergonomics; work boots; muscle activity

1. Introduction

Physical overexertion is one of the primary events or type of exposure that leads to injuries, illnesses, and fatalities in occupational settings. Physical exertions are a common occurrence due to the nature of the type and timing of the work performed, especially in settings such as construction, manufacturing, warehousing, mining, and manual material handling. Such physical exertions and physical demands in combination with the hazardous work conditions contribute to an increase in the risk of occupational falls [1], which are one of the leading causes of both non-fatal and fatal injuries at the workplace. The human postural control system is responsible for maintaining an upright, erect, and stable stance and works to prevent such falls and fall-related injuries. However, the sensory and motor systems of the postural control system can be impacted by several extrinsic (external)

and intrinsic (internal) factors. Several previous studies in the literature have reported the impact of extrinsic factors, such as occupational footwear and the impact of intrinsic factors such as muscular fatigue, on postural control and stability. Postural stability has been reported to be impacted by occupational footwear such as work boots [2–5], firefighter boots [6,7], and military boots [8–11] when exposed to various simulated occupational workloads. As such, muscular fatigue quantified by torque [6], muscle activity through electromyography (EMG) measures during postural stability tasks, but only after a fatiguing workload [10,12], perceived exertion [5,10,13], and increased energy expenditure [14] have been reported due to occupational footwear type and workload type. From the perspective of extrinsic factors, the negative impact on postural stability in the above-mentioned studies has been attributed to the design characteristics of these types of safety and protective footwear, as the design and material modifications in these types of occupational footwear may be more pronounced for the neuromuscular system to automatically correct and compensate. From the perspective of intrinsic factors, the negative impact on postural stability in the above-mentioned studies has been attributed to muscular exertions or fatigue, during which there is an impairment of the proprioceptive/somatosensory system's ability to control posture. Footwear design features play a critical part in maintaining postural stability and preventing falls. Some of them include: a heel-to-toe drop and heel height, where lower heel-to-toe drops and heel heights support the neutral position of ankle and foot aiding postural stability; a midsole and insole, where a thin and firm midsole and a textured insole aids proprioception and somatosensory feedback; footwear mass, where the lower the mass, the less energy is expended; boot shaft height, where the higher the height, the better the proprioception and somatosensory feedback supporting better postural stability [15]; and outer sole texture, thread and material, which impacts the slip-trip induced fall risk [16,17].

Most types of occupational safety footwear are designed in compliance with the standards put forth by the American National Standards Institute (ANSI-Z41-1991). The primary concern of occupational footwear relates to safety, and the design features and characteristics may not be suitable for optimal human biomechanics. With occupational footwear and workload significantly influencing postural stability as well as physiological and muscular exertion, the current research team have previously conducted multiple studies and analyzed the impact of three types of work shoes (steel-toed work boots—ST; tactical work boots—TB; low-top slip resistant shoes—LT) on postural stability, muscular exertion, and energy expenditure when exposed to different physiological workloads. One of the first studies from the current research team reported the impact of these three types of occupational footwear on postural stability assessed using the sensory organization test (SOT) on the NeuroCom Equitest when exposed to prolonged durations (4 h) of walking with postural stability measured every 30 min. Postural stability performance was quantified using center of pressure (COP) excursions-derived postural sway parameters during the SOT tests. The use of LT compared to ST and TB was found to be more detrimental to postural stability, for which the design of the footwear, such as the absence of an above-ankle boot shaft despite its lower mass, was suggested as rationale for the findings [2]. Additionally, postural stability was found to be lower over time, with continuous standing and walking attributed due to the workload placed on the postural control system [2]. The secondary analysis from the same study using SOT equilibrium (EQ) scores reported similar findings in footwear differences, with LT demonstrating poor postural stability. However, there were no significant differences over time, suggesting that SOT EQ scores alone might not be sufficient to detect postural stability changes [4], especially since COP postural sway parameters identified differences over the 4 h workload [2]. While postural stability measures have already been reported, lower extremity EMG muscle activity from the same study during maximal exertions and during SOT postural stability tasks have not been analyzed or reported yet. The EMG analysis will further aid in the understanding of the behavior of these types of occupational footwear when exposed to simulated occupational workloads.

A second follow-up study using the same three types of footwear with the addition of a barefoot condition analyzed postural stability during dynamic postural perturbations using the motor control test (MCT) on the NeuroCom Equitest. COP latencies in response to postural perturbations as well as EMG measures were analyzed. Even with no imposed physical workload, TB and ST demonstrated significantly lower muscle activity compared to LT, suggesting the need for only minimal muscular effort and adjustments to maintain postural stability [3]. After addressing the impact of these types of occupational footwear on postural stability with an extended duration low-intensity workload and no workload conditions, a third follow-up study analyzed the impact of these types of occupational footwear (ST and TB) under an acute high-intensity workload on both muscular exertions and postural stability. A decreased pain threshold immediately after the workload and a decreased EMG MVIC 15 min post-workload were reported as consequences of the acute muscle soreness and fatigue, but no footwear differences were seen [13]. However, postural stability quantified by the COP postural sway during a modified clinical test of sensory interaction on balance (CTSIB-M), showed that donning the ST led to a better postural stability compared to TB, but only in the balance conditions that had no visual and altered somatosensory feedback [5]. When visual and somatosensory feedback were optimally available, there were no differences between the types of footwear. The design features on the ST, with a thin-hard mid-sole, an elevated boot shaft, and a low heel height were suggested as positive design characteristics to aid postural stability [5]. Finally, during a fourth study, ST and TB were compared for physiological energy expenditure during simulated workloads that included horizontal and graded treadmill walking–running protocols [14]. The ST was reported to cause a significant increase in oxygen consumption quantified by a significant increase in absolute oxygen consumption (VO_2) [14].

In addition to the current research team's findings on the biomechanical impact of work boots, another team of researchers from the University of Wollongong, Australia have addressed the biomechanical and psychological impact of mining work boots [18–23]. Additionally, similar to the current study, previous research has used both postural sway and EMG measures to compare footwear types in standing tasks (barefoot, stable control shoes, and unstable shoes) [24]. Unstable shoes were reported to cause an increase in postural sway as well as an increase in muscle activity among the lower extremity muscles [24]. As such, there are emerging studies within the literature that have aided understanding of the behavior and the design suggestions for work boots, with more relevant future research warranted. Accordingly, there is still a dearth of literature on the impact of occupational footwear on lower extremity muscle activity, especially when exposed to long durations of low-intensity workloads similar to the predominant type of assembly line occupations that require workers to stand and walk continuously for long shifts wearing occupational footwear. Therefore, the purpose of this paper is to analyze previously unreported EMG muscle activity data from the first occupational footwear 4 h standing/walking study during maximal exertions and postural stability tasks when exposed to a low-intensity 4 h prolonged duration simulated workload while wearing three different types of occupational footwear (ST, TB, LT). Based on the previous literature, it was hypothesized that EMG muscle activity during maximal exertions will be lower, and during postural stability tasks EMG muscle activity will be less efficient over the 4 h prolonged duration standing/walking and between the types of footwear.

2. Materials and Methods

2.1. Participants and Study Design

The study is an original research study using a repeated measures study design with randomized footwear assignment. The current paper reports previously unreported EMG analysis from the occupational footwear 4 h walking study. Fourteen healthy adult males (age: 23.6 ± 1.2 years; height: 181 ± 5.3 cm; mass: 89.2 ± 14.6 kg) with no history of any orthopedic or neurological disorders completed the study in a repeated measures design. Additionally, these participants did not have any prior experience in physically

exerting occupations, which is one of the limitations of the study and is addressed in the limitations section. Each participant was tested in three separate testing sessions with all three types of occupational footwear (ST, TB, LT) (Figure 1) when exposed to a low intensity simulated occupational workload, separated by a minimum of 72 h. All participants read and signed the informed consent to participate in the study based on the University of Mississippi's institutional review board (IRB) approved protocol (IRB Protocol# 11-150) and completed a brief familiarization protocol with the balance and lower body muscular exertion experimental protocol.

Figure 1. Three types of occupational footwear: **Left** (top and bottom)—Steel-toed work boots (ST), **Middle** (top and bottom)—Tactical work boot (TB) and **Right** (top and bottom)—Low-top slip resistant shoe (LT).

2.2. Experimental Procedures

During each testing session, after a brief initial dynamic warm-up, participants were provided with one type of the occupational footwear that was based on a randomized, but foot size matched assignment. Participants were then prepared for the EMG measurement by skin abrasion and alcohol cleaning to lower skin impedance, after which bipolar surface electrodes were placed on the muscle bellies of the vastus medialis (VM), the medial hamstrings (MH), the tibialis anterior (TA), and the medial gastrocnemius (MG) with the ground electrode placed on the tibial tuberosity using a surface EMG for the non-invasive assessment of muscles (SENIAM) guidelines and the recommendations from De Luca's "the use of surface EMG in biomechanics" for electrode orientation and placement [25,26]. Participants then performed three trials of maximal voluntary isometric contraction (MVIC) for 5 s for each of the above four muscles. During VM and MH MVIC, participants were seated on an exercise bench with a footrest and during TA and MG MVIC, participants were in a standing position with the forefoot fixed by a foot strap to ensure the MVIC trials were performed in the middle of the range of motion for both knee and ankle joints. EMG data were collected using the Noraxon Telemyo T2400 G2 wireless EMG system (Noraxon Inc., Scottsdale, AZ, USA) with a sampling frequency of 1500 Hz. On completion of the MVIC trials, participants stepped on the NeuroCom Equitest (NeuroCom International Inc., Clackamas, OR, USA) and performed the first four conditions of the SOT as a pre-test measure (PRE). The four SOT conditions included eyes open—EO; eyes closed—EC; eyes open sway referenced vision—EOSRV and eyes open sway referenced platform—EOSRP, as they best represented the reliance on visual and somatosensory for maintaining postural stability. The triggering starting and stopping of the EMG measures during the SOT trials was accomplished by a wireless sync between the NeuroCom Equitest and the Noraxon through a transistor–transistor logic (TTL) pulse.

On completion of the PRE-testing, participants walked for a total of 4 h (240 min) at a self-selected pace and a self-selected path on an even flat vinyl flooring. Participants were allowed to stand briefly in one place for a few minutes during the 4 h walking but were never allowed to sit or rest. During this 4 h walking period, MVIC and SOT trials were repeated for eight more times in 30 min intervals. Thus, a total of 9 testing measures were performed (PRE, 30 min, 60 min, 90 min, 120 min, 150 min, 180 min, 210 min, 240 min). The end of the 240th minute of testing marked the completion of the first testing session in one of the randomly assigned types of occupational footwear. This method of standing/walking for a prolonged duration and testing in 30 min intervals was adopted from previous studies attempting to simulate occupational workloads in specific occupations such as roofers and railroad workers [27,28]. The exact same procedures were repeated for the other two randomly assigned types of footwear with a minimum of 72 h of rest in between testing sessions to avoid any undue muscular fatigue.

2.3. Data and Statistical Analysis

Raw EMG data from all four lower extremity muscles (VM, MH, TA, MG) during both MVIC and SOT trials were filtered using a band pass filter at 20–250 Hz and a full wave rectification was performed before further analysis. Mean muscle activity was calculated for all three trials of MVIC and three trials of each SOT testing from which an average of the three trials was used as an EMG dependent variable. Additionally, the percentage activation of MVIC (%MVIC) during all trials of the SOT tests was calculated normalized to the MVIC. MVIC dependent variables included mean VM MVIC, mean MH MVIC, mean TA MVIC, and mean MG MVIC during all nine time points of testing. SOT postural stability-dependent variables included both mean muscle activity (mean VM, mean MH, mean TA, mean MG) and %MVIC (%MVIC VM, %MVIC MH, %MVIC TA, %MVIC MG) for each SOT testing condition during all nine time points of testing. All the above-mentioned EMG-dependent variables were analyzed using a 3 (footwear—ST, TB, LT) $\times$ 9 (time points—PRE, 30 min, 60 min, 90 min, 120 min, 150 min, 180 min, 210 min, 240 min) repeated measures analysis of variance (RM-ANOVA) individually for each SOT testing condition (EO, EC, EOSRV, EOSRP). If a significant main effect for footwear or time or a significant interaction was found, it was followed up with post-hoc pairwise comparisons or simple effects comparisons, respectively, using a Bonferroni correction. Effect size calculations were also performed with partial eta squared (ηp^2) for RM-ANOVA-derived main effects and Hedges' g for pairwise comparisons for the significant main effect differences. The interpretation of ηp^2 was based on small ($\eta p^2 = 0.01$ to <0.06), medium ($\eta p^2 = 0.06$ to <0.14), and large ($\eta p^2 > 0.14$) effect sizes, while the interpretation of Hedges' g was based on trivial (Hedges' g < 0.2), small (Hedges' g = 0.2 to <0.5), medium (Hedges' g = 0.5 to 0.8), and large (Hedges' g > 0.8) effect sizes, respectively. The alpha level was set at $p < 0.05$ and all statistical analyses were performed using SPSS v26 (IBM SPSS, Armonk, NY, USA).

3. Results

The RM-ANOVA revealed significant differences for the MVIC EMG measures but was limited to only the time main effect with no significant differences between the types of occupational footwear. A time main effect significant difference was identified for the mean VM MVIC [$F_{(8, 104)} = 4.113$; $p < 0.0001$, $\eta p^2 = 0.240$] (Figure 2); for the mean MH MVIC ($F_{(8, 104)} = 2.770$; $p = 0.008$, $\eta p^2 = 0.176$) (Figure 3), and for the mean MG MVIC ($F_{(8, 104)} = 2.453$; $p = 0.018$, $\eta p^2 = 0.159$) (Figure 4). No significant differences were evident in TA during MVIC. The post-hoc pairwise comparisons for the mean VM MVIC revealed greater mean muscle activity for 30 min compared to 60 min (Hedges' g = 0.26), and 240 min (Hedges' g = 0.36) measures. For the mean MH MVIC, the pairwise comparisons did not reveal any significant differences and the mean MG MVIC 30 min had greater mean muscle activity than 240 min (Hedges' g = 0.38) measures. Additionally, there were no significant differences for the main effects of footwear and time and no significant differences for

footwear and time interaction for VM, MH, TA, and MG during the SOT postural tasks. An example of the EMG during the postural stability tasks is provided in Figure 5 for the mean MG during the EC condition of the SOT. Although the mean muscle activity based on descriptive statistics was increasing over time on average, none of the changes were significantly different between the footwear types and between the different time points of testing. Complete data from the study for the mean muscle activity during MVIC and the mean and %MVIC muscle activity during all four SOT conditions, for all four muscles, for all three types of occupational footwear during all nine time points of testing are presented as a supplemental file as Tables S1–S9.
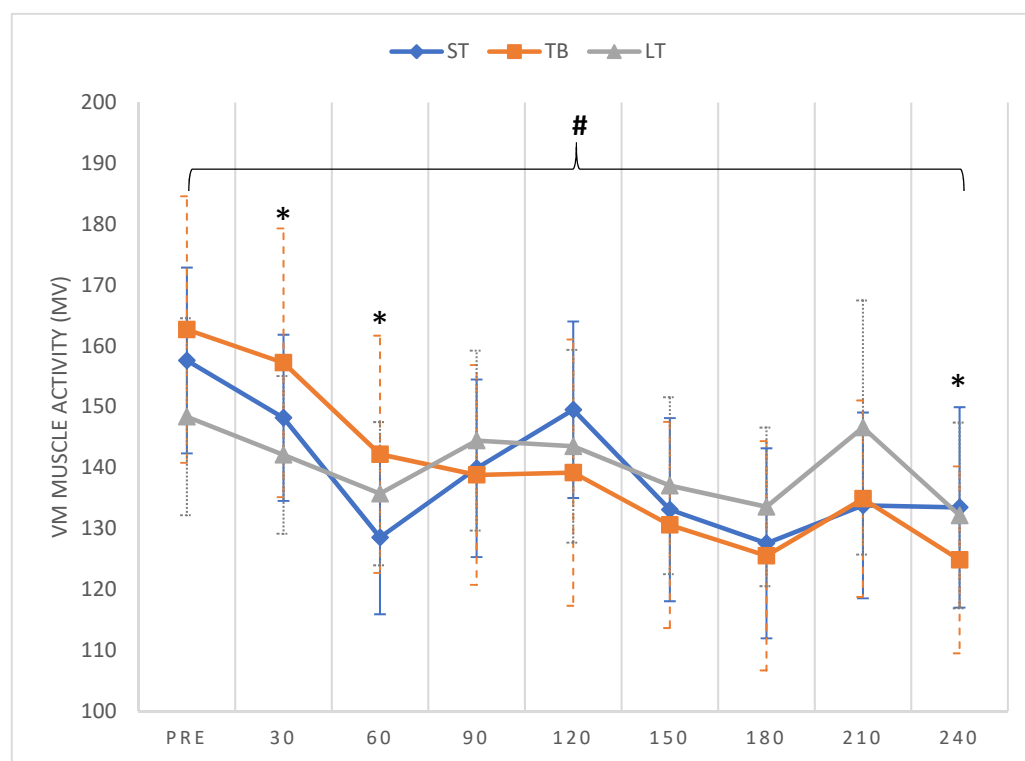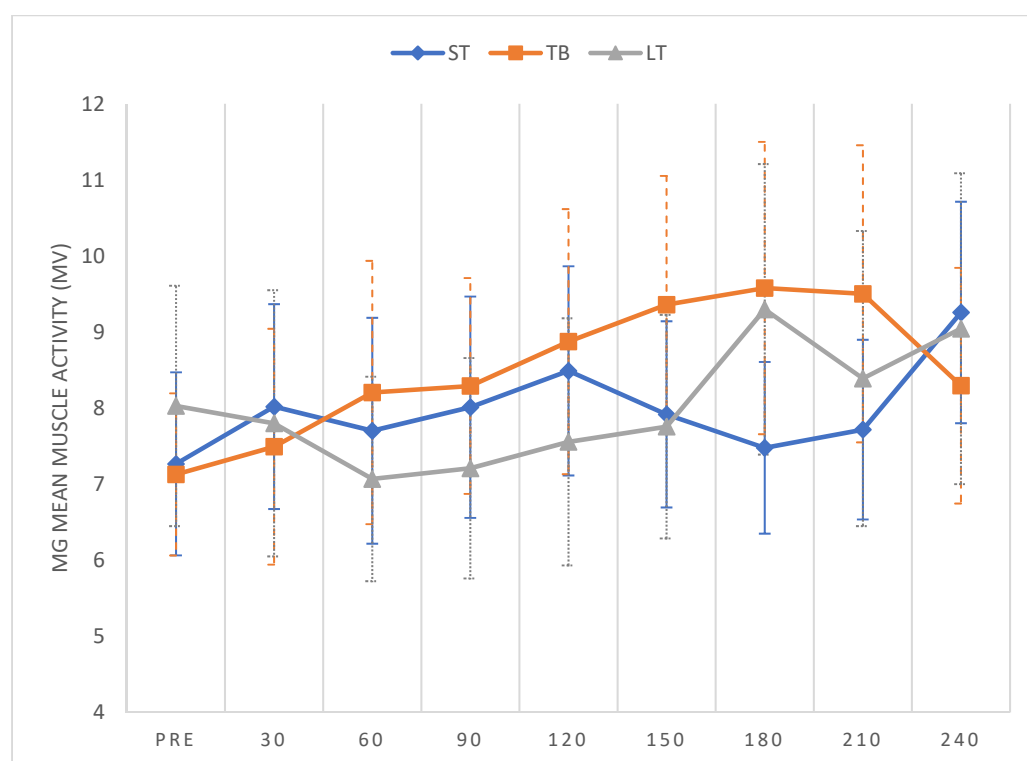

Figure 2. Mean muscle activity for vastus medialis (VM) for three types of occupational footwear across 4 h (PRE to 240th minute) during maximal voluntary isometric contractions (MVICs). # represents the significant main effect for time and * represents significant differences in pairwise comparisons at $p < 0.05$ level. Bars represent standard errors.
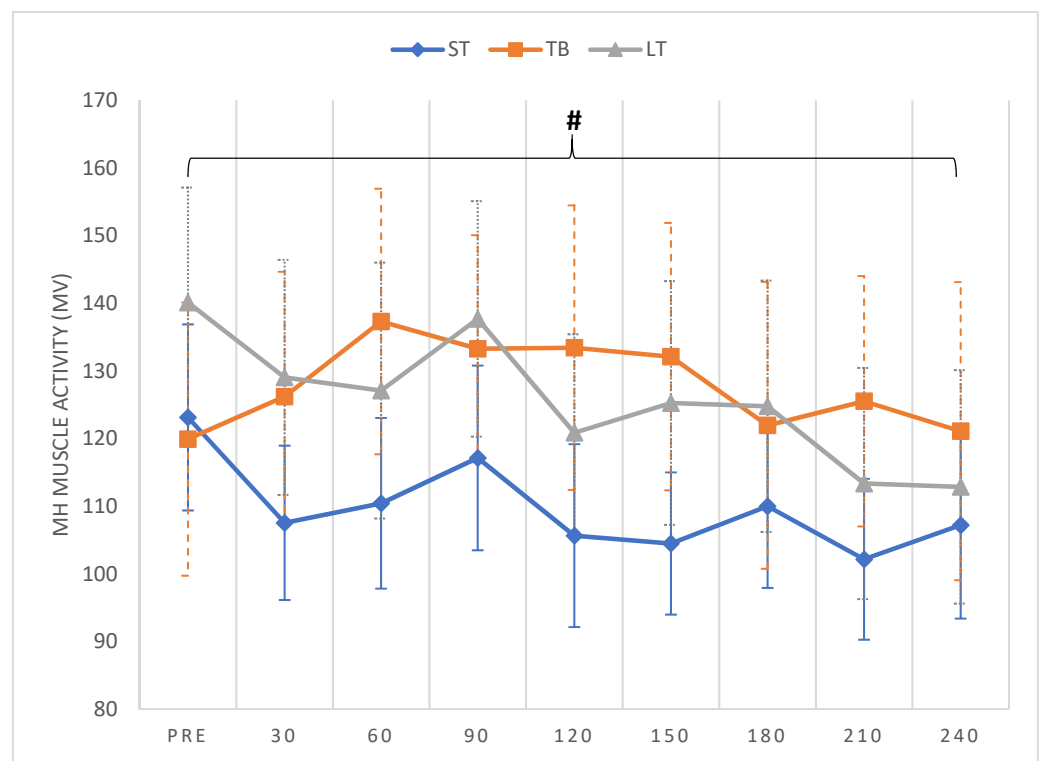
Figure 3. Mean muscle activity for medial hamstrings (MH) for three types of occupational footwear across 4 h (PRE to 240th minute) during maximal voluntary isometric contractions (MVICs). # represents significant main effect for time at $p < 0.05$ level. Bars represent standard errors.
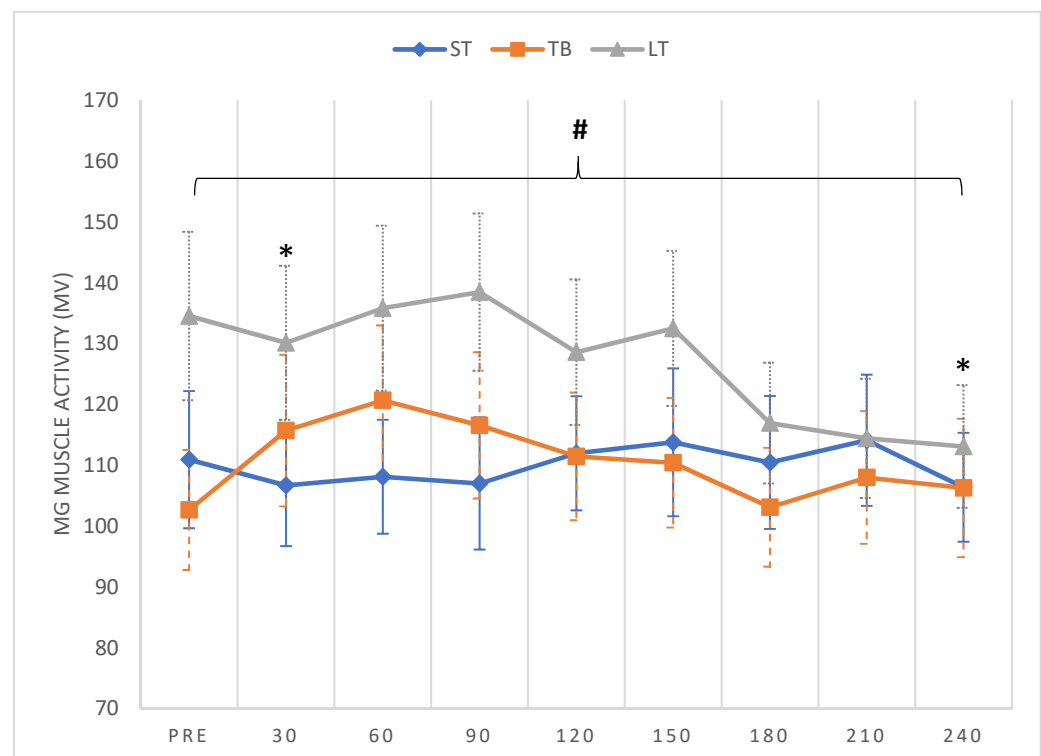
Figure 4. Mean muscle activity for medial gastrocnemius (MG) for three types of occupational footwear across 4 h (PRE to 240th minute) during maximal voluntary isometric contractions (MVICs). # represents significant main effect for time and * represents significant differences in pairwise comparisons at $p < 0.05$ level. Bars represent standard errors.

Figure 5. Mean muscle activity for medial gastrocnemius (MG) for three types of occupational footwear across 4 h (PRE to 240th minute) during the eyes closed (EC) condition of the sensory organization test (SOT). Bars represent standard errors.

4. Discussion

The purpose of this paper was to analyze previously unreported EMG muscle activity from four lower extremity muscles (vastus medialis (VM), medial hamstrings (MH), tibialis anterior (TA), and medial gastrocnemius (MG) during both maximal excursions and during postural stability tasks when exposed to three different types of occupational footwear and a low-intensity simulated occupational workload that involved 4 h of walking. Based on the current findings, the occupational footwear did not play a role in influencing EMG mean muscle activity during maximal exertions and was only influenced by the low-intensity workload with significant differences in knee extensors, flexors, and ankle plantar flexors. The occupational footwear and the low-intensity workload did not impact mean muscle activity and the percentage of MVIC muscle activation during SOT postural stability tasks.

In regards to the extrinsic factor of footwear, the current findings suggest that the design differences in occupational footwear, while they have been shown previously to impact postural stability [2], did not influence muscle activity both during MVICs and during the SOT. The footwear design features such as the heavier mass on ST compared to TB and LT, the elevated boot shaft on ST and TB compared to LT, the heel height, the mid-sole thickness etc., were all attributed to the differences previously seen in postural stability [2–5,29–32]. The increased proprioceptive and somatosensory feedback due to the elevated boot-shaft by providing support around the ankle joint and minimizing postural sway [32], the firm hard mid-sole on the ST and TB aiding greater somatosensory feedback to minimize postural sway [31], and the lower heel height creating an optimal stable foot–ankle position on the ST and TB have all been identified as possible favorable design features for maintaining optimal postural stability during SOT and MCT [2,3] and efficient EMG muscle activity when subjected to postural perturbations [3]. However, muscle activity during MVICs and SOT did not reveal any significant differences between the types of footwear, where the low-intensity prolonged duration walking task may not have been sufficient to cause significant changes in MVICs. In addition, the positive design

features of the occupational footwear may have contributed to the better maintenance of postural stability, without a significant increase in muscle activity. Even though the differences were not statistically significant, the muscle activity of MG for LT during MVIC was greater compared to the other two types of footwear, for which the lower, below-ankle boot shaft height might be the reason for the increased muscle activity in MG due to the restrictions in the range of motion and the position held during the MVIC impacted by the boot shaft height [3]. The EMG muscle activity differences seen in these types of footwear were reported previously during MCT dynamic postural perturbations [3]. It appears that the significantly lower and more efficient muscle activity especially in TB reported during MCT previously [3] did not transfer over during SOT in the current analysis. The MCT's dynamic platform movements probably elicited more footwear differences than the more static SOT. However, combing the previously reported postural stability data from the SOT [2] and the current EMG findings, the footwear design differences play a role in postural stability, but these postural stability changes were not accompanied by significant changes in muscle activity. When exposed to horizontal and graded treadmill walking–running workloads, the same types of occupational footwear showed differences in energy expenditure, with ST exhibiting a greater absolute VO_2 compared to TB [14] but the EMG did not show significant differences between the types of footwear during maximal exertions when exposed to an acute treadmill Bruce protocol [13]. As such, EMG measures alone may not be sufficient to identify differences in these types of occupational footwear. Additionally, footwear type does not always have an impact on lower extremity muscle activity [13] and even if it does, the impact may predominantly be on the smaller intrinsic lower extremity muscles, rather than the larger extrinsic muscles [24]. As such, there is a further need for biomechanical and physiological measures to be assessed and analyzed simultaneously in order to have a comprehensive understanding of their behavior.

In regard to the intrinsic factor of the workload, the current findings suggest that the simulated low-intensity occupational workload induced lower muscle activity in knee flexors, knee extensors, and ankle plantar flexors only during maximal exertions but was not sufficient enough to elicit significant changes in muscle activity during postural stability tasks. Muscular fatigue resulting from a strenuous physical activity has been related to the deterioration of the ability of muscles to produce and sustain a required output of muscle activity [33]. This is true, especially when performing maximal exertions during MVIC, thus providing a rationale for the significantly lower muscle activity seen at the end of the 4 h duration. Although a time main effect was present and the mean muscle activity decreased over time, pairwise comparisons identified only minimal significant differences across time, and the results need to be interpreted with caution as there was not a linear, significant decline in muscle activity over time. This could be due to the nature of the workload performed at a self-selected pace and path of standing and walking for 4 h, but this study did not analyze the direct workload with physiological measures. Muscular fatigue has also been shown to reduce the activity of the proprioceptive system [33,34] resulting in an abated ability to maintain postural equilibrium [35–37]. However, there were no significant differences identified during any SOT condition, suggesting there was not a greater requirement for muscle activation over the 4 h, even when the sensory systems of vision and proprioception were either absent (EC) or distorted (EOSRV and EOSRP). The control of posture is a complex but efficient interaction between sensory and motor processes, and in an erect stance, only a relatively small amount of muscular activity is required [38]. However, with the addition of a physical workload, increased effort from the lower extremity in maintaining balance is required. This increase in EMG muscle activity during a static stance was seen in the heavier standard military boot compared to the lighter tactical military boot only after an acute heavy-intensity load carrying workload [10]. However, the current workload was a low-intensity occupational simulation and did not impact muscle activity both between the types of footwear and over time across the 4 h duration. The negative impact of strenuous physical workload on postural stability has predominantly been based on acute high-intensity workloads that

usually involve greater than 50% of maximal voluntary contractions and greater than 33% of maximal aerobic capacity; however, occupational workloads similar to the current study where muscular fatigue is induced over a prolonged duration at a low rate of less than 15% of maximal voluntary contractions and less than 33% of maximal aerobic capacity [39], provide a rationale for the observed findings in the current study. Even though muscle activity during postural stability tasks was not influenced by the low-intensity workload, previous findings from the current study reported decrements in postural stability over the 4 h duration [2], suggesting that postural stability can change with any significant difference in muscle activity.

There are several limitations to this study. As previously mentioned, the study did not directly measure energy expenditure over the 4 h walking period as the only directions given to the participants were to walk at a self-selected pace and path, with brief periods of standing allowed. However, they were not allowed to sit or rest throughout the entire testing period. This was in order to mimic an occupational workload. However, the study did not have any added cognitive or manual handling tasks to the physical workload. Additionally, this current study focused on larger extrinsic muscles of the lower extremity rather than smaller intrinsic muscles, which have been previously reported to be impacted by footwear type during postural stability tasks [24]. Hence, future research should also focus on analyzing the impact of footwear and prolonged standing/walking on the lower extremity intrinsic musculature. Finally, healthy college-aged individuals served as the participants and may not best represent the entire working population, but they at least represented the amateur workforce. Members of the healthy, young population free of any neuromuscular or musculoskeletal abnormalities were chosen so any significant differences seen could be attributed to the footwear and/or the workload. Previous research has also reported both sensory and mechanical effects due to footwear types with alterations in muscle activity, joint kinematics, and kinetics during gait [40,41]. As such, future research combining biomechanical and physiological measures under these prolonged durations of standing and walking tasks is warranted, which can be critical for predicting injury risk.

5. Conclusions

The findings from the current analysis suggest that occupational footwear type does not influence lower extremity muscle activity both during maximal exertions and during postural stability tasks. The 4 h occupational workload only influenced lower extremity muscle activity during maximal exertions, leading to significantly lower muscle activity over the time course of the workload, but it did not impact the muscle activity during postural stability tasks. Based on previous findings from the same study, both occupational footwear and the prolonged durations of standing and walking task have been reported to impact postural stability measures during the SOT. Contrary to postural stability findings, the footwear design characteristics did not appear to play a role in lower extremity muscle activity, suggesting the requirement of assessment methods other than EMG measures to identify the overall behavior of these types of occupational footwear when exposed to a prolonged duration occupational workload. The fact that the 4 h occupational workload was lower in intensity than other strenuous workloads did not appear to cause sufficient localized neuromuscular fatigue to influence muscle activity during postural stability tasks but caused significant decrements in muscle activity during maximal exertions. The current findings add to the body of literature of occupational footwear biomechanics and offer design suggestions for occupational footwear when exposed to simulated occupational workloads.

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