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Lower Limb Muscle Activation in Young Adults Walking in Water and on Land

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Abstract: Previous research has shown that exercise interventions requiring increased activation of the tibialis anterior (TA), the primary ankle dorsiflexor, can improve walking performance in individuals with foot drop. Correspondingly, heightened drag forces experienced during walking performed in water may augment TA activation during the swing phase of gait, potentially leading to improved walking gait on land. Therefore, this study aimed to compare surface electromyographic (sEMG) activation in the TA and medial gastrocnemius (GM) during gait performed in water versus on land. Thirty-eight healthy, recreationally active young adults, comprising 18 females and 20 males, participated in the study. Each participant completed 2 min walking trials under five conditions: land 2.5 mph, land 3.5 mph, water 2.5 mph, water 3.5 mph, and water 3.5 mph with added jet resistance. Stride kinematics were collected using 2-dimensional underwater motion capture. TA and GM, muscle activation magnitudes, were quantified using sEMG root-mean-square (RMS) amplitudes for both the swing and stance phases of walking. Additionally, TA and GM co-activation (Co-A) indices were estimated. Two-way within-subjects repeated measures analyses of variance were used to evaluate the main effects of and interactions between the environment and walking speed. Additionally, paired sample t-tests were conducted as a secondary analysis to investigate differences between walking in water at 3.5 mph with and without added jet resistance. Main effects and interactions were observed across various stride kinematics and sEMG measures. Notably, TA sEMG RMS during the swing phase of walking gait performed at 2.5 mph was 15% greater in water than on land ($p < 0.001$). This effect increased when walking gait was performed at 3.5 mph (94%; $p < 0.001$) and when jet resistance was added to the 3.5 mph condition (52%; $p < 0.001$). Furthermore, TA Co-A was increased during the stance phase of gait in water compared to on land ($p < 0.001$), while GM Co-A was reduced during the swing phase ($p < 0.001$). The findings of this study offer compelling evidence supporting the efficacy of aquatic treadmill walking as a potential treatment for individuals suffering from foot drop. However, further research is needed to evaluate whether a causal relationship exists between heightened TA activation observed during aquatic treadmill walking and improvements in voluntary dorsiflexion during gait.

Keywords: electromyography; aquatic; treadmill; foot drop

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1. Introduction

Foot drop, prevalent in various neurological disorders, significantly heightens the risk of falls and diminishes individuals' physical independence [1–4]. This condition, marked by a decreased capacity to dorsiflex the foot during walking, results in a dragging or slapping motion of the foot against the ground. Foot drop is typically attributed to weakness in ankle dorsiflexors or over-activation in ankle plantar flexors [3]. Consequently, individuals affected by foot drop often exhibit slower walking speeds and encounter challenges navigating uneven terrain like stairs [2–4]. Compensatory mechanisms, such as hyper-flexion of the knee and hip joints (e.g., "steppage gait" and "hip hiking"), can lead to improper skeletal loading and injury over time [5,6]. Furthermore, a foot drop may restrict activities of daily living, contributing to a decline in functional independence and overall quality of life [1].

Foot drops can stem from various neurological conditions affecting the central or peripheral nervous systems, including stroke, multiple sclerosis, cerebral palsy, spinal stenosis, peripheral neuropathy, Charcot–Marie–Tooth disease, or localized compression of the peroneal nerve [7]. Traumatic events such as spinal cord injuries, fibular fractures, or knee dislocations can also lead to foot drop [8]. Additionally, it may develop due to surgical positioning or prolonged habits like sitting with legs crossed [2]. Although the incidence of foot drop is not precisely documented, it is evident that it affects diverse populations. Given the varied causes and affected populations, it is important to explore a range of treatment options to accommodate individuals with different levels of mobility.

Common treatment options include nerve decompression surgery, ankle–foot orthoses, and functional electrical stimulation [2–4]. Surgical interventions are invasive and primarily target peripheral causes like peroneal nerve compression or direct trauma (e.g., fracture, dislocation, etc.) [2]. While ankle–foot orthoses effectively enhance ambulatory function, they may limit ankle mobility, potentially causing discomfort and muscle contracture over time [9,10]. In contrast, functional electrical stimulation enables individuals with foot drops to move the affected ankle through its entire range of motion by activating the tibialis anterior (TA), the primary muscle responsible for dorsiflexion, using external electrical stimuli applied to the peroneal nerve. Research indicates that functional electrical stimulation has an immediate "orthotic effect" on walking performance, increasing toe clearance and walking speed while the device is in use [3,9,10]. Moreover, it may have lasting effects on TA excitability by strengthening corticospinal connections [11]. However, functional electrical stimulation poses challenges like difficulty in device application and skin irritation from electrodes [12,13]. Further research is needed to determine if voluntary control of the TA can be improved without resorting to surgery or uncomfortable assistive devices.

Gait training stands out as a promising non-invasive approach to enhance dorsiflexor function. In a study by Willerslev-Olsen et al. [14], children with cerebral palsy exhibited a significant increase in maximal voluntary dorsiflexion torque and toe lift at the end of the swing phase after daily incline treadmill walking for 30 days. The authors proposed that uphill walking required greater voluntary activation of the TA during the swing phase, facilitating toe lift in preparation for foot strike. The gait training also induced significant shifts in corticospinal drive, evident from increased intramuscular coherence in the beta and gamma frequency bands recorded from the TA, which may have contributed to the observed improvements in gait mechanics. A similar study in spinal cord injury patients using transcranial magnetic stimulation found increased TA excitability after a 16-week, body weight-supported treadmill walking intervention [15]. Moreover, improvements in corticospinal tract function exhibited a significant positive correlation with improvements in walking function, as assessed by the Walking Index for Spinal Cord Injury II (WISCI II). While the precise adaptation mechanism remains unclear, the authors propose that rigorous daily treadmill training involving voluntary activation of leg muscles may enhance neural control of the TA and improve toe clearance during the swing phase of walking.

For individuals with limited mobility, engaging in intensive treadmill training on land may be impractical [16,17]. Conversely, aquatic treadmill walking is considered a safe alternative that enhances movement confidence and accommodates various levels of functional capacity [18,19]. Water immersion introduces an upward buoyant force, offloading body weight and reducing the impact of potential falls [20]. Additionally, a review focused on individuals with Parkinson's disease noted a decrease in fear of falling during exercise performed in an aquatic environment [21]. Aside from Parkinson's disease, prior research supports aquatic exercise as an effective intervention for addressing motor symptoms in individuals with various other neurological diseases, including stroke, multiple sclerosis, dementia, and cerebral palsy [22]. In essence, water provides a secure environment for individuals with limited mobility to exercise with increased confidence.

Moreover, the heightened drag forces associated with walking in water may demand increased voluntary activation of the TA compared to walking on land. In both aquatic and dry land environments, fluid drag forces act on the body to impede movement. However, the density of the fluid, in this case, water, significantly influences the magnitude of the fluid drag force. Given the higher density of water compared to air, an individual experiences greater resistance to motion in water than on land [23]. Additionally, fluid drag force intensifies with the relative velocity of the object in the fluid. During the swing phase of the walking stride cycle, the foot attains a higher linear velocity relative to more proximal lower-extremity body segments, akin to the tip of a windmill. This elevated velocity increases fluid drag, particularly at the anterior foot and ankle. Consequently, walking on an aquatic treadmill should necessitate greater TA activation to overcome heightened drag forces at the distal lower limb and lift the toes before foot strikes. Further, as indicated by Willerslev-Olsen et al. [14], the repetitive activation of lower-limb muscles during exercises that demand heightened TA activation may enhance neural control of the ankle by strengthening corticospinal connections.

Results from previous studies examining differences in TA muscle activation between aquatic and land treadmill gaits are mixed. For instance, in studies where walking in water was performed at slower speeds compared to land or at self-selected speeds, no significant differences in peak or average TA activation have been reported [24–27]. In a study by So et al. [28] investigating running gait while controlling for stride rate across participants, significantly greater TA activation was observed for running in water than on land, with deeper water immersion depths also leading to greater activation. However, it is noteworthy that a significant difference was observed only in the left leg, with no differences noted in the right leg. Moreover, Silvers et al. [29] observed an increase in the absolute duration of activity and total TA activation between water and land running at matched speeds. However, the study did not find a significant difference in TA activation when expressed as a percentage of maximal voluntary contraction between the two environments. It is important to note that in the study, muscle activation measures were derived by averaging data from 10 full stride cycles without separately analyzing the swing and stance phases. Additionally, it is important to consider that findings from investigations focusing on muscle activation during running may not directly apply to walking gait. For instance, during the swing phase of running, the knee and hip exhibit greater flexion to minimize the moment arm of the leg, facilitating a faster swing phase. This altered posture may lead to reduced drag forces opposing dorsiflexion, which contrasts with the more fully extended leg position typically observed during the swing phase of walking.

The observation of greater TA activation during running at matched stride rates and speeds does, however, reflect a current need for investigation into TA activation during walking performed in water and on land at matched speeds. Moreover, there is a scarcity of research on plantar flexor (i.e., gastrocnemius) activation, and, to our knowledge, no prior study has provided insights into muscle co-activation (Co-A) between the TA and medial gastrocnemius (GM) during walking in water. Improving the ability to dorsiflex the foot involves either enhancing TA activation or minimizing GM Co-A. Consequently, it is crucial to assess the activation of both muscles. To our knowledge, no study has compared distal leg muscle activation patterns during both the stance and swing phases of gait between aquatic and land treadmill walking. This represents a notable gap in the current literature.

The purpose of this study was to compare stride kinematics and surface electromyography (sEMG) of the TA and GM in young adults performing aquatic and land treadmill walking at matched speeds. We hypothesized that (1) TA activation would be greater when walking in water than on land at matched speeds, attributed to increased fluid drag opposing the forward movement of the distal lower limb. We anticipated (2) no significant differences in the magnitude of TA activation between walking environments during the stance phase at matched speeds. Additionally, we hypothesized that (3) GM activation would be greater during aquatic walking than on land during the stance phase. This expectation stemmed from the assumption that the GM would play a more substantial role in propelling the body forward in the presence of increased fluid drag. However, we also anticipated (4) no significant difference in GM activation between environments during the swing phase. Given the relationship between relative velocity and fluid drag, we further hypothesized that (5) observed differences in muscle activation magnitude between aquatic and land walking would be more pronounced at the 3.5 mph speed condition than the 2.5 mph speed condition. The findings of this study may offer valuable insights into how the aquatic environment influences the activation of muscles spanning the ankle joint during walking and provide supporting evidence for including aquatic exercise as a component of treatment for foot drop.

2. Materials and Methods

2.1. Participants

An a priori power analysis was conducted using G*Power (version 3.1; $f = 0.25$, $\alpha = 0.05$, $\beta = 0.80$). The analysis revealed that a minimum sample size of 34 participants would be required to detect a moderate effect size of $f = 0.25$. Consequently, we opted to recruit a convenience sample of 38 healthy, recreationally active adults from the local community (Table 1). In this study, we defined recreationally active as being able to walk continuously for a minimum of 20 min without the use of an assistive device. To ensure eligibility for inclusion, participants were required to be between the ages of 18 and 35 years old and have no history of neurological diseases with motor symptoms (e.g., stroke, multiple sclerosis, recent concussion, etc.). Additionally, participants were asked to confirm that they were not currently experiencing any physical discomfort or injury that could impact their ability to walk. Furthermore, individuals who had undergone surgical interventions on their lower limbs or trunk within the past two years or had previously experienced hip, knee, or ankle ligament tears were excluded from the study. Participants were recruited voluntarily, and prior to participation, they provided written consent via signature after reviewing an informed consent document approved by the university Institutional Review Board.

Table 1. Participant characteristics.

Sex	*n*	Age (years)	Height (cm)	Body Mass (kg)
Collapsed	38	22.6 (2.2)	172.5 (8.5)	72.4 (14.9)
Female	18	21.7 (1.8)	167.6 (6.9)	65.4 (12.9)
Male	20	23.4 (2.2)	177.0 (7.2)	78.8 (14.0)

Data are reported as mean (SD).

2.2. Procedures

Participants attended two separate sessions at a movement research clinic. In the initial session, they became familiar with the study protocol. During the familiarization process, participants underwent anthropometric measurements, including height, weight, and the determination of leg dominance. Leg dominance was assessed by asking participants which leg they preferred to kick a ball [30]. Additionally, participants practiced walking on the aquatic treadmill to acquaint themselves with the equipment and walking conditions. The aquatic treadmill familiarization consisted of walking for 4 min at a moderate 2.5 mph pace. During the aquatic walking, participants were given the following standard verbal instructions: “Walk as you normally would on land. Try to keep one foot in contact with the ground at all times, and try not to bounce or float between steps. You may walk with your arms at your sides, but do not use your arms to swim forward.” All practice trials were monitored by a member of the research team, with augmented feedback provided if necessary.

Participants then attended an experimental session scheduled for at least 24 h after familiarization. Upon arrival, participants underwent surface electromyography (sEMG) preparation. This involved gently shaving and cleaning the designated sEMG sites with an alcohol swab. Subsequently, adhesive waterproof electrodes (Cometa Mini Wave, Cometa SRL, Milan, Italy) were carefully positioned over the tibialis anterior (TA) and gastrocnemius medialis (GM) muscles of their dominant leg, adhering to established guidelines from the Surface Electromyography for Non-Invasive Assessment of Muscles (SENIAM) project [31]. Once the electrode placement was completed, adhesive waterproof tape was used to further secure the electrodes, minimizing any potential movement artifacts during the study.

In the experimental protocol, participants walked for 2 min at both 2.5 mph and 3.5 mph, with the order of speeds randomized, both on land and in water. For a supplementary analysis, participants walked for an additional 2 min in water at 3.5 mph with 75% water jet resistance applied to the anterior torso. The supplementary bout of water walking was performed after the completion of the 2.5 mph and 3.5 mph walking conditions. Walk speeds were determined from existing literature, which indicates that the average preferred walk-run transition speed for young adults falls within the range of approximately 4.25–4.70 mph [32]. sEMG data were collected during walking at a sampling rate of 2000 Hz [29]. Video data of participants' lower legs and feet were captured at 100 Hz using an underwater camera (Miqus Underwater, Qualisys AB, Sweden) and time-synchronized with sEMG data.

Land walking was performed on an instrumented treadmill (Tandem Treadmill, AMTI, Watertown, MA, USA), and walking in water was performed on an adjustable-depth aquatic treadmill (2000 Series, HydroWorx, Middletown, PA, USA). The water temperature was maintained at 29.5 °C ± 0.2 °C, which is considered thermoneutral for aquatic exercise [33]. Participants walked in the water while submerged to the level of the xiphoid process. Participants performed land walking first to prevent movement of the electrodes while drying off and also to negate the potential impact of thermoregulation on muscle activation. A 5 min rest period was provided between land and water walks. An additional 2 min rest time was provided before participants underwent the supplemental water walking with jet resistance condition. Participants wore a swimsuit or compression shorts and walked barefoot in the water but wore athletic shoes while walking on land. While gait mechanics may differ between walking barefoot versus shod, we aimed to maximize ecological validity by selecting footwear conditions that closely match those used in typical real-world environments.

2.3. Data Analysis

Data for ten full, consecutive stride cycles were extracted from the last 20 s of each walking condition. A stride cycle was defined as two subsequent foot strikes occurring on the dominant foot. Stance and swing phases were differentiated using frame-by-frame video analysis. Stance was defined as foot strike to toe-off, and swing was defined as toe-off to foot strike. sEMG signals were processed in MATLAB (version R2023a, The Mathworks Inc., Natick, MA, USA). sEMG signals were passed through a band-pass fourth-order recursive Butterworth filter (10–500 Hz). sEMG signals were then pared to isolate data corresponding with the stance and swing phases of all ten stride cycles. TA and GM muscle activation magnitudes for each swing and stance phase were estimated by taking the root-mean-square (RMS) of filtered sEMG signals. RMS activation magnitudes were then averaged across the 10-stride cycles, with mean values for the swing and stance phases passed on for statistical analysis. The Co-A indices for the stance and swing phases were estimated by taking a linear ratio of antagonist to agonist activation magnitude. The TA was designated as the agonist during the swing phase and the antagonist during the stance phase. In contrast, the GM was designated as the agonist during the stance phase and the antagonist during the swing phase. Stance time (s), swing time (s), and stride rate (strides/s) were estimated using foot strike timing data acquired through video analysis. Stride length (m) was estimated from a standard gait velocity equation (Equation (1)).

$$\text{gait velocity} = \text{stride length} \times \text{stride rate} \tag{1}$$

2.4. Statistical Analysis

Statistical analysis was performed using RStudio (Version 1.1.456). Intraclass correlation coefficients (ICC) and corresponding 95% confidence intervals (95% CI) were used to evaluate the inter-trial reliability of sEMG measurements. ICCs were estimated from a single measure, absolute agreement, two-way mixed effects model. ICC values were classified as excellent reliability (>0.90), good reliability (0.75–0.90), moderate reliability (0.50–0.75), or poor reliability [34]. Prior to hypothesis testing, we confirmed the normality of the data using the Shapiro–Wilk test. For all dependent measures, the main effects and interactions between environment (land × water) and walking speed (2.5 mph × 3.5 mph) were evaluated using a two-way within-subjects analysis of variance (ANOVA). Post hoc analysis of significant interactions was performed using Tukey HSD pairwise comparisons. Post hoc analysis of the main effects was performed using paired *t*-tests. For the supplemental analysis, differences between walking in water at 3.5 mph with and without 75% jet resistance were evaluated using paired sample *t*-tests. Cohen's *d*-effect sizes were estimated using mean differences and pooled standard deviations. The interpretation of Cohen's *d* values followed an established scale from the literature [35]. All hypothesis tests were conducted using an alpha-type I error threshold of 0.05.

3. Results

3.1. Inter-Trial Reliability

The inter-trial reliability of the measures recorded here was moderate to excellent, with the exception of stance phase Co-A (water 3.5 mph with jets), swing phase GM RMS (water 3.5 mph; water 3.5 mph with jets), and swing phase Co-A (water 3.5 mph; water 3.5 mph with jets; $p < 0.001$; see Tables 2 and 3).

Table 2. Inter-trial reliability of measures during the stance phase of walking.

Environment	Measure	2.5 mph	3.5 mph	3.5 mph + Jets
Land	Stance time (s)	0.71 (0.61–0.81)	0.94 (0.91–0.96)	
	TA RMS (μV)	0.57 (0.45–0.70)	0.69 (0.58–0.80)	
	GM RMS (μV)	0.84 (0.77–0.90)	0.90 (0.85–0.94)	
	Co-A (%)	0.65 (0.54–0.77)	0.80 (0.72–0.88)	
Water	Stance time (s)	0.87 (0.81–0.92)	0.81 (0.74–0.88)	0.77 (0.68–0.85)
	TA RMS (μV)	0.73 (0.63–0.82)	0.78 (0.70–0.86)	0.69 (0.59–0.80)
	GM RMS (μV)	0.63 (0.51–0.75)	0.56 (0.44–0.69)	0.83 (0.75–0.89)
	Co-A (%)	0.61 (0.49–0.73)	0.66 (0.54–0.77)	0.46 (0.34–0.61)

TA = tibialis anterior; GM = medial gastrocnemius; RMS = root-mean-square amplitude; Co-A = co-activation index.

Table 3. Inter-trial reliability of measures during the swing phase of walking.

Environment	Measure	2.5 mph	3.5 mph	3.5 mph + Jets
Land	Swing time (s)	0.87 (0.81–0.92)	0.89 (0.84–0.93)	
	TA RMS (μV)	0.82 (0.74–0.89)	0.87 (0.80–0.92)	
	GM RMS (μV)	0.61 (0.49–0.73)	0.87 (0.81–0.92)	
	Co-A (%)	0.66 (0.55–0.77)	0.81 (0.73–0.88)	
Water	Swing time (s)	0.88 (0.83–0.93)	0.80 (0.71–0.87)	0.81 (0.73–0.88)
	TA RMS (μV)	0.89 (0.83–0.93)	0.88 (0.82–0.92)	0.92 (0.88–0.95)
	GM RMS (μV)	0.84 (0.77–0.90)	0.35 (0.24–0.50)	0.46 (0.34–0.61)
	Co-A (%)	0.83 (0.76–0.90)	0.46 (0.34–0.61)	0.45 (0.33–0.60)

TA = tibialis anterior; GM = medial gastrocnemius; RMS = root-mean-square amplitude; Co-A = co-activation index.

3.2. Interactions

Significant environment × speed interactions were observed for stance time ($F = 44.63$, $p < 0.001$), swing time ($F = 25.03$, $p < 0.001$), stance phase GM RMS ($F = 17.45$, $p < 0.001$), stance phase Co-A ($F = 17.20$, $p < 0.001$), swing phase TA RMS ($F = 36.23$, $p < 0.001$; Figure 1), swing phase GM RMS ($F = 19.56$, $p < 0.001$), and swing phase Co-A ($F = 14.42$, $p < 0.001$; Table 4). Environment × speed interactions were not observed for stride length ($F = 0.59$, $p = 0.446$), stride rate ($F = 1.26$, $p = 0.269$), or stance phase TA RMS ($F = 1.17$, $p = 0.287$; Table 4).

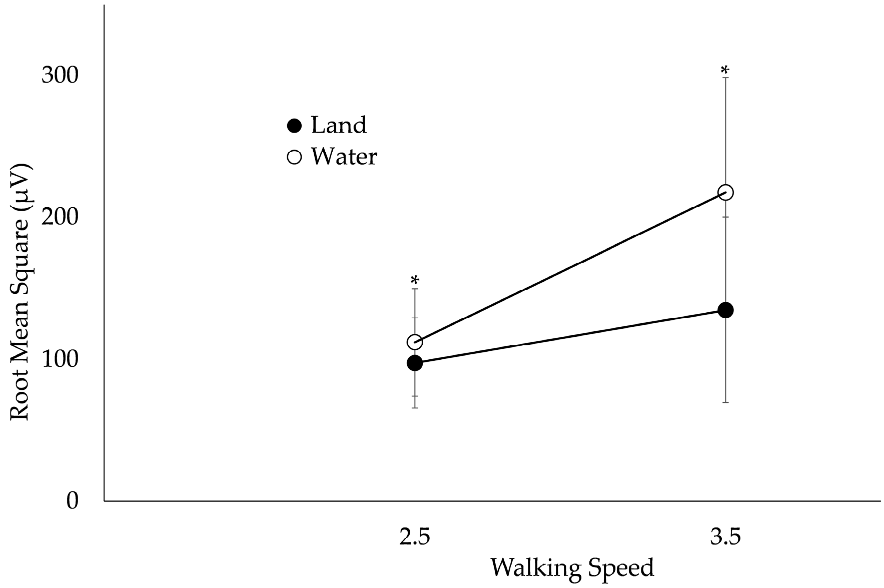

Figure 1. Speed × environment interaction on tibialis anterior muscle activation during the swing phase of walking gait.

Table 4. Post hoc comparisons on significant environment × speed interactions.

Measure	Land 2.5 mph	Land 3.5 mph	Water 2.5 mph	Water 3.5 mph
Stance time (s)	0.75 (0.04)	0.64 (0.04) [a]	0.81 (0.06) [a,b]	0.63 (0.04) [a,c]
Swing time (s)	0.38 (0.03)	0.36 (0.03)	0.62 (0.07) [a,b]	0.57 (0.05) [a,b,c]
GM RMS Stance (µV)	116.9 (42.4)	144.9 (64.8) [a]	56.4 (24.3) [a,b]	110.4 (41.5) [b,c]
Co-A Stance (%)	59.7 (27.8)	79.2 (38.1) [a]	135.3 (75.0) [a,b]	110.4 (56.8) [a,b]
TA RMS Swing (µV)	97.5 (31.9)	134.8 (65.3) [a]	112.1 (37.8) [a]	217.4 (81.0) [a,b,c]
GM RMS Swing (µV)	18.5 (17.8)	17.9 (13.9)	8.7 (7.4) [a,b]	18.1 (7.8) [c]
Co-A Swing (%)	19.6 (16.1)	14.4 (9.8) [a]	8.7 (8.7) [a,b]	9.1 (5.3) [a,b]

[a] significantly different from land 2.5 mph ($p < 0.05$); [b] significantly different from land 3.5 mph ($p < 0.05$); [c] significantly different from water 2.5 mph ($p < 0.05$); TA = tibialis anterior; GM = medial gastrocnemius; RMS = root-mean-square amplitude; Co-A = co-activation index. Data are presented as mean (SD).

3.2.1. Kinematics

Post hoc comparisons revealed that stance time was significantly longer in water than on land at 2.5 mph ($p < 0.001$), with no difference observed between environments at 3.5 mph ($p = 0.936$). Stance time was also significantly longer for the 2.5 mph water condition compared to all other conditions ($p < 0.001$; Table 4). Swing time was significantly longer for the 2.5 mph water condition compared to all other conditions ($p < 0.001$) and for the 3.5 mph water condition compared to both speed conditions on land ($p < 0.001$). No significant difference was observed between the 2.5 mph and 3.5 mph land conditions ($p = 0.221$; Table 4).

3.2.2. Stance Phase sEMG

Stance phase GM RMS was significantly greater for the 3.5 mph land condition than all other conditions ($p < 0.001$–0.039). Stance phase GM RMS was also significantly lower for the 2.5 mph water condition compared to the 2.5 mph land and 3.5 mph water conditions ($p < 0.001$). No significant difference was observed between the 3.5 mph water and 2.5 mph land conditions ($p = 0.927$; Table 4). Stance phase Co-A was significantly greater for the 3.5 mph land condition compared to the 2.5 mph land condition ($p = 0.002$). Stance phase Co-A was also significantly greater for both speed conditions in water compared to both speed conditions on land ($p < 0.001$). No significant difference was observed between speeds in water ($p = 0.171$; Table 4).

3.2.3. Swing Phase sEMG

Swing phase TA RMS was significantly greater in water than on land at matched speeds ($p < 0.001$–0.024). Swing phase TA RMS was also significantly greater for the 3.5 mph water condition compared to both 2.5 mph water and land conditions ($p < 0.001$) and for the 3.5 mph land condition compared to the 2.5 mph land condition ($p = 0.028$). No significant difference was observed between the 2.5 mph water and 3.5 mph land conditions ($p = 0.321$; Table 4). Swing phase GM RMS was significantly lower for the 2.5 mph water condition than all other conditions ($p = 0.005$–0.015). No other significant differences were observed ($p = 0.997$–1.000; Table 4). Swing phase Co-A was significantly greater for the 2.5 mph land condition compared to the 3.5 mph land condition ($p = 0.011$) and significantly lower for both speed conditions in water compared to both speed conditions on land ($p < 0.001$). No significant difference was observed between the 2.5 mph and 3.5 mph water conditions ($p = 0.914$).

3.3. Main Effects

3.3.1. Environment

Significant main effects of environment were observed on several dependent measures, including stance time ($F = 15.68$, $p < 0.001$), swing time ($F = 811.53$, $p < 0.001$), stride rate ($F = 733.18$, $p < 0.001$), stride length ($F = 446.97$, $p < 0.001$), stance phase GM RMS ($F = 59.11$, $p < 0.001$), stance phase Co-A ($F = 41.26$, $p < 0.001$), swing phase TA RMS ($F = 49.91$, $p < 0.001$), and swing phase Co-A ($F = 13.42$, $p < 0.001$; Table 5).

Table 5. Central tendency and dispersion results collapsed across speed and environment.

Measure	Land	Water	2.5 mph	3.5 mph
Stance time (s)	0.70 (0.07)	0.72 (0.10) ^{a}	0.78 (0.06)	0.63 (0.04) ^{b}
Swing time (s)	0.37 (0.03)	0.59 (0.07) ^{a}	0.50 (0.14)	0.46 (0.11) ^{b}
Stride length (m)	1.41 (0.16)	1.73 (0.19) ^{a}	1.44 (0.20)	1.71 (0.19) ^{b}
Stride rate (strides $\times$ s^{-1})	0.95 (0.08)	0.77 (0.09) ^{a}	0.79 (0.10)	0.93 (0.10) ^{b}
TA RMS Stance (μV)	79.7 (30.9)	80.3 (38.0)	61.0 (22.1)	99.0 (34.3) ^{b}
GM RMS Stance (μV)	130.9 (56.2)	83.4 (43.4) ^{a}	86.6 (45.9)	127.7 (56.8) ^{b}
Co-A Stance (%)	69.4 (34.5)	122.9 (67.2) ^{a}	97.5 (67.8)	94.8 (50.5)
TA RMS Swing (μV)	116.1 (54.4)	164.8 (82.2) ^{a}	104.8 (35.5)	176.1 (84.1) ^{b}
GM RMS Swing (μV)	18.2 (15.8)	13.4 (8.9) ^{a}	13.6 (14.4)	18.0 (11.2) ^{b}
Co-A Swing (%)	17.0 (13.5)	8.9 (7.2) ^{a}	14.1 (14.0)	11.8 (8.3) ^{b}

^{a} significantly different from land ($p < 0.05$); ^{b} significantly different from 2.5 mph ($p < 0.05$); TA = tibialis anterior; GM = medial gastrocnemius; RMS = root-mean-square amplitude; Co-A = co-activation index. Data are presented as mean (SD).

Post hoc comparisons revealed that stance time ($p < 0.001$), swing time ($p < 0.001$), stride length ($p < 0.001$), stance phase Co-A ($p < 0.001$), and swing phase TA RMS ($p < 0.001$) were significantly greater in water compared to land, while stride rate ($p < 0.001$), stance phase GM RMS ($p < 0.001$), and swing phase Co-A ($p < 0.001$) were significantly greater on land compared to in water. There were no main effects of the environment on stance phase TA RMS ($F = 0.03$, $p = 0.866$) or swing phase GM RMS ($F = 3.39$, $p = 0.074$; Table 5).

3.3.2. Speed

Significant main effects of speed were observed on several dependent measures, including stance time ($F = 1283.66$, $p < 0.001$), swing time ($F = 97.50$, $p < 0.001$), stride rate ($F = 964.41$, $p < 0.001$), stride length ($F = 957.12$, $p < 0.001$), stance phase TA RMS ($F = 161.27$, $p < 0.001$), stance phase GM RMS ($F = 95.23$, $p < 0.001$), swing phase TA RMS ($F = 81.55$, $p < 0.001$), swing phase GM RMS ($F = 11.21$, $p = 0.002$), and swing phase Co-A ($F = 5.72$, $p = 0.022$; Table 5).

Post hoc comparisons revealed that stride rate ($p < 0.001$), stride length ($p < 0.001$), stance phase TA RMS ($p < 0.001$), stance phase GM RMS ($p < 0.001$), swing phase TA RMS ($p < 0.001$), and swing phase GM RMS ($p = 0.002$) were significantly greater for the 3.5 mph speed conditions, while stance time ($p < 0.001$), swing time ($p < 0.001$), and swing phase Co-A ($p = 0.013$) were greater for the 2.5 mph speed conditions. There was no main effect of speed on stance phase Co-A ($F = 0.23$, $p = 0.634$; Table 5).

3.4. Effects of Added Water Jet Resistance

Paired-sample t-tests revealed that stride rate ($p = 0.011$), stance phase TA RMS ($p < 0.001$), stance phase GM RMS ($p < 0.001$), swing phase TA RMS ($p < 0.001$), and swing phase GM RMS ($p < 0.001$) were significantly greater for water walking performed with jet resistance compared to without. Additionally, stance time ($p < 0.001$) and stride length ($p = 0.017$) were significantly lower for water walking performed with jet resistance compared to without. No significant differences were observed for swing time ($p = 0.316$), stance phase Co-A ($p = 0.069$), and swing phase Co-A ($p = 0.971$; Table 6).

Table 6. Central tendency and dispersion results were compared between water walking conditions at 3.5 mph with and without jet resistance.

Measure	Water 3.5 mph	Water 3.5 mph + Jets
Stance time (s)	0.63 (0.04)	0.60 (0.05) [a]
Swing time (s)	0.57 (0.05)	0.57 (0.06)
Stride length (m)	1.87 (0.13)	1.83 (0.14) [a]
Stride rate (strides $\times$ s^{-1})	0.84 (0.06)	0.86 (0.09) [a]
TA RMS Stance (μV)	101.2 (40.3)	152.7 (61.9) [a]
GM RMS Stance (μV)	110.4 (41.5)	206.2 (98.2) [a]
Co-A Stance (%)	110.4 (56.8)	92.4 (60.7)
TA RMS Swing (μV)	217.4 (81.0)	330.4 (121.7) [a]
GM RMS Swing (μV)	18.1 (7.8)	27.7 (10.2) [a]
Co-A Swing (%)	9.1 (5.3)	9.1 (4.0)

[a] significantly different from water 3.5 mph ($p < 0.05$); TA = tibialis anterior; GM = medial gastrocnemius; RMS = root-mean-square amplitude; Co-A = co-activation index. Data are presented as mean (SD).

4. Discussion

This study aimed to investigate lower-limb muscle activation during walking gaits performed by young adults in water versus on land. Our hypothesis that TA activation would be greater during the swing phase of aquatic treadmill walking compared to land treadmill walking at matched speeds was confirmed. Additionally, we hypothesized that the difference in TA activation between aquatic treadmill walking and land treadmill walking would be more pronounced at 3.5 mph compared to 2.5 mph, which was also confirmed. We observed a significant 15% increase in TA RMS sEMG during the swing phase while walking in water compared to walking on land at 2.5 mph (Cohen's $d = 0.42$). The increase in TA RMS sEMG was considerably more pronounced for walking in water compared to on land at 3.5 mph (123%; Cohen's $d = 1.13$). We also observed meaningful effects on swing phase TA activation between water walking conditions. For instance, TA RMS sEMG increased 94% (Cohen's $d = 1.77$) during walking in the water at 3.5 mph compared to 2.5 mph. Additionally, TA RMS sEMG significantly increased by 52% (Cohen's $d = 1.11$) for walking in water at 3.5 mph performed with 75% jet resistance compared to without jet resistance.

Indeed, our findings regarding the heightened TA activation during the swing phase of water walking compared to land walking are compatible with the effects of fluid drag. Fluid drag is known to have a linear relationship with fluid density and a quadratic relationship with relative speed. In our study, the water temperature was carefully regulated and maintained at approximately 29.5 °C, resulting in a water density estimate of 995–996 kg/m^3 [36]. This density is approximately 800 times greater than that of ambient air [37]. In addition, by increasing the walking speed in water or by adding jet resistance, fluid drag (water's resistance to the motion of the foot) increases quadratically, further amplifying the muscle activation necessary to maintain the position of the foot [23].

Our findings indicate that TA activation during water walking is modulated by either increasing the walking speed or by introducing jet resistance opposing the direction of the walking gait. We observed an increase in stride rate both on land and in water when walking speed was raised to 3.5 mph compared to 2.5 mph (+15%, Cohen's d = 1.40), which coincided with decreases in both stance and swing time, particularly in water. These findings align with the notion that increased lower limb velocity was accompanied by increased fluid drag, attributable to the quadratic relationship between the velocity of the foot relative to the surrounding water. Stride rate also increased for the 3.5 mph water walking condition when jet resistance was added; however, this increase appeared to be largely due to a reduction in stance time as there was no significant difference in swing time between the 3.5 mph water walking and 3.5 mph water walking with added jet resistance. This indicates that the application of jets led to greater motor drive as participants exerted more effort to maintain a stationary walking position on the aquatic treadmill. Anecdotally, many participants self-reported localized fatigue in the TA after completing the 3.5 mph water walking condition with jets, which supports the observed increase in TA activation.

Our hypothesis regarding greater GM activation in water compared to land during the stance phase of walking gait was not supported by the findings of this study. GM RMS sEMG was lower during the stance phase of water walking compared to land walking, with a more substantial effect observed at the matched 2.5 mph speed condition (+52%; Cohen's d = 1.82) relative to the 3.5 mph speed condition (+24%; Cohen's d = 0.65). While we expected the GM to play a larger role in propelling the body forward in the presence of increased fluid drag, the findings suggest that the buoyancy effect of water immersion, which offloads body weight, reduces GM activation during stance. Interestingly, as fluid drag increases with walking speed, differences in GM activation between water and land walking decrease. We also hypothesized that there would be no significant difference in GM activation during the swing phase of walking in water compared to on land. This hypothesis was partially supported, with no significant difference observed between water and land walking at the matched 3.5 mph speed condition. Curiously, however, GM RMS sEMG was significantly reduced for the 2.5 mph water walking condition relative to all other conditions. This observation may reflect a differential balance of contributions from buoyancy and fluid drag expressed at different walking speeds in water.

While our study did not specifically hypothesize about Co-A indices, we did observe that the Co-A of the GM during the swing phase was greater on land compared to in water, irrespective of the matched speed conditions at 2.5 mph (56%; Cohen's d = 0.88) and 3.5 mph (37%; Cohen's d = 0.70). Swing phase Co-A of the GM was influenced by a combination of increased TA RMS sEMG across both speed conditions and reduced GM RMS sEMG for the environment-matched 2.5 mph speed condition. This suggests the potential that net ankle torque is shifted toward dorsiflexion when walking is performed in water compared to on land at 2.5 mph. We also observed that the Co-A of the TA during the stance phase was greater in water compared to on land (44%; Cohen's d = 1.05), a finding that was explained by reduced GM activation, given that there was no significant difference in TA activation between land and water conditions.

Collectively, the medium-to-large effects observed in TA and GM activation during water walking compared to land support the notion of accentuated TA activation when walking is performed in water. However, it is important to contextualize these findings within the existing body of research. Prior studies have not observed significant differences in TA activation between land and water walking performed at self-selected speeds [24,25]. This lack of disparity may be attributed to participants' tendency to opt for slower walking speeds in water [24]. Consequently, mismatched speeds resulting in a greater speed of walking on land, paired with a reduction in fluid drag in the water, likely contribute to the absence of a significant effect of the environment on TA activation [24–27].

To target TA activation through aquatic treadmill walking exercise, it is important to optimize walking intensity through variables such as speed, water resistance application, and water depth. Moreover, further research is warranted to determine whether TA activation magnitude, duration, or a combination of both has a greater impact on longitudinal biomechanical and neural adaptations. While our study focused on RMS sEMG, future studies could explore muscle activation in terms of absolute duration and total activation [38]. Furthermore, it is essential to customize optimal walking conditions for specific populations, especially those with neurological or mobility limitations. Further research is needed to explore the optimal combination of walking speed, water resistance, and water depth to maximize TA activation and toe lift during the swing phase while also considering movement constraints. This is particularly relevant when extrapolating our study findings to populations with foot drops. For example, although our study introduced jet resistance only at 3.5 mph, future investigations should examine its implementation at slower speeds and with a sufficient immersion depth [28], which may better suit individuals with limited mobility or gait impairments. Given the observed influence of walking speed and jet resistance on the effect sizes, clinical populations such as those with foot drops may walk at a slower pace. Increasing the jet resistance may be an important consideration when extrapolating the findings of the current study to individuals with foot drops. It is also reasonable to assume that individuals with foot drop experience more drag opposing dorsiflexion during the swing phase, as foot drop results in a greater projected area. This increase in drag could enhance TA activation during the swing phase, especially if individuals with foot drop are unable to walk at a 2.5 mph pace in water due to their limitations. It is also important for future research to consider the water immersion depth, with chest-deep immersion recommended, as increased water depth has been observed to enhance the effect of fluid resistance on muscle activation during gait performed in water [28].

Several study limitations need to be acknowledged. Firstly, a convenient sample of healthy, recreationally active adults was chosen, limiting the generalizability of the findings to clinical populations. Future research should include individuals with conditions commonly associated with foot drop, such as cerebral palsy, multiple sclerosis, or spinal cord injuries, to assess the consistency of cross-sectional findings and the potential effectiveness of aquatic treadmill walking in reducing foot drop. Additionally, while increased TA activation during the swing phase in water was observed, this study did not establish a causal link between increased TA activation and biomechanical or neural adaptations in dorsiflexor function. Future studies should investigate the effects of aquatic treadmill interventions on TA excitability, voluntary dorsiflexor torque, and toe lift/clearance while walking, as demonstrated in previous research [14,15]. Lastly, it is important to consider the limitations of sEMG when interpreting the findings. Internal noise and movement artifacts stemming from factors like fat, skin, and muscle can directly influence sEMG recordings, potentially impacting the quality of the signal collection.

5. Conclusions

We conducted a cross-sectional investigation to examine the disparities in stride kinematics and muscle activation patterns of the TA and GM during walking gait performed by young adults in water compared to on land at matched speeds. We found that TA activation was higher during the swing phase of walking in water compared to on land, and it increased with higher water walking speed and the application of jet resistance. Conversely, GM activation was reduced during the stance phase of walking in water compared to on land, resulting in a higher TA Co-A. The results support the potential use of aquatic treadmill walking as a treatment option for individuals with foot drops. However, further research is needed to determine if the observed increase in TA activation during aquatic treadmill walking translates to improvements in voluntary dorsiflexor control. Future studies should explore population-specific aquatic walking intensities needed to effectively increase TA activation before generalizing the findings of our study to clinical populations, particularly those frequently affected by foot drop.

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