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Comparing Lower-Limb Muscle Activity During Gait Performed in Water Versus on Land

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COMPARING LOWER-LIMB MUSCLE ACTIVITY DURING GAIT

PERFORMED IN WATER VERSUS ON LAND

by

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Abstract

The purpose of this study was to compare lower-limb muscle activation during gait, performed in water versus on land, in order to provide preliminary evidence for the benefit of aquatic treadmill walking in treating individuals with foot drop. Foot drop is a debilitating symptom of several neurological disorders characterized by the inability to dorsiflex the foot while walking. Generally, it is due to weakness in the ankle dorsiflexor muscles and/or increased tone in the plantar flexor muscles. Previous research has found that exercise interventions that demand greater than normal activation of the tibialis anterior (TA) (i.e., the primary ankle dorsiflexor) may improve walking performance in individuals with foot drop. Correspondingly, higher drag forces associated with walking in water may also facilitate increased activation of the TA during the swing phase of gait, potentially leading to similar improvements in gait. Thus, the current study compared surface electromyographic activity in the TA and medial gastrocnemius (GM) during gait performed in water versus on land. Thirty-eight healthy, recreationally active adults completed the study. Each participant walked under five conditions (Land 2.5 mph, Land 3.5 mph, Water 2.5 mph, Water 3.5 mph, and Water 3.5 mph + Jets) for 2-min each while muscle activity in the TA and GM were recorded using surface electromyography. A two-way within-subjects analysis of variance was used to evaluate main effects and interactions. As a secondary analysis, paired samples *t*-tests were used to assess differences between walking in water with and without jet resistance. TA activity during the swing phase of gait was greater in water than on land and this effect increased with greater walking velocity and the application of jet resistance. Furthermore, GM activity during the stance phase of gait was lower in water compared to land. The results of this study provide evidence in support of aquatic treadmill walking as a potential treatment for individuals with foot drop. Additional research is needed to

establish if a causal relationship exists between increased TA activity during exercise and improvements in voluntary dorsiflexion during gait in individuals with foot drop.

Introduction

Foot drop is a serious condition associated with many neurological disorders, which increases fall risk and decreases the independence of individuals affected (Aprile et al., 2005; Elisabeth Carolus et al., 2019; Graham, 2010; Stewart, 2008). Foot drop is characterized by the inability to dorsiflex the foot (i.e., flex upward at the ankle) while walking, causing it to drag or slap against the ground while walking. Generally, this is due to weakness in the muscles responsible for dorsiflexion and/or overactivity in the muscles responsible for plantar flexion (i.e., flexing the foot downward at the ankle) (Graham J, 2010). Consequently, individuals with foot drop walk more slowly and have difficulty navigating unlevel environments such as stairs (Elisabeth Carolus et al., 2019; Graham J, 2010; Stewart, 2008). Additionally, compensatory mechanisms such as hyperflexion in the knee and hip ("steppage gait" and "hip hiking") can lead to improper skeletal loading and injury over time (Błażkiewicz et al., 2017; Wiszomirska et al., 2017). Moreover, foot drop may limit activities of daily living, leading to a decline in functional independence and quality of life (Aprile et al., 2005).

Foot drop can be caused by several neurological conditions affecting either the central or peripheral nervous systems, including stroke, multiple sclerosis, cerebral palsy, spinal stenosis, peripheral neuropathy, Charcot-Marie-Tooth disease, or local compression of the peroneal nerve (*Foot Drop Information Page | National Institute of Neurological Disorders and Stroke*, n.d.). Foot drop may also result from direct trauma such as a spinal cord injury, fibular fracture, or knee dislocation. Moreover, foot drop can develop because of positioning during surgery or even from habitually sitting with legs crossed (Elisabeth Carolus et al., 2019). While the incidence of foot drop has not been reported in the literature, it is evident that foot drop affects a wide range of populations. Considering the diversity of foot drop causes and the populations affected, it is important to investigate various treatment options to accommodate individuals with different levels of mobility.

Common treatment options include nerve decompression surgery, ankle-foot orthoses, and functional electrical stimulation (Elisabeth Carolus et al., 2019; Graham, 2010; Stewart, 2008). Surgical interventions are invasive and limited to treating peripheral causes of foot drop such as peroneal nerve compression or direct trauma (e.g., fracture, dislocation, etc.) (Elisabeth Carolus et al., 2019). Ankle-foot orthoses effectively improve ambulatory function, but they restrict ankle mobility which may lead to discomfort and muscle contracture over time (Kluding et al., 2013; Sheffler et al., 2006). In contrast, functional electrical stimulation enables individuals with foot drop to move the affected ankle through its entire range of motion by activating the tibialis anterior (TA), the primary muscle responsible for dorsiflexion, with an external electrical stimulus applied to the peroneal nerve. Research has shown that functional electrical stimulation has an immediate "orthotic effect" on walking performance, increasing toe clearance and walk speed while the device is in use (Graham, 2010; Kluding et al., 2013; Sheffler et al., 2006). Additionally, functional electrical stimulation may have lasting effects on TA excitability through strengthening corticospinal connections (Everaert et al., 2010). However, the disadvantages of functional electrical stimulation include difficulty putting on the device and skin irritation from the electrodes (Bulley et al., 2011; Taylor et al., 1999). Further research is needed to determine if voluntary control of the TA can be improved without surgery or the use of uncomfortable assistive devices.

Gait training may be an effective alternative for improving dorsiflexor function without nerve stimulation or surgery. Willerslev-Olsen et al. (2015) found that maximal voluntary dorsiflexion torque and toe lift at the end of the swing phase increased significantly in children with cerebral palsy after walking daily on an incline treadmill for 30 days. The authors proposed that walking uphill required greater voluntary activation of the TA during the swing phase to lift the toes in preparation for foot strike. In the same study, researchers found that intramuscular coherence in the beta and gamma frequency bands recorded from the TA increased, suggesting that intensive gait training leads to changes in corticospinal drive (i.e., central drive). The changes observed in central drive may be responsible for the improvements in gait function. A similar study used transcranial magnetic stimulation to measure changes in motor-evoked potentials in spinal cord injury patients after a gait training (Thomas & Gorassini, 2005). Researchers found that neural pathways to the TA became more excitable following a 16-week, body weight-supported treadmill intervention. Moreover, improvements in corticospinal tract function were positively and significantly correlated with improvements in walking function as assessed by the Walking Index for Spinal Cord Injury (WISCI II). While the adaptation mechanism is not fully understood, the authors suggest that intensive daily treadmill training involving voluntary activation of leg muscles may improve neural control of the TA and improve toe clearance during the swing phase while walking. However, individuals with limited mobility due to neurological disorders may be unable to perform intensive treadmill training on land (Michael et al., 2005; Weerdesteyn et al., 2008).

Conversely, aquatic treadmill walking is considered to be a safe alternative that boosts movement confidence and accommodates various levels of functional capacity (Becker, 2009; Iliescu et al., 2020). Specifically, water immersion introduces an upward buoyant force that acts to offload body weight and reduce the impact of potential falls. A review of aquatic exercise in people with Parkinson's disease reported that fear of falling is reduced in the aquatic environment (Cugusi et al., 2019). In addition to those with Parkinson's disease, aquatic exercise has been used as an effective intervention for managing motor symptoms in individuals with other neurological diseases including stroke, multiple sclerosis, dementia, and cerebral palsy (Becker, 2020). In short, water provides a safe environment for people with limited mobility to perform exercise with greater confidence. Furthermore, increased drag forces associated with walking in water may require greater voluntary activation of the TA compared to walking on land. As supported by Willerslev-Olsen et al. (2015), the repetitive activation of lower-limb muscles during exercise that requires greater TA activity may improve neural control of the ankle through strengthening corticospinal connections.

Drag forces act on the body to resist movement while walking in water and on land. However, when the velocity, shape, and surface area of the object in fluid are matched, the magnitude of the drag force depends on the density of the fluid (Pöyhönen et al., 2000). Therefore, the same individual experiences greater resistance to motion in water than on land due to the higher density of water compared to air. Furthermore, drag force increases as the relative velocity of the object in fluid increases. Similar to the tip of a windmill, the foot has a greater velocity during the swing phase of the walking stride cycle compared to the rest of the lower limb. Consequently, the greatest fluid resistance is present at the anterior foot and ankle during swing. As a result, walking on an aquatic treadmill should require higher TA activity to overcome increased drag at the distal aspect of the lower limb and lift the toes prior to foot strike. However, to design an effective aquatic intervention for foot drop, we must first understand lower-limb muscle activation patterns in water.

Results from previous studies examining the differences in TA muscle activity between aquatic and land treadmill walking are mixed. For example, at self-selected speeds there are no significant differences in peak or average muscle activity of the TA (Heywood et al., 2016; Masumoto et al., 2004, 2008). Conversely, when comparing matched walking speeds, Shono et al. (2007) and Lau et al. (2022) found that TA activity was significantly higher in water than on land. Likewise, Silvers et al. (2014) found an increase in the absolute duration of activity and total activation of the TA between water and land running at matched speeds. However, the previous study found no difference in TA activity magnitude (expressed as a percentage of maximal voluntary contraction) between environments. Importantly, the previous study measured muscle activity while running. Due to increased flexion at the knee and hip during the swing phase while running, drag forces opposing dorsiflexion would not be as great as they would if the leg were fully extended during swing, as it is while walking. The mixed findings on TA activity in water warrant further investigation. Additionally, few studies have measured plantar flexor (i.e., gastrocnemius) activity and no study has reported estimates of muscle coactivation between the TA and medial gastrocnemius (GM) during gait performed in water. Since the ability to dorsiflex the foot can be improved by increasing activation of the TA and/or reducing the antagonist action of the GM, it is important to measure the activity of both muscles. To our knowledge, a comparison between distal leg muscle activation patterns during land and aquatic treadmill walking, for both the stance and swing phases of gait, is a gap in the current literature.

The purpose of this study was to compare surface electromyography (sEMG) activity in the TA and GM during the stance and swing phases of gait between aquatic and land treadmill walking at matched speeds. We hypothesized that average TA activation during the swing phase would be greater while walking in water than on land, at matched speeds, due to the increased drag on the distal lower limb. We did not expect significant differences in TA activity between environments during the stance phase. We also hypothesized that GM activity would be greater in water than on land during the stance phase to propel the body forward in the presence of increased drag, but that no difference would be present between environments during the swing phase. Given the relationship between relative velocity and fluid drag, we also hypothesized that the difference in muscle activation magnitude between walking on land and in water would be larger at 3.5 mph versus 2.5 mph. The findings of this study provided insight into the influence of the aquatic environment on dorsiflexor activation and provide evidence in support of its potential as a treatment for foot drop.

Methods

Participants

An a priori power analysis was conducted in G*Power (f = 0.25, α = 0.05, β = 0.80). It was determined that, to detect a moderate effect size of f = 0.25, a minimum sample of 36 participants was necessary to reject the null hypothesis with 80% statistical power. Thus, we recruited a convenience sample of 38 healthy, recreationally active adults (Table 1). All participants completed the study protocol in full. In this study, we defined recreationally active as being able to walk for at least 20 minutes without the use of an assistive device. To be included in the study, participants had to be between the ages of 18-35 and meet the established criteria for recreational activity. Participants were excluded from the study if: (a) they reported a history of neurological disease expressing motor symptoms (e.g. stroke, multiple sclerosis, recent concussion, etc.), (b) they reported current physical discomfort or an injury that affects their ability to walk, (c) they reported having a surgical intervention on the lower limbs or trunk in the prior two years, (d) reported having torn a hip, knee, or ankle ligament in the past. Written informed consent was obtained from all participants via signature on an informed consent document approved by the University institutional review board.

Table 1. Participant Characteristics

Sex	n	Age(years)	Height(cm)	Body Mass (kg)
Total	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).

Experimental Design

This study utilized a cross-sectional, repeated-measures research design comparing within-subject differences in TA and GM muscle activity between environments (water vs. land) and walking velocity (2.5 mph vs. 3.5 mph). Each participant served as their own control, performing both the water and land trials at each of the two walking velocities.

Instruments

Land trials were performed on a Tandem Treadmill (AMTI, Watertown, MA, USA) that was located in a Motion Analysis Laboratory, while aquatic trials were performed in a HydroWorx 2000 Series pool (HydroWorx, Middletown, PA, USA). The HydroWorx pool contains an 8 x 12-ft underwater treadmill platform with variable floor depth. The water temperature was maintained at 29.5°C $\pm$ 0.2°C, a thermoneutral range for aquatic exercise (McArdle et al., 1992). Participants were submerged to the level of the xiphoid process during aquatic trials. Previous studies have shown that this depth is optimal for minimizing float time during the non-contact phase of the stride cycle (Rutledge et al., 2007).

Muscle activity of the TA and GM were recorded using a 16-channel waterproof sEMG system (Cometa Mini Wave, Cometa SRL, Milan, Italy). The raw sEMG signals were collected at 2000 Hz (Silvers et al., 2014). Video data were captured from the sagittal-plane view at 100 Hz using an underwater camera (Miqus Underwater, Qualisys AB, Sweden) and synchronized with the sEMG recordings.

Procedures

Participants underwent a separate familiarization session before experimental testing to acclimate to the aquatic treadmill. Anthropometric data including height, weight, foot length, and foot width were also collected at this time. The experimental testing for each participant was completed in a single session at least 24 hours after the familiarization session. To begin, electrode sites were shaved, cleaned with an alcohol swab, and adhesive waterproof electrodes were then be placed over the TA and GM of the dominant leg according to guidelines established by the Surface Electromyography (sEMG) for Non-Invasive Assessment of Muscles (i.e., SENIAM) project (Hermens et al., 2000). The dominant leg was defined as the leg with which the participant responded that they would kick a ball. Adhesive, waterproof tape was used to secure the electrodes in order to minimize movement artifact.

In the experiment, participants walked for 2-min at both 2.5 mph and 3.5 mph (randomized order) on land and in water (8-min total walk time) while sEMG was collected. For a secondary analysis, participants also walked for an additional 2-min at the 3.5 mph speed (water only) with 75% jet resistance. Walk speeds were determined based on previous literature which suggests that the average preferred walk-run transition speed for young adults is approximately 4.25 – 4.70 mph (Diedrich & Warren, 1995). The selected speeds required the participants to walk quickly while maintaining the pendulum gait characteristic of walking. To prevent movement of the electrodes while drying off, and for participant comfort, each participant performed the land trials first. This may have helped to negate the potential impact of thermoregulation on muscle activity. To reduce the potential impact of fatigue, a 5-min rest period was included after completing the land trials and before starting the aquatic trials. An additional 2-min rest time was included before the jet resistance condition. For the water trials, participants were instructed to walk as they would on land to minimize float time. Additionally, no arm swimming was allowed. Participants wore a swimsuit or compression shorts and walked barefoot during the water trials but wore athletic shoes during the land trials. 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.

Data Reduction

Ten full, consecutive stride cycles from the last 20 seconds of each trial were selected for analysis. A stride cycle was defined as foot strike to foot strike on the dominant foot. The 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.

Data Processing and Analysis

sEMG signals were processed in MATLAB (The Mathworks Inc., Natick, MA, USA). sEMG signals were passed through a band-pass 4^{th} order recursive Butterworth filter (10-500 Hz). sEMG signals were then pared down to contain signal data corresponding with each stance and swing phase for each of the ten consecutive stride cycles according to the timing determined from video data. For both muscles (TA and GM) in each condition (land 2.5 mph, water 2.5 mph, etc.), 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 phase passed on for statistical analysis. Coactivation (Co-A) indices for both stance and swing phases in each condition were estimated by taking a linear ratio of antagonist to agonist activation magnitude. The TA was considered the agonist during the swing phase and antagonist during the stance phase, while the GM was considered the agonist during the stance phase and the antagonist during the swing phase. Lastly, kinematic variables including stance time (s), swing time (s), stride length (m), and stride rate (strides/s) were computed and averaged across the 10 stride cycles for each condition. Mean values for kinematic variables were passed on for statistical analysis.

Statistical Analysis

Intraclass correlation coefficients (ICC) and their 95% confidence intervals (95% CI) were used to assess inter-trial reliability of sEMG measurements using a single measure, absolute agreement, 2-way mixed effects model. All statistical procedures were performed using RStudio (Version 1.1.456). For all dependent measures, main effects and interactions between environment (land × water) and walking speed (2.5 mph × 3.5 mph) were evaluated using a 2-way within-subjects analysis of variance (ANOVA). Any significant interaction effects were followed up with a post-hoc analysis using Tukey HSD pairwise comparisons. Upon the observation of main effects of environment or speed, post-hoc analysis was performed using paired *t*-tests. As a secondary analysis, paired samples t-tests were used to assess differences between walking in water at 3.5mph with jet resistance and without for each dependent measure.

All hypothesis tests were conducted using an alpha type I error threshold of 0.05. ICC values were classified as excellent reliability (>0.90), good reliability (0.75-0.90), moderate reliability (0.50-0.75), or poor reliability (<0.50) (Koo & Li, 2016).

Results

Inter-trial Reliability

Inter-trial reliability of measures was moderate to excellent, with exception of Co-A for water 3.5 mph with jets during the stance phase, GM RMS during the swing phase for water 3.5 mph and 3.5 mph with jets, and Co-A during the swing phase for water 3.5 mph and 3.5 mph with jets ($p < 0.001$; see Tables 2-5).

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

Measure	2.5 mph	3.5 mph
Stance time	0.710 (0.607 – 0.810)	0.938 (0.907 – 0.963)
TA RMS	0.572 (0.453 – 0.701)	0.691 (0.584 – 0.795)
GM RMS	0.843 (0.774 – 0.902)	0.901 (0.853 – 0.940)
Co-A	0.654 (0.543 – 0.767)	0.803 (0.722 – 0.876)

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

Table 3. Inter-trial reliability of water measures during the stance phase.

Measure	2.5 mph	3.5 mph	3.5 mph + jets
Stance time	0.872 (0.813 – 0.921)	0.814 (0.736 – 0.883)	0.772 (0.682 – 0.854)
TA RMS	0.725 (0.625 – 0.821)	0.783 (0.696 – 0.861)	0.693 (0.587 – 0.797)
GM RMS	0.629 (0.514 – 0.747)	0.559 (0.439 – 0.691)	0.826 (0.752 – 0.891)

Co-A	0.607 (0.490 – 0.730)	0.655 (0.544 – 0.768)	0.463 (0.342 – 0.606)

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

Table 4. Inter-trial reliability of land measures during the swing phase.

Measure	2.5 mph	3.5 mph
Swing time	0.869 (0.809 – 0.919)	0.889 (0.836 – 0.932)
TA RMS	0.817 (0.740 – 0.885)	0.865 (0.804 – 0.917)
GM RMS	0.606 (0.489 – 0.729)	0.868 (0.809 – 0.919)
Co-A	0.663 (0.552 – 0.774)	0.806 (0.725 – 0.877)

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

Table 5. Inter-trial reliability of water measures during the swing phase.

Measure	2.5 mph	3.5 mph	3.5 mph + jets
Swing time	0.883 (0.829 – 0.928)	0.796 (0.713 – 0.871)	0.811 (0.732 – 0.881)
TA RMS	0.886 (0.833 – 0.930)	0.876 (0.818 – 0.923)	0.921 (0.882 – 0.952)
GM RMS	0.839 (0.768 – 0.899)	0.351 (0.238 – 0.499)	0.464 (0.343 – 0.607)
Co-A	0.834 (0.762 – 0.896)	0.464 (0.343 – 0.608)	0.454 (0.334 – 0.598)

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

ANOVA

Central tendency and dispersion results for dependent measures are presented in Table 6.

Table 6. Central tendency and dispersion collapsed across environment and walking speed.

Measure	Mean (SD)
Stance time (s)	0.71 (0.09)
Swing time (s)	0.48 (0.13)
Stride length (m)	1.57 (0.24)
Stride rate (strides*s^{-1})	0.86 (0.12)
TA RMS Stance (μV)	80.0 (34.5)
GM RMS Stance (μV)	107.2 (55.4)
Co-A Stance (%)	96.2 (59.6)
TA RMS Swing (μV)	140.5 (73.6)
GM RMS Swing (μV)	15.8 (13.0)
Co-A Swing (%)	12.95 (11.50)

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

Interactions

Significant environment x speed interactions were found for stance time (F = 44.63, p < 0.001), swing time (F = 25.03, p < 0.001), GM RMS Stance (F = 17.45, p < 0.001), Co-A Stance (F = 17.20, p < 0.001), TA RMS Swing (F = 36.23, p < 0.001), GM RMS Swing (F = 19.56, p < 0.001), and Co-A Swing (F = 14.42, p < 0.001). The results of the post-hoc comparison revealed that stance time was significantly greater in water than on land at 2.5 mph (p < 0.001), but no significant difference was found between environments at 3.5 mph. Swing time was significantly greater at 2.5 mph than 3.5 mph in both environments (p < 0.001), but the effect of speed was greater in water than on land. GM RMS Stance was significantly greater on land than in water at both speed conditions ($p = 0.001$), but the difference between environments was greater at 2.5 mph. Co-A Stance was significantly greater at 3.5 mph than 2.5 mph on land ($p = 0.002$), but no significant difference between speeds was found in water. TA RMS Swing was significantly greater in water than on land at both 2.5 mph ($p = 0.024$) and 3.5 mph ($p < 0.001$) speed conditions, but the effect of environment was greater at 3.5 mph. GM RMS Swing was significantly greater on land than in water at 2.5 mph ($p = 0.015$), but no significant difference between environments was found at 3.5 mph. Lastly, Co-A Swing was significantly greater at 2.5 mph than 3.5 mph on land ($p = 0.011$), but no significant difference between speed conditions was found in water (Table 7). No significant environment x speed interactions were found for stride length ($F = 0.59$, $p = 0.446$), stride rate ($F = 1.26$, $p = 0.269$), or TA RMS Stance ($F = 1.17$, $p = 0.287$).

Table 7. 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)[a]	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 (9.83[a]	8.7 (8.7)[a,b]	9.11 (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).

Main Effects

ANOVA revealed a significant main effect of environment 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), GM RMS Stance (F = 59.11, p < 0.001), Co-A Stance (F = 41.26, p < 0.001), TA RMS Swing (F = 49.91, p < 0.001), and Co-A Swing (F = 13.42, p < 0.001). Post-hoc comparison revealed that stance time (p < 0.001), swing time (p < 0.001), stride length (p < 0.001), Co-A Stance (p < 0.001), and TA RMS Swing (p < 0.001) were significantly greater in water than on land, while stride rate (p < 0.001), GM RMS Stance (p < 0.001), and Co-A Swing (p < 0.001) were greater on land (Table 8). There was no main effect of environment on TA RMS Stance (F = 0.03, p = 0.866) or GM RMS Swing (F = 3.39, p = 0.074).

A significant main effect of speed was observed for 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), TA RMS Stance (F = 161.27, p < 0.001), GM RMS Stance (F = 95.23, p < 0.001), TA RMS Swing (F = 81.55, p < 0.001), GM RMS Swing (F = 11.21, p = 0.002), and Co-A Swing (F = 5.72, p = 0.022). Post-hoc comparisons revealed that stride rate (p < 0.001), stride length (p < 0.001), TA RMS Stance (p < 0.001), GM RMS Stance (p < 0.001), TA RMS Swing (p < 0.001), and GM RMS Swing (p = 0.002) were significantly greater at 3.5 mph, while stance time (p < 0.001), swing time (p < 0.001), and Co-A Swing (p = 0.013) were greater at 2.5 mph (Table 8). There was no main effect of speed on Co-A Stance (F = 0.23, p = 0.634).

Table 8. 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*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).

Secondary Analysis

Paired-samples t-tests revealed TA RMS Swing was significantly greater with jet resistance than without ($p < 0.001$). GM RMS Stance was also significantly greater with jet resistance than without ($p < 0.001$). Additionally, significant differences were observed between conditions for stance time ($p < 0.001$), stride length ($p = 0.017$), stride rate ($p = 0.011$), TA RMS Stance ($p < 0.001$), and GM RMS Swing ($p < 0.001$; Table 9). No significant differences between conditions were found for swing time ($p = 0.316$), Co-A Stance ($p = 0.069$), or Co-A Swing ($p = 0.971$).

Table 9. Central tendency and dispersion results for water 3.5 conditions (with jets versus without jets).

Measure	Water 3.5mph	Water 3.5mph + 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*s^{-1})	0.84 (0.06)	0.86 (0.06)[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).

Discussion

This study aimed to evaluate differences in lower-limb muscle activity during gait performed in water versus on land as an early investigation of the potential of aquatic treadmill walking as an intervention for foot drop. We hypothesized that TA activity during the swing phase of gait would be greater in water than on land and that the effect of environment would be larger at faster speeds. The findings of the present study support this hypothesis. Using sEMG recordings with time-synchronized video data, we found that healthy, recreationally active adults have greater average TA activity magnitude (TA RMS) during the swing phase while walking in water versus on land. The difference in TA activity during swing between water and land was small (14.7 ± 4.88μV; Cohen's *d* = 0.42) at the 2.5 mph speed condition. In contrast, the difference in TA activity during swing was much greater at 3.5 mph (82.7 ± 11.61μV; Cohen's *d* = 1.13). Considering that the effect of environment is small at slower walking speeds, this may provide a rationale as to why other studies did not observe significant differences in TA activity when comparing between land and water at self-selected speeds (Heywood et al., 2016; Masumoto et al., 2004, 2008).

Furthermore, TA activity significantly increased in water at 3.5 mph with the addition of 75% jet resistance (Table 9). Anecdotally, many participants self-reported local fatigue in the TA after completing the jet condition. More research is needed to determine the optimal environment x speed x jet resistance combination for increasing TA excitation and improving toe lift during the swing phase in populations experiencing foot drop. Shono et al. (2007) found that TA activity in elderly women was significantly higher in water than on land while walking against water flow at relatively slow speeds (≤ 1.5 mph). In the current study, jet resistance was only introduced at the 3.5 mph speed in water as supplementary exploration. Future studies should experiment with implementing jet resistance at slower speeds that might be more suitable for individuals with gait impairments.

As expected, there was no difference in TA activity during the stance phase between environments. Since the TA is not a primary mover during stance, it remains less active during this phase in both environments. Interestingly, we did observe that TA activity during stance was significantly greater at 3.5 mph than 2.5 mph in both environments (Table 8). Previous research supported the observed main effect of speed on TA activity during stance, which found that coactivation increases as gait velocity increases in young adults (Hortobágyi et al., 2009).

Interestingly, we found that stride rate is greater on land and stride length is greater in water (Table 8). The increase in stride length is likely due to increased swing time in water, with greater differences at 2.5 mph (Table 7). This suggests that healthy adults take longer steps while walking in water and that the duration of the swing phase is also increased in water, especially while walking at 2.5 mph. Consequently, the duration of TA activity during swing is likely greater at 2.5 mph in water, but the magnitude of activity is greater at faster speeds in water. Further research is needed to determine whether magnitude or duration, or a combination of the two, has a greater impact on neural adaptations. While it was beyond the scope of this project, other studies have reported muscle activity in terms of absolute duration and total activation (mean area under the curve) (Silvers & Dolny, 2011), and a similar approach could be explored in subsequent studies.

We also hypothesized that GM activity would be greater in water than on land during the stance phase. Contrary to our hypothesis, we found that healthy, recreationally active adults have greater average GM activity magnitude (GM RMS) during the stance phase while walking on land versus in water (Table 8). However, GM activity during stance at 3.5 mph in water was not statistically different from GM activity on land at 2.5 mph (Table 7). These findings suggest that offloading body weight through water immersion reduces GM activity during stance, but as drag forces increase with speed, GM activity in water begins to approach normal values on land. Interestingly, GM activity during swing only increased with speed in water. Reduced GM activity during swing was observed in water at 2.5 mph, but that activity increased to match land values at 3.5 mph (Table 7).

In a broader context, the findings of this study support the idea that an aquatic treadmill intervention might be an effective modality for treating individuals with foot drop. Since foot drop may result from inactivity in the dorsiflexors, or overactivity in the plantar flexors, a treatment modality that maximizes activation of the TA while minimizing activity in the antagonist muscles might be effective. Considering that our study found reduced GM activity during swing in water at 2.5 mph, an intervention employing similar speeds may be most suitable for individuals with foot drop due to overactive plantar flexors.

Analysis of Co-A ratios revealed a higher degree of TA Co-A during stance in water versus land at both the 2.5 mph and 3.5 mph speeds (Table 7). The results also show that GM Co-A during swing was greater on land versus in water (Table 8). Thus, providing further evidence that GM activity is reduced, and TA activity is elevated during the swing phase in water. However, it is important to mention that ICC values for GM RMS during swing indicate poor reliability at all 3.5 mph conditions in water (Table 5). This suggests that there was greater stride-to-stride variability in the GM sEMG signal during swing while walking in water at faster speeds. Interestingly, GM RMS showed greater reliability during the stance phase in similar conditions (Table 3), when the GM was acting as the primary mover. Lastly, since Co-A during swing is a linear ratio of GM RMS to TA RMS, the ICC values for Co-A during stance are also poor.

Limitations

A convenient sample of healthy, recreationally active adults was selected to conduct a preliminary comparison of lower-limb muscle activity between environments in young, healthy individuals. Therefore, this limits the generalizability of findings to clinical populations. Given that the broader purpose of this study was to investigate the potential of aquatic treadmill walking as a modality for treating foot drop, future studies should sample from clinical populations who commonly experience foot drop such as individuals with cerebral palsy, multiple sclerosis, or spinal cord injuries to determine if aquatic treadmill walking is associated with reduced footdrop.

Furthermore, while we found evidence of increased TA activity during swing in water, this project did not establish a link between increased TA activity and biomechanical or neural adaptations in dorsiflexor function. Similar to research by Thomas and Gorassini (2005) and Willerslev at al. (2015), future studies should examine the effect of an aquatic treadmill intervention on TA excitability, voluntary dorsiflexor torque, and toe lift/clearance while walking.

Lastly, the limitations of sEMG should be considered when interpreting the findings of this study, especially with regards to GM activity and Co-A ratios during swing at faster speeds in water. It is likely that there was some movement artifact relating to the fat, skin, and muscle on which the sEMG electrodes were attached.

Conclusions

The purpose of this study was to provide a preliminary cross-sectional investigation into the potential of aquatic treadmill walking as a modality for treating individuals with foot drop. Specifically, we compared electromyographic activity in the TA and GM during gait performed in water versus on land. Evidence was found that TA activity during swing is greater in water than on land and Co-A of the TA is greater during stance in water v. land. The effect of environment is exaggerated at faster speeds, and with the addition of jet resistance. Future studies should investigate the effect of added jet resistance at speeds more suitable for clinical populations. Furthermore, GM activity during stance is reduced in water compared to land. The findings provide evidence in favor of aquatic treadmill walking as a potential modality for treating individuals with foot drop. More evidence is needed to establish if the increased TA activity during aquatic treadmill walking results in improvements in voluntary dorsiflexor control. Future studies should assess differences in lower-limb muscle activity between environments in populations more commonly affected by foot drop.

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