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Comparison of Vestibular Contribution During Stair Ascent, Stair Descent, and Treadmill Locomotion

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Comparison of vestibular contribution during stair ascent, stair descent, and treadmill locomotion

By

Alex Kern, LAT, ATC

A plan B research project submitted in partial fulfillment of the requirements for the degree

of

MASTER OF SCIENCE

in

Health and Human Movement

Approved:

**Utah State University
Logan, UT**

2018

Abstract: Stair negotiation accounts for 16% of fall related traumatic spinal cord injuries and the vestibular system plays a crucial role in fall avoidance through its role in maintaining balance during human locomotion, but its role in stair negotiation is unclear. The aim of this study was to compare the vestibular contribution to a) stair negotiation versus locomotion and b) stair ascent to stair descent in healthy young participants. It was hypothesized that, with cadence held constant, vestibular influence would increase during stair negotiation when compared to locomotion. We also expected the vestibular influence to increase during stair descent vs. ascent. The timing of muscle responses were expected to change due to differences in gait patterns. Differences in vestibular influence were analyzed using time-dependent coherence and time-independent cross-correlation. To probe the vestibular system, we used a random bandwidth limited white noise stimulus and recorded EMG signals from muscles of the lower leg, upper leg, and hip. Participants completed 300 steps on the treadmill at a cadence of 76 steps/min and a velocity of 0.4 m/s (0.9 m/h). Participants also ascended and descended a 9-step staircase for a total of 300 steps at a cadence of 76 steps/min. When comparing stair negotiation to locomotion, we found increases in mean peak coherence values across most muscles, however, only the soleus was significantly different during locomotion vs. ascent ($.146 \mp .080$ vs. $.259 \mp .105$, $p = 0.001$). Surprisingly, we also observed decreases in coherence values during stair descent vs. ascent. Finally, differences were seen in the timing of muscle responses when comparing the three conditions. For example, the medial gastrocnemius showed greatest vestibular influence before peak EMG recording during locomotion and stair ascent, but after peak EMG recording on descent. Our findings suggest that the magnitude of difference in the influence of the vestibular system is muscle-dependent. Also, the findings of this study suggest that we rely more heavily on the vestibular system during ascent than descent.

Introduction

Fall-induced injuries among the elderly are a major public health issue, with incidences on stairs being especially dangerous. Indeed, 51% of falls that result in a traumatic brain injury occur while negotiating stairs (Boye et al, 2014), and they are three times more likely to produce a traumatic brain injury than falls while walking (Hwang et al, 2015). However, traumatic brain injury is not the only risk that the elderly population faces. Falls on stairs can lead to other injuries such as hip fractures or even death (Nagata, 2014). Furthermore, the elderly are not the only group at risk of injury. Across all age groups, stair negotiation accounts for 16% of fall related traumatic spinal cord injuries (Chen et al, 2016). Unfortunately, little is known about what causes loss of balance and what role biological systems play in maintaining balance to prevent falls. To date, most research has focused on vision's contribution to stair negotiation (Miyasike-daSilva and McIlroy, 2016; Foster et al, 2015; Schofield et al, 2017), but has neglected the contributions made by the other sensory systems. For example, the vestibular system's contribution to balance control increases as stability decreases (Britton et al, 1993; Fitzpatrick et al, 1994). Given the vestibular system's role during locomotion (Brandt, 2000), it is expected to make a similar or greater contribution during stair negotiation. This is because stair negotiation is inherently more unstable, when compared to level walking, due to the increased challenge to stabilize the head (Cromwell and Wellmon, 2001). However, the vestibular system's role during stair negotiation is unknown. For this reason, we wanted to investigate how the vestibular system aids in the maintenance of balance during locomotion and stair negotiation. Once we have a sound understanding of the vestibular system's role in balance throughout these tasks, future work can begin investigating how vestibular interventions might be used to prevent falls and address this public health problem.

The vestibular system senses acceleration and velocity of the head, allowing individuals to adjust their physical orientation in the event of an unexpected body movement. The vestibular organs are located in the inner ear and are comprised bilaterally of three semicircular ducts (horizontal, anterior, and posterior) and two otolith organs (sacculae and utricle). The semicircular ducts sense angular acceleration and velocity of the head. The otolith organs lie against the walls of the inner ear, between the semicircular ducts and the cochlea, and are sensitive to linear acceleration and orientation relative to gravity. When the head moves, vestibular signals are sent, via the eighth cranial nerve, to the brainstem (Fitzpatrick and Day, 2004) from which the signals are used to generate a motor response in the muscles of the body to maintain balance and/or generate movement.

The vestibular system plays a crucial role in maintaining balance during human locomotion through its influence over muscles in the lower limb and this role increases as stability decreases (Britton et al, 1993; Fitzpatrick et al, 1994). Typically, the vestibular role in movement control is investigated by using small amounts of electrical current to stimulate the vestibular nerve. The electric current is applied behind both ears resulting in modulation of the firing rate of the nearby vestibular nerve (Goldberg et al, 1982; Kim & Curthoys, 2004; Aw et al, 2008). The brain interprets the electrical stimulus as an unplanned head movement towards the cathodal electrode which is compensated for by a physical sway in the frontal plane, or roll, towards the anodal side (Fitzpatrick and Day, 2004). Commonly, electric vestibular stimulation is conducted using a step shaped waveform, where the current is held at a constant level for a short period of time, then turned off. More recently several studies have used a continuous random waveform that has been referred to as stochastic vestibular stimulation (Dakin et al 2007, 2009, 2011).

Stochastic vestibular stimulation (SVS) is often delivered using a bipolar binaural electrode configuration resulting in a small but continuously varying disturbance to medio-lateral balance when the head is facing forward. SVS has several benefits over traditional step stimuli in that its use reduces testing times, decreases postural disturbance, and facilitates the extraction of the time varying vestibular reflex modulation that occurs during cyclic activities, such as locomotion (Blouin et al, 2011). Recently, SVS was used to reveal that the vestibular influence over muscular activation in the lower limb (Dakin et al, 2007) is dependent on the phase of the stride cycle (Blouin et al, 2011; Dakin et al, 2013). The authors found that SVS-EMG coupling is present in muscles spanning the hip, knee, and ankle joints. More importantly, the strength of the SVS-EMG correlation in each muscle was found to be dependent on the phase in the gait cycle. For example, the largest SVS-EMG coupling was found in the medial gastrocnemius during the stance phase. This study also found a decrease in SVS-EMG coupling with increased locomotor velocity and cadence.

Recently, Dakin et al. (2013) investigated the time-dependent vestibular influence on muscles of the leg during locomotion. The authors found that vestibular signals influence the excitation of different muscles at different times depending on the phase of the gait cycle. From these findings it was concluded that the vestibular system contributes to the control of balance during locomotion in a phase specific manner. Here we used methods similar to that of Dakin et al. (2013) to observe how the vestibular system contributes to balance during stair negotiation and compare its influence to that observed during locomotion.

This study will serve as the first step in understanding the vestibular role in stair negotiation and will serve as the foundation to examine its role in falls. The aim of this study is to compare the vestibular contribution to a) stair negotiation versus locomotion and b) stair

ascent to stair descent in healthy young participants. It was hypothesized that, with cadence held constant, vestibular influence will increase during stair negotiation when compared to treadmill locomotion. Also, it was expected that the vestibular influence would increase during stair descent vs. ascent. This will be shown by increases in time-dependent coherence and cross-correlation amplitude. The timing of muscle responses were also expected to change due to differences in gait patterns between stair negotiation and locomotion.

Methods

Participants

This study used 10 healthy college aged participants (height 171 ± 8 cm, mass 69 ± 10 kg; 5 women, 5 men, aged from 21 – 28 yr.), and this sample size was determined based on the effect size observed by Dakin et al. (2013). Participants were recruited from the Utah State University student population by word of mouth. The experimental protocol was explained to each participant, their written informed consent was obtained, and they were coded with a 5-digit alphanumeric label. Participants also filled out a questionnaire to rule out any conditions that would contraindicate the use of stimulation, such as epilepsy. This research project was approved by Utah State University's Institutional Review Board (Protocol #7952).

Procedures

Stimulus: Stochastic electrical stimulation was delivered in a bipolar binaural electrode configuration in order to modulate the firing rate of the nearby vestibular nerve (Goldberg et al, 1982; Kim & Curthoys, 2004; Aw et al, 2008). The electrodes were coated with a thin layer of conductive gel and placed behind the ears, held there using transparent tape. Tape was also applied to the backside of each ear to prevent discomfort from the stimulus on the earlobe itself.

The electrodes were secured using prewrap, wrapped around the head and ears. In order to restrict the stimulus-induced perturbation to the frontal plane, each participant was required to maintain their Frankfurt plane 18° nose up relative to the horizontal plane to maximize the amplitude of the vestibular responses in the mediolateral direction (Fitzpatrick and Day, 2004). To monitor head orientation during trials, two stickers were placed on the side of the participant's head at an 18° angle. These markers allowed us to visualize if the subject was maintaining the proper head position. We allowed participants to look down when they were at the top of the staircase, in order to get their bearings. However, they were required to keep the proper head angle while actively ascending and descending the staircase, and we corrected them whenever their gaze was too low or high.

Before the start of the testing trials, subjects were acclimated to the stimulus by slowly increasing the intensity over consecutive 3 second trials. The first trial was a 1 mA step-shaped waveform, followed by 2 mA of step-shaped waveform. The next three trials increased from ∓ 3 to ∓ 5 mA but using the noisy bandwidth waveform that was to be used during the experiment. During stair negotiation, the vestibular stimulus consisted of a 22 second noisy bandwidth limited waveform (0-25 Hz) of varying amplitude and peaking at ∓ 5 mA.

Stairs: Participants ascended and descended a 9-step staircase over 78 separate trials, for a total of 312 steps. The staircase's dimensions (6 ¾" vertical rise and 12" tread) are within the international residential code for stairways (Anon, 2015). Participants were required to ascend and descend the staircase with one foot per step without assistance while being paced by a metronome at a cadence of 76 steps/min. This cadence was tested during pilot studies and chosen to be similar to that used on the treadmill in previous studies done by Dakin et al. (2013) while also providing a comfortable pace for walking on the staircase. A two-count pause was taken at

the end of ascent and start of descent to distinguish the two separate conditions during data analysis. Participants could take breaks at the end of each trial, as needed. They also wore heel and toe switches, secured with tape and prewrap, to indicate heel contact and toe off for each step cycle. To prevent falling, participants wore a harness tethered to a freely moving track above the stairs. This cable was on a moving track in order to remain directly over each participant, and it was part of a quadruped cage surrounding the staircase. The tether was linked to participants by an assisted braking device that could lock the tether length in the event of a fall. During each trial, a research assistant shadowed each participant to adjust the length of the tether and to lock it in the event of a fall. A second research assistant was also present to ensure that any loose wires attached to the participant did not get caught in the staircase or surrounding structures. There was a hand railing on one side of the staircase for participants to hold onto if they felt unsteady.

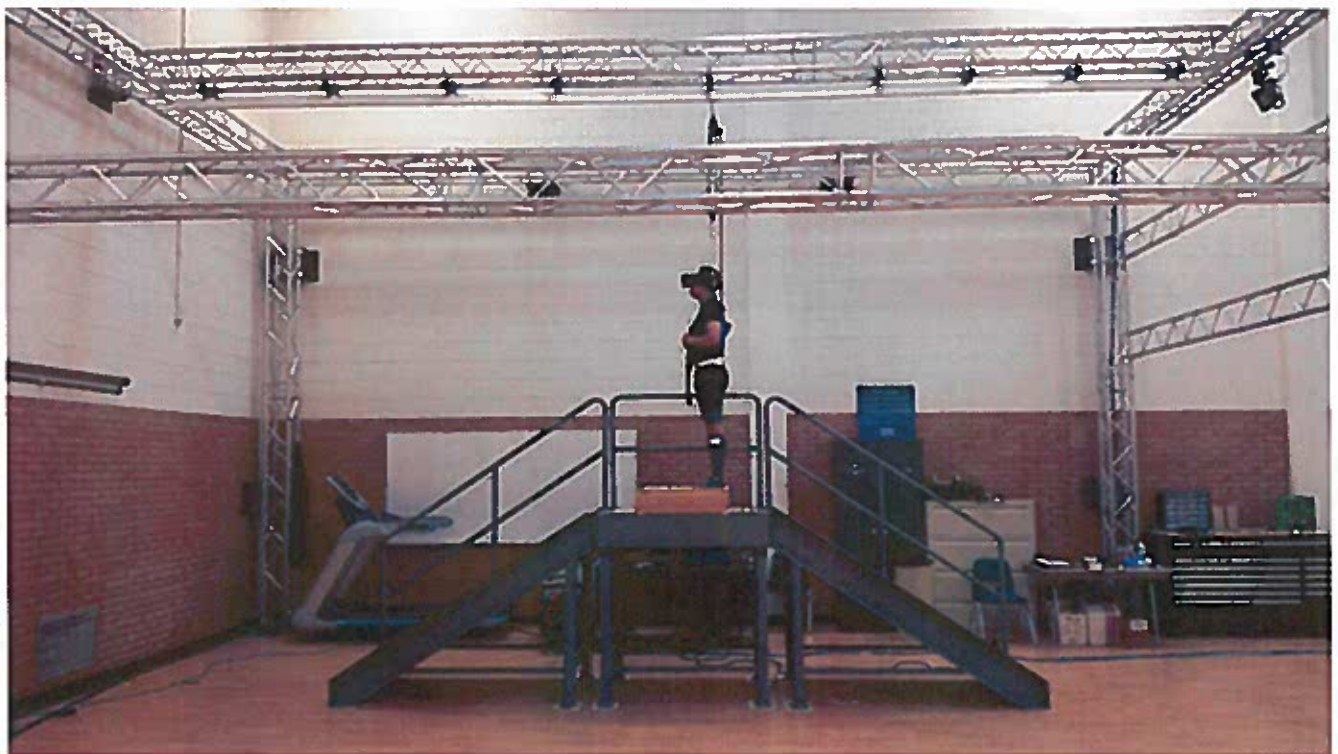


Figure 1: Image of the experimental setup illustrating a participant wearing the harness tethered to a freely moving track above the stairs to prevent falls.

Treadmill: Participants completed 380 steps on the treadmill at a cadence of 76 steps/min and a velocity of 0.4 m/s, controlled using a metronome. This cadence has been adjusted from previous locomotor work done by Dakin et al. in 2013 to assist in more comfortable stair negotiation. The stimulus bandwidth and amplitude were the same as the stairs but lasted for 10 minutes to coincide with the walking duration. This duration and stimulus bandwidth are ideal to observe the effect of the vestibular stimulus on leg muscles (Blouin et al, 2011; Dakin et al, 2013).

Electromyography: Surface electromyography was recorded to quantify the size and selectivity of vestibular influence on lower limb muscle activity. EMG was bandpass filtered from 20-450Hz during collection using wireless sensors. Electrodes were placed on the following eight muscles: tibialis anterior, soleus, medial gastrocnemius, vastus medialis, rectus femoris, semitendinosus, biceps femoris, and gluteus medius. Only muscles on the left side of the body were recorded as we assumed bilateral symmetry similar to that found during similar tasks, such as locomotion (Blouin et al, 2011; Dakin et al, 2013). Electrode contact areas were shaven and cleaned with rubbing alcohol to increase the quality of recordings. Then electrodes were placed in a parallel orientation to the direction of the muscle fiber, and manual muscle testing was performed to avoid cross talk. Electrodes were then secured into place using transparent tape and prewrap.

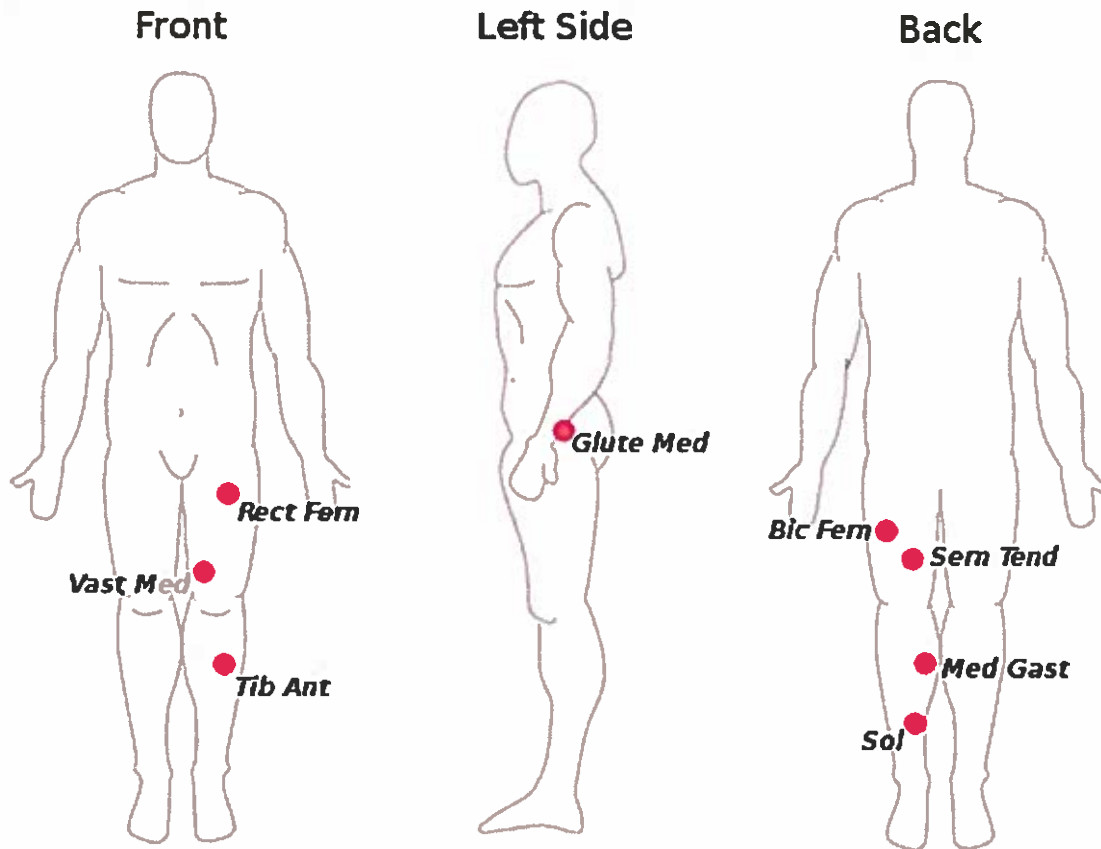


Figure 2: Location of EMG electrodes on the left side of the lower extremity. Electrodes placed on: tibialis anterior (Tib Ant), soleus (Sol), medial gastrocnemius (Med Gast), vastus medialis (Vast Med), rectus femoris (Rect Fem), semitendinosus (Sem Tend), biceps femoris (Bic Fem), and gluteus medius (Glute Med). Electrodes were placed parallel to the direction of the muscle fibers.

Motion Tracking (not presented in this thesis): A seven camera Vicon motion tracking system was used to measure the position of the body throughout trials. This system allowed us to measure the variance in postural motion that is due to vestibular stimulation. Due to limited number of cameras, only the left side of the body was recorded. Cameras were placed behind, above, beside, and in front of participants to record, then reconstruct the participants' kinematic behavior. Markers for motion tracking were placed on the following anatomical landmarks: head of the left 5th metatarsal, left lateral heel, left lateral malleolus, left lateral knee, left greater trochanter of the hip, bilateral posterior superior iliac spine, base of the sacrum, and 7th cervical

vertebrae. Markers were applied using an adhesive spray and transparent tape to secure in place, and the location of each marker was identified on the skin in case the marker were to fall off.

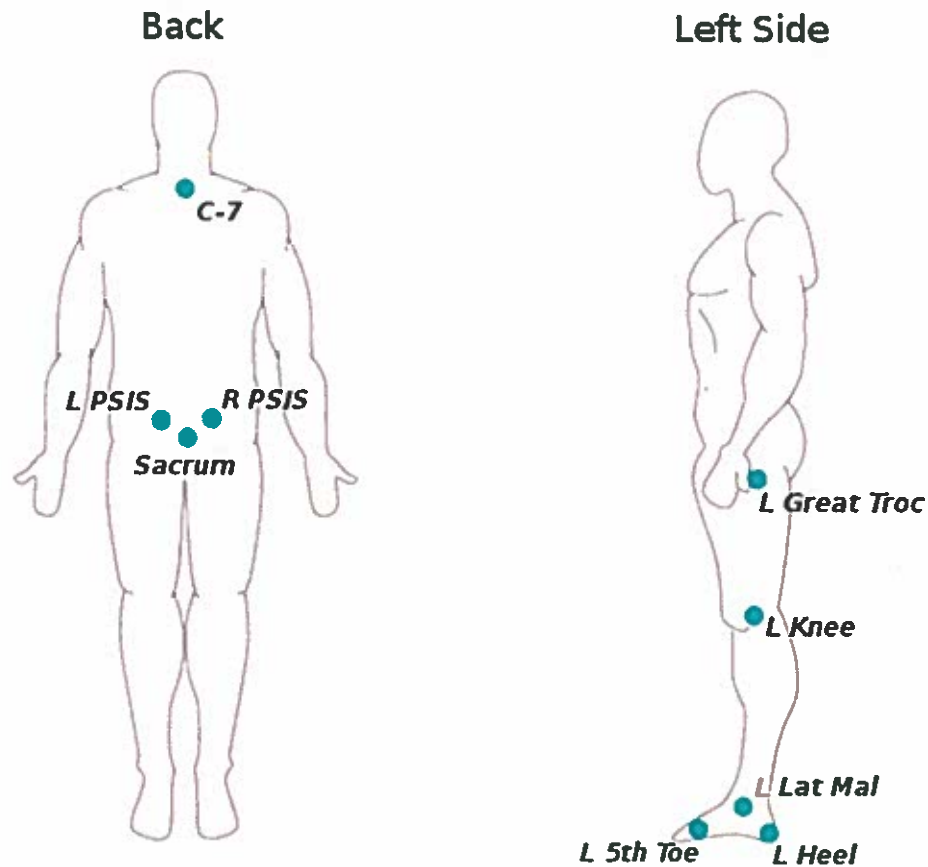


Figure 3: Locations of the motion tracking markers on the left side and back of the body. Markers were placed on: head of the left 5th metatarsal (L 5th Toe), left lateral malleolus (L Lat Mal), left heel (L Heel), left lateral knee at the joint-line (L Knee), left greater trochanter (L Great Troc), left posterior superior iliac spine (L PSIS), base of the sacrum (Sacrum), right posterior superior iliac spine (R PSIS), and 7th cervical vertebrae (C-7). Skin was labeled with a pen to ensure same location if the marker were to fall off.

Data Analysis

Signal Analysis: In order to quantify the relationship between vestibular stimulation and its effect on muscle activity, we calculated the time-independent cross-correlation and time-dependent coherence between the two signals. First, electromyography data were full wave rectified, then the data was cut into steps based on visual identification of toe contact via a toe

force sensor placed under the head of the first metatarsal in MATLAB (MathWorks, Version 2016a). On the treadmill, a step began when the left toe struck the treadmill and ended when the left toe struck the treadmill on the subsequent step. On the staircase, the step began when the left toe made contact with the first stair, and it ended when the left toe made contact with the subsequent stair. In order to compare each step, the step lengths were normalized to the length of the average step for each individual participant. 50% of the previous step and 50% of the following step of each step were added to each step to act as a buffer to prevent error when calculating the coherence at the start and end of the step. Once all steps were extracted, they were treated like trials and were used to calculate the cross-correlation and coherence.

Cross-correlation is a linear measure on a scale of -1 to 1 which measures the magnitude of the relationship between two signals over time. The cross-correlation can provide a measure of the time delay from when the stimulus is applied to when it influences muscle activation, as well as, the directionality of this influence (Blouin et al, 2011; Dakin et al, 2013). Once broken into steps the time-dependent cross-correlation was calculated by cross correlating the rectified EMG data with the stimulus for each time point in the step cycle, then averaging the cross-correlation across steps to get a single 'time-dependent' cross-correlation for each condition in each subject. The pooled cross-correlation was then averaged over the step cycle to get a 'time-independent' 2-D curve for each condition in each subject. Essentially, this time-independent cross-correlation measure hides modulation over the step cycle and instead shows the behavior averaged over the step cycle.

Time-dependent coherence is a linear measure on a scale of 0 to 1, which measures the magnitude of the relationship between two signals in the frequency domain. A coherence of 1 indicates a perfect linear relationship and a coherence of 0 indicates independence. To calculate

the time-dependent coherence, we again used the data segmented into steps (300 in total) for each condition and for each participant. The time-dependent coherence was calculated by decomposing the data using a Morlet wavelet transform and estimating coherence at each point in time based on the methods by Zhan et al (2006). This analysis was performed and averaged across the 300 steps on the staircase for both ascent and descent and 300 steps on the treadmill for each participant and then across each participant. The extra steps were available to substitute steps in which the participant stumbled or erred.

Data Reduction and Statistical Analysis: Time-dependent coherence data was determined different from zero when it exceeded a 95% confidence limit. The confidence limit for the coherence was derived from the number of steps (Halliday et al, 1995; Blouin et al, 2011). One-way repeated measures ANOVA was used to compare both peak coherence and response timing within a muscle compared between treadmill locomotion, stair ascent, and stair descent. Post hoc tests using least significant difference with a Bonferroni corrected p-value was used to evaluate the individual differences between conditions. To simplify the analysis the ANOVA was run on only the soleus, medial gastrocnemius, and semitendinosus muscles. The other muscles recorded did not exhibit unique periods of modulation over the step cycle, so we did not compare them. Alpha for both measures was set to $p = 0.05$.

Results

Coherence:

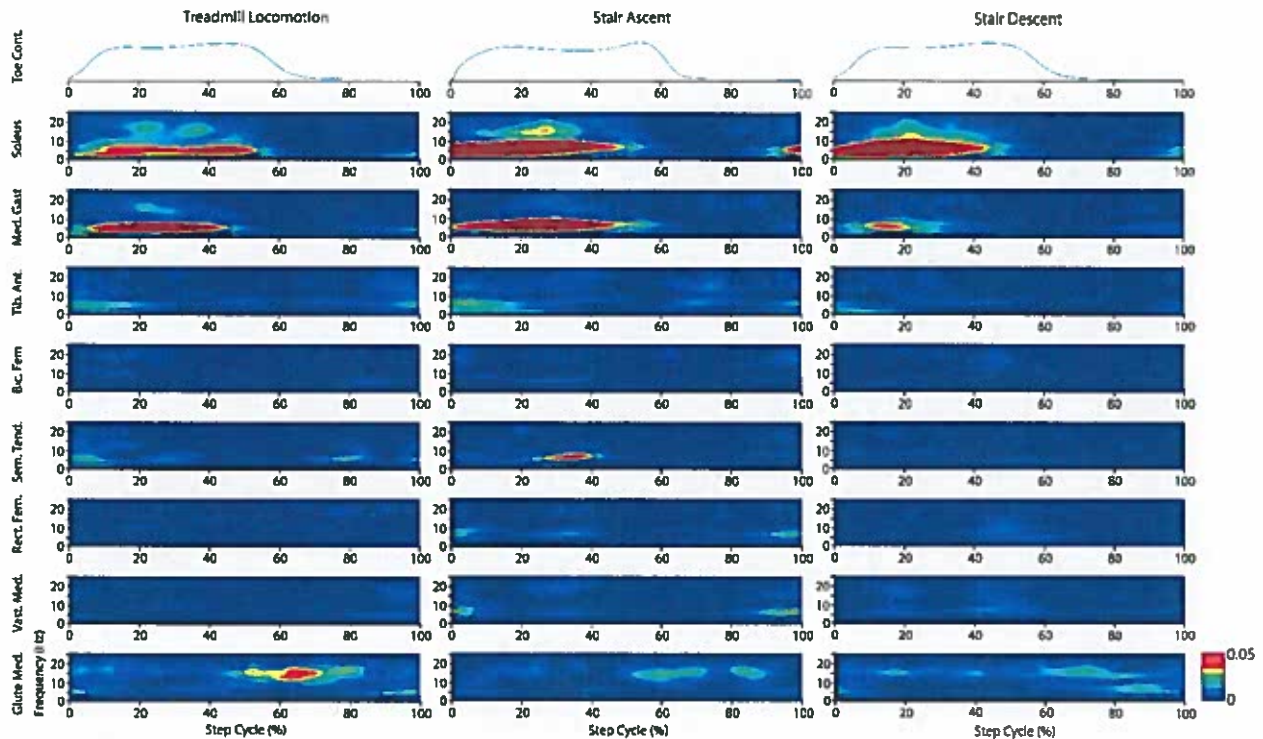


Figure 4: Pooled coherence for each of the three conditions in each of the recorded muscles ($n = 10$). The top row displays the average toe force sensor which delineates the average step cycle across all subjects. Step cycle in the top row begins at left toe contact and ends with left toe contact on the subsequent step. Toe-off occurs at about 55%-60% of the step cycle. Rows in descending order: toe contact (Toe Cont), soleus (Soleus), medial gastrocnemius (Med Gast), tibialis anterior (Tib Ant), biceps femoris (Bic Fem), semitendinosus (Sem Tend), rectus femoris (Rect Fem), vastus medialis (Vast Med), and gluteus medius (Glute Med). For coherence data: x-axis: percentage of the step cycle (%), y-axis: frequency, z-axis: coherence magnitude. The coherence displayed has been thresholded by 95% confidence limit. Coherence values below the confidence limit were set to zero and present as regions of dark blue. Red regions represent largest coherence. This pooled coherence figure has been added for illustrative purposes only. Exact values for average peak coherence is presented in Table 1. Average peak coherence is greater during stair ascent than locomotion in all muscles except the gluteus medius, and peak coherence during stair descent is lower than stair ascent in all muscles.

Coherence data were analyzed to compare the vestibular influence across all three conditions (Figure 4). The muscles showing the largest coherence over the step cycle were the soleus, medial gastrocnemius, and gluteus medius. The soleus and medial gastrocnemius exhibited the highest coherence with the stimulus from early stance phase to just before toe-off

whereas the gluteus medius exhibited the highest coherence with the stimulus just after toe-off during the beginning of swing-phase, but primarily during treadmill locomotion. Our findings showed that average peak coherence for all muscles except the gluteus medius was higher during stair ascent than locomotion. However, while gluteus medius decreased 0.046 during stair ascent compared to locomotion the differences were not significant ($.112 \pm .070$ vs. $.158 \pm .074$, $p = .233$). Also, differences were not significant during locomotion vs. descent ($.158 \pm .074$ vs. $.099 \pm .059$, $p = .056$).

For the soleus, peak coherence amplitude was $.146$ vs. $.259$, compared between locomotion vs. stair ascent, respectively. Repeated measures ANOVA found that mean peak coherence differed significantly between conditions ($F_{(2,18)} = 21.6$, $p = 0.001$). Follow-up comparisons revealed that coherence magnitude during treadmill locomotion was significantly lower than during stair ascent ($.146 \pm .080$ vs. $.259 \pm .105$, $p = 0.001$). Stair ascent coherence was also significantly larger than stair descent ($.259 \pm .105$ vs. $.176 \pm .076$, $p = 0.001$). Locomotion vs. stair descent was not significantly different ($.146 \pm .080$ vs. $.176 \pm .076$, $p = 0.476$). However, mean peak coherence during stair descent was slightly larger than during locomotion.

The medial gastrocnemius showed an increase of 0.04 during stair ascent compared to locomotion ($.170 \pm .109$ vs. $.130 \pm .053$, $p = .316$) and semitendinosus increased 0.039 during stair ascent compared to locomotion ($.095 \pm .063$ vs. $.056 \pm .036$, $p = .286$). For the medial gastrocnemius, peak coherence differed significantly overall between conditions ($F_{(2,18)} = 6.54$, $p = 0.007$). However, pairwise comparison revealed no significant differences between any combination of the three conditions. Treadmill locomotion was slightly lower than ascent ($.130$

$\bar{x} \pm .053$ vs. $.170 \bar{x} \pm .109$, $p = 0.316$) and larger than descent ($.130 \bar{x} \pm .053$ vs. $.083 \bar{x} \pm .052$, $p = 0.062$). Stair ascent was also larger than descent, ($.170 \bar{x} \pm .109$ vs. $.083 \bar{x} \pm .052$, $p = 0.062$).

For the semitendinosus, peak coherence differed significantly between conditions ($F_{(2,18)} = 6.48$, $p = 0.008$). Post hoc tests using the Bonferroni correction revealed significantly larger peak coherence during stair ascent than stair descent ($.095 \bar{x} \pm .063$ vs. $.032 \bar{x} \pm .009$, $p = 0.030$). There were no significant differences between treadmill locomotion and stair negotiation for the semitendinosus. However, mean coherence during locomotion was lower than ascent ($.056 \bar{x} \pm .036$ vs. $.095 \bar{x} \pm .063$, $p = 0.286$) and larger than descent ($.056 \bar{x} \pm .036$ vs. $.032 \bar{x} \pm .009$, $p = 0.176$). It is worth noting, that these comparisons are only looking at differences in peak coherence for coherence values already determined to be significantly different from zero by a 95% confidence limit seen as dark blue in Fig. 4. Anything above the 95% confidence limit appears as color increasing to dark red being the largest coherence values.

When comparing the timing of the muscle responses in the soleus, repeated measures ANOVA with a Greenhouse-Geisser correction for sphericity found that mean percent step cycle differed significantly between conditions ($F_{(1.196,10.761)} = 6.69$, $p = 0.022$). Post hoc tests revealed a significantly earlier peak coherence during stair descent than locomotion ($17.7 \bar{x} \pm 6.07$ vs. $30.2 \bar{x} \pm 13.5$, $p = .040$). Locomotion vs. ascent ($30.2 \bar{x} \pm 13.5$ vs. $21.1 \bar{x} \pm 3.66$, $p = .191$) and ascent vs. descent ($21.1 \bar{x} \pm 3.66$ vs. $17.7 \bar{x} \pm 6.07$, $p = .153$) were not significantly different in their timing of peak. The timing of muscular response for the medial gastrocnemius only produced a significant difference between the three conditions when one outlier subject was removed from the data set ($F_{(2,18)} = 3.82$, $p = .044$). This outlier was identified by SPSS as a value that was more than three interquartile ranges higher than the upper quartile, denoted by a star (*). However, post hoc

comparisons produced no significant differences between any of the three conditions. There were no significant differences in muscle response timing for the semitendinosus between conditions.

Averages	Coherence		% Step Cycle	
	mean	stdev	mean	stdev
TREADMILL				
Sol	0.14596	0.079826	30.21739	13.52432
Med Gast	0.129885	0.053109	25.46584	8.570278
Tib Ant	0.057387	0.025896	48.81988	38.29439
Bic Fem	0.037246	0.014946	66.70807	33.14143
Sem Tend	0.056052	0.035806	42.20497	36.12449
Rect Fem	0.048107	0.044612	61.36646	33.44253
Vast Med	0.049453	0.024946	56.45963	39.84278
Glute Med	0.158327	0.073929	42.67081	26.31893
ASCENT				
Sol	0.259245	0.105421	21.11801	3.659973
Med Gast	0.170006	0.10878	27.63975	7.148106
Tib Ant	0.065161	0.030879	34.40994	31.15333
Bic Fem	0.063901	0.036593	30.2795	35.59489
Sem Tend	0.095188	0.063455	29.50311	7.998527
Rect Fem	0.05197	0.024515	27.76398	27.07201
Vast Med	0.062806	0.028223	66.18012	37.60414
Glute Med	0.112337	0.069692	45.06211	26.57753
DESCENT				
Sol	0.175562	0.076123	17.70186	6.066288
Med Gast	0.082722	0.05178	25.68323	26.272
Tib Ant	0.05891	0.024549	16.64596	16.75945
Bic Fem	0.037533	0.006932	27.04969	15.3722
Sem Tend	0.031858	0.008645	49.84472	17.31303
Rect Fem	0.042288	0.012901	45.06211	26.13721
Vast Med	0.049161	0.017443	42.20497	26.8579
Glute Med	0.098803	0.058786	56.24224	26.68497

Table 1. Peak coherence amplitudes and time points in the step cycle averaged over all individual subjects. Means and standard deviations presented for both conditions of all muscles. Average peak coherence is greater during stair ascent than treadmill locomotion in all muscles except the gluteus medius. Average peak coherence tends to be greater during stair ascent than stair descent in all muscles. For percent step cycle, the soleus and medial gastrocnemius present relatively low variation in timing compared to all other muscles due to their relatively large responses. The semitendinosus also presented with a low standard deviation during stair ascent.

When comparing stair ascent vs. descent, every muscle recorded demonstrated larger average max coherence amplitude during ascent than descent. The coherence values presented in

Table 1 are the average peak coherences for the 10 participants studied. This table also presents the point in the step cycle where that peak coherence occurs.

Cross-Correlation:

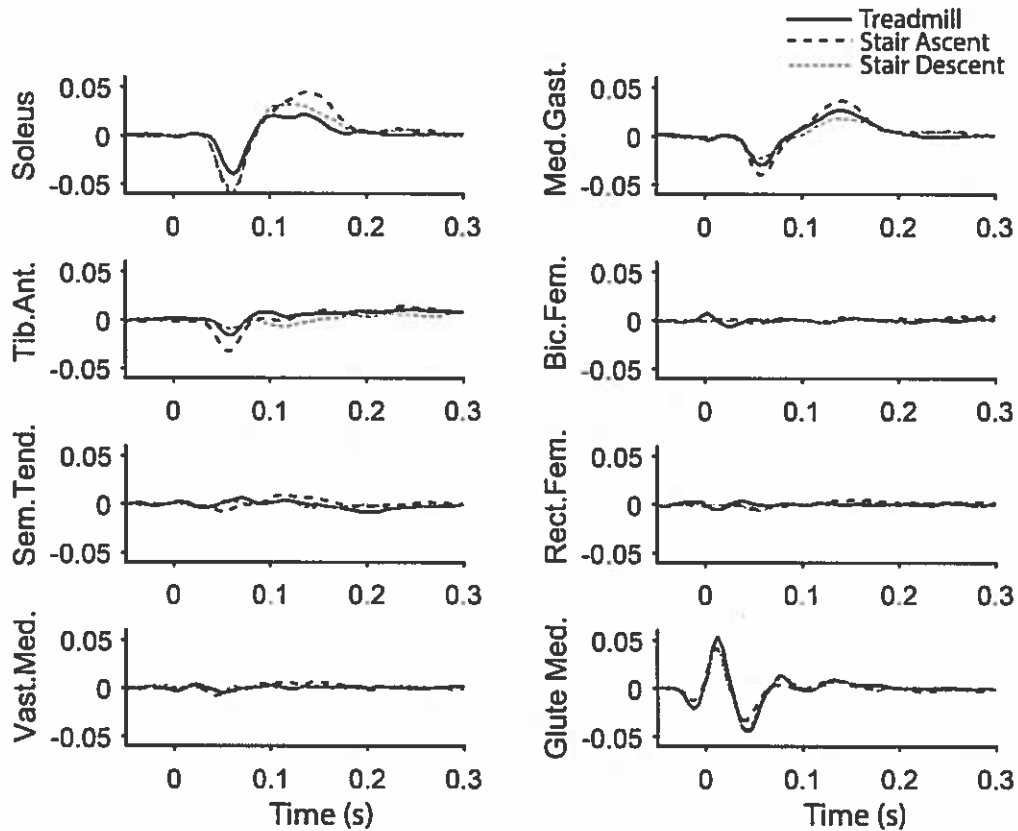


Figure 5: Average cross-correlations across time between the three conditions. y-axis is correlational magnitude, and x-axis is time in seconds. This illustrates the time lag of muscle response relative to the stimulus application, as plotted on the x-axis. A negative going waveform signifies muscle inhibition and a positive going waveform signifies muscle activation. The muscles are labeled as follows: soleus (Sol), medial gastrocnemius (MedGast), tibialis anterior (TibAnt), biceps femoris (BicFem), semitendinosus (SemTend), rectus femoris (RectFem), vastus medialis (VastMed), and gluteus medius (GluteMed).

The cross-correlation between rectified muscle activity and the stimulus was also calculated in order to provide a measure of the timing of vestibular influence on stairs vs. level walking (Figure 5). Figure 5 shows the average time delay from when the stimulus is applied to when it influences muscle activation. It also shows the directionality of the influence that the stimulation has over muscle activity. Due to the positive vestibular signals for anode right

currents, a positive going waveform indicates that anode right currents induced a facilitation of the muscle response (Dakin et al, 2007). The soleus, medial gastrocnemius, and tibialis anterior all exhibit negative going waveforms initially indicating muscular inhibition with anode right currents. However, the gluteus medius exhibited an initial positive going waveform, indicating an activation response. The muscles of the hamstrings and quadriceps were minimally correlated with the stimulus. Figure 5 also shows greater magnitude cross-correlations in the soleus, medial gastrocnemius, and tibialis anterior during stair negotiation than locomotion. However, in the gluteus medius it shows greater magnitude during locomotion than stair negotiation, which is consistent with our coherence findings.

EMG:

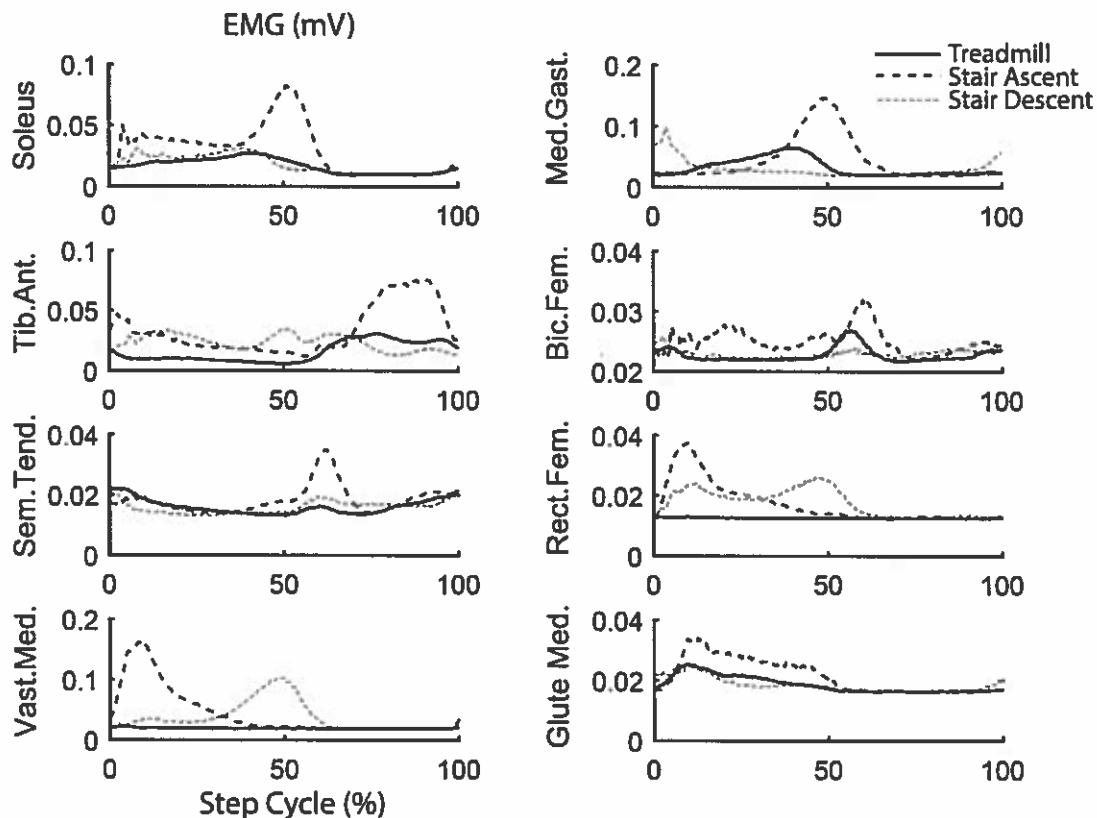


Figure 6: Muscle activity (EMG) across the step cycle for the three conditions. y-axis is in millivolts, and x-axis shows the percent of the step cycle from left toe-contact to subsequent left toe-contact. Toe-off is at approximately 55-60%.

Muscle activation patterns for the plantar flexors of the foot, soleus and medial gastrocnemius, exhibit very similar patterns between muscles across the three conditions. During locomotion, both show a steadily rising muscle activation until just before toe-off, followed by minimal activity during swing phase. Maximum ascent muscle recordings for these two muscles occur at approximately 50% of the step cycle, while during descent it occurs just after toe-contact. The tibialis anterior, which dorsiflexes the foot, shows the greatest activity during ascent while in swing phase until just before toe-contact. Similarly, on the treadmill the largest area of activity is during the late swing phase of the gait cycle, just before foot contact. During ascent the hamstring muscles, semitendinosus and biceps femoris, experience the greatest activation during and just after toe-off. However, the antagonist quadriceps muscle produce the greatest activity just after toe-contact of ascent and just before toe-off of descent. For all three conditions, the gluteus medius shows maximum activity just after toe-contact, and a steady decrease into minimum activity during swing phase.

Discussion

This study has provided the first in-depth analysis of vestibular influence during stair negotiation as compared to treadmill locomotion. The aim of this study was to compare the vestibular contribution to a) stair negotiation versus locomotion and b) stair ascent to stair descent in healthy young participants. It was hypothesized that, with cadence held constant, vestibular influence would increase during stair negotiation when compared to treadmill locomotion. The timing of muscle responses was also expected to change due to differences in gait patterns between stair negotiation and locomotion. Comparing stair negotiation to locomotion, we found increases in mean peak coherence values across most muscles, however,

only the soleus was significantly different during locomotion vs. ascent. Surprisingly, we also observed decreases in coherence values during stair descent vs. ascent. Finally, differences in the timing of muscle responses between the three conditions were observed. For example, the medial gastrocnemius showed greatest vestibular influence before peak muscle activation during locomotion and stair ascent, but after peak muscle activation during descent.

Coherence:

Increases in coherence indicate greater covariance between the stimulus and muscle activation resulting from a phase-locked relationship between the EMG envelope and the stimulus (Figure 4). Periods of increased coherence have been suggested to represent times in the gait cycle when the vestibular signals are most influential over balance control (Blouin et al, 2011). When compared to the study by Dakin et al. (2013), which used similar methods to the current study, it can be seen that our results are consistent with theirs. For example, the soleus and medial gastrocnemius had the greatest vestibular influence during the stance phase of walking until just before toe-off. However, peak coherence magnitude was lower in this study than theirs (0.146 ± 0.08 vs. 0.16 ± 0.13 , respectively for the soleus). This could be due to methodological differences such as our participants being harnessed while the participants in their study were not. The harness may have provided increased support, decreasing the need to maintain balance, which would decrease coherence by reducing reliance on vestibular cues to control the muscles. Since the vestibular system's contribution to balance control increases as stability decreases (Britton et al, 1993; Fitzpatrick et al, 1994), increased stability from the harness would decrease vestibular reliance. However, it is currently unclear whether the harness provided any increased or inhibited stability for our participants.

Cross-Correlation:

Cross-correlation is useful to understand the directionality (excitatory vs. inhibitory) of muscular responses resulting from the vestibular stimulus. The directionality of the response depends on the orientation of the head relative to the feet (Fitzpatrick and Day, 2004). With the head facing forward, vestibular stimulation will have the strongest influence in the frontal plane causing the sensation of unbalance towards the cathodal (-) side which is compensated for by a physical sway towards the anodal (+) side (Fitzpatrick and Day, 2004). The correlational magnitudes are greater in muscles that have a greater influence in frontal plane stability, such as the soleus, medial gastrocnemius, and gluteus medius (Fig. 5). Of note, in this study the soleus was inhibited followed by activation, which is opposite the finding by Dakin et al. (2013). This is due to the opposite polarity between legs when the head is facing forward (Dakin et al, 2011). While the vestibular system has an excitatory influence on one muscle, it also has an inhibitory influence over the same muscle at the same time on the contralateral side. Contralateral muscles are simultaneously inhibited and activated to produce the side-to-side physical sway in response to the vestibular stimulus. While these muscles lend to this physical sway, muscles of the quadriceps and hamstrings, which play a greater role in the sagittal plane, have lower correlational magnitudes implying less contribution to balance in the frontal plane.

Locomotion vs. Stair Negotiation

It was hypothesized that vestibular influence would be greater during stair negotiation than during locomotion because stair negotiation is associated with decreased stability compared to level walking, due to the increased challenge to stabilize the head (Cromwell and Wellmon, 2001). Also, stair walking requires a greater percentage of stance time throughout the total stride

when compared to level walking (McFayden and Winter, 1988). Our findings suggest there may be a mild increase in vestibular reliance on stairs in some muscles. While all muscles, except the gluteus medius, have a greater average peak coherence during stair ascent than during treadmill walking, the only significant increase was observed in the soleus. This increase is likely due to the fact that the greatest point of instability during ascent is at toe-off of the contralateral leg when the total body weight is given to the ipsilateral leg while all three joints are flexed (McFadyen and Winter, 1988). This period coincides with the point in the coherence data where the coherence for the soleus and medial gastrocnemius is greatest during stance phase (Fig. 4). In addition, while we did not statistically analyze the differences in the cross-correlational data, it mirrors the coherence data displaying greater peak correlational magnitude for all muscles except the gluteus medius during stair negotiation. It is currently unclear why the gluteus medius is the only muscle to trend towards larger vestibular modulation during locomotion than stair negotiation. These differences were not significant, however.

Stair Ascent vs. Stair Descent

Predictable disturbances to posture can be anticipated and compensated for in advance, requiring no vestibular contribution. However, unpredictable disturbances cannot be anticipated and require vestibular cues for post-disturbance compensation (Brooks and Cullen, 2014). A recent study done by MacNeilage and Glasauer (2017) used this idea to create a model to approximate the reliance on sensory (vestibular) and motor signals, based on the predictability of posture. They suggest that vestibular signals should receive more weight at slower walking speeds due to linear head acceleration being less predictable. Their predictive model suggests that vestibular reliance should be greater for walking down stairs vs. up stairs, however, our

findings would suggest otherwise. Peak coherence amplitude tended to be greater during stair ascent than descent for every muscle recorded, with significant differences in the soleus and semitendinosus. While it is unknown exactly why there would be less vestibular modulation during descent than ascent, there are two mechanical advantages during descent that could provide the answer. First, the body is in a more stable position due to the center of mass being more centrally located with respect to the foot (McFadyen and Winter, 1988). Also, during mid-swing of descent all three joints (ankle, knee, and hip) are in extension which prepares the body for weight acceptance (McFadyen and Winter, 1988). Another important difference between ascent and descent is how our body propels itself up and down stairs. Walking up stairs mostly requires concentric contractions to push and pull our body, while walking down stairs requires controlled falling by eccentric contractions (McFadyen and Winter, 1988). This controlled falling may render descent passively stable, similar to models described by Kuo (2002). Kuo used mechanical models to analyze the energy costs of walking and found that propulsion at toe-off and torques at the hip create a cycle where the motion of the swing leg is passive and requires minimal muscle activation. This may be similar during stair descent, when the swing leg falls passively to the subsequent stair, and only requires the muscular activation to clear the intermediate step (McFadyen and Winter, 1988). These mechanical advantages during stair descent could explain why there appears to be less vestibular influence than during ascent, but further examination in future studies is needed.

Timing of Muscle Activation

Peak EMG recordings did not always coincide with the peak coherence values observed in Fig. 4 and Table 1. Stimulus-muscle coherence increases when there is an increase in

vestibular influence on muscle activity and may not be directly related to modulation in muscle activity levels. Here we found that peak vestibular influence on muscle activity often does not occur when the muscle is most active, suggesting that these muscles' contribution to stability is higher at these points than when they are at their peak contraction. For this reason, the vestibular system may have its greatest influence over muscle activity when the muscle is not at its peak contraction.

As expected, the EMG patterns for the three conditions clearly demonstrated varying muscle activity patterns, suggesting different gait patterns which may influence the timing of muscle responses to vestibular stimulation. When comparing the EMG recordings of the medial gastrocnemius (Fig. 6) to the coherence magnitudes (Fig. 4), variations can be seen across all three conditions for both measures. For both treadmill walking and stair ascent, the majority of the coherence occurs during stance phase, ending just before peak muscle activity; slightly earlier during locomotion than ascent. However, during descent the majority of coherence occurs just after peak muscle activity. This suggests an increased role in balance control for the medial gastrocnemius as it is supporting the body being lowered during eccentric contraction.

While adjusted statistical testing only suggests significant differences between locomotion and stair descent in the soleus, it fails to reflect the size and location of the coherence areas as a whole, rather than just the location of the peak. As an example, the pooled coherence in Fig. 4 illustrates that coherence is greatest for the medial gastrocnemius in stair ascent throughout the entire stance phase. Whereas, during descent the majority of the coherence is only seen briefly, shortly after toe-contact. As explained earlier, peak EMG recordings do not necessarily align with areas of increased coherence. This is due to the complex relationship between moments of increased need for balance and muscular activity. Interestingly, peak EMG

recording for the soleus and medial gastrocnemius during ascent occurs around 50% in the step cycle, when the foot is plantar flexing to push off the step. Just after this, at about 60% the tibialis anterior is most active in order to dorsiflex the foot to clear the intermediate step (McFadyen and Winter, 1988). However, the largest areas of coherence for these three muscles occurs at the beginning of toe-contact and throughout the stance phase, primarily for the soleus and medial gastrocnemius. As discussed earlier, this supports the idea that we are most unstable during stance phase of ascent and suggests that vestibular signals are used to stabilize the ankle before push-off (Blouin et al, 2011). Of note, a similar phenomenon is seen in the semitendinosus during ascent, where peak coherence occurs just before peak muscle activation. However, in this case the knee is being flexed during swing-phase to help clear the intermediate step (McFadyen and Winter, 1988) and may require increased balance control towards the end of stance-phase to help maintain body posture.

Conclusions

This study has provided the first insight into how the vestibular system influences muscles of the lower leg as we negotiate stairs. Previous studies have only evaluated this relationship during level walking, and these findings have expanded on those to better understand the extent and timing of vestibular influences on stairs. Our findings suggest that the magnitude of difference in the influence of the vestibular system is muscle-dependent. While the soleus showed significant differences between the three conditions, muscles such as the gluteus medius and quadriceps did not appear to have any significant difference. This may be explained by differing gait patterns requiring modified balance requirements in each muscle. Surprisingly, the findings of this study also suggest that we rely more heavily on the vestibular system during

ascent than descent. This may be due to biomechanical advantages our bodies have descending vs. ascending stairs. This study provides the first steps into understanding the vestibular systems contribution to stair negotiations which could aid in future efforts to prevent falls during stair negotiation.

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