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Cortical Activation during Standing Balance in Modern Dancers

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An abstract of
A thesis submitted to the Faculty of the Emory College of Arts and Sciences
of Emory University in partial fulfillment
of the requirements for the degree of
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Abstract

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Understanding when and how cortical resources are engaged in balance recovery is crucial for assessing balance health prior to diagnosed balance impairments. Following a balance perturbation, a large negative peak (N1) can be recorded by electroencephalography (EEG). We interpret N1 as an error assessment signal that may reflect varying reliance on cortical resources. Analysis of cortical neuronal oscillatory frequency can further characterize EEG activity. The beta band (13-30 Hz) is associated with movement initiation and sensorimotor integration crucial for balance. Here, we investigate whether the N1 response or beta power differ in sensorimotor integration and balance ability experts – professional modern dancers. Because modern dance emphasizes responding to sensory information and correcting balance errors without missing a beat, I hypothesize modern dancers require less cortical sensorimotor integration to maintain balance than non-dancers. I predict that N1 amplitudes following balance perturbations will be smaller in modern dancers than controls and scale with perturbation difficulty. I also predict that dancers will have less sensorimotor cortical beta power following a challenging balance perturbation, and that beta will scale with balance ability within both groups. Testing this hypothesis, we used EEG to record cortical activity throughout a series of support-surface perturbations at varying difficulty. Balance ability, assessed as the mean distance traveled across a narrowing beam was also compared against N1 amplitudes. Overall, N1 amplitudes scaled with perturbation difficulty ($p<0.01$), but there was a trend toward dancers having larger N1 amplitudes than controls despite having better balance. There was a visually observed decrease in beta power 150-250 ms post-perturbation, which is after the N1 response. These results differ slightly from my prediction because dancers' N1 amplitudes did not scale with balance ability and were larger than controls while beta power was still less in dancers. N1 amplitude alone may not be a biomarker of worse performance on balance tasks, but rather an error assessment signal that can be sensitized by training.

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Chapter 1

Background

1.1 Introduction

Falls are the leading cause of mortality and morbidity in individuals over the age of 65 [1]. While there are many clinical measures used to assess fall risk in this population, the best predictor of a fall is a still previous fall [2]. Current fall risk assessments are based on indirect, biomechanical measures such as number of steps taken to recover balance [3]. However, there is evidence suggesting that balance health decline begins before measurable changes to biomechanical outcome measures assessed in the clinic. When automatic control of balance is insufficient to maintain balance alone, cortical resources have been shown to become engaged as well, presumably to compensate for inadequate subcortical control [4, 5, 6, 7]. Because these cortical resources can compensate effectively initially, visible balance impairment does not manifest until cortical resources are no longer enough to compensate for balance decline. Understanding when and how balance-correcting cortical resources are recruited in individuals without motor impairment can inform future balance assessments used in impaired populations. Here, we investigate whether cortical responses to balance perturbations differ in a cohort with fine-tuned sensorimotor integration and balance ability – professional modern dancers. Because modern dance training emphasizes responding to sensory information and correcting balance errors without missing a beat, dancers' balance correcting responses may be more automatic than nondancers'.

1.2 Cortical Control of Balance

Understanding when and how cortical resources are engaged in balance recovery can lead to a better grasp on factors that determine an individual's overall balance health prior to noticeable balance impairments.

During a destabilizing event, an automatic, brainstem-mediated balance correcting muscle response is evoked, followed by longer latency muscle activity that may be cortically mediated [4, 8, 9]. Individuals with lower balance ability can use cortical resources to compensate when the automatic response is insufficient [5, 6, 7]. Specifically, recruited cortical resources may aide in the sensorimotor integration required to sense a loss of balance and then recruit muscles to counteract it [10]. In healthy young adults (HYA), these resources are typically only recruited in very difficult postural conditions, such as those that require a reactive step to regain balance [6].

Reliance on cortical control for maintaining balance is thought to increase with age [4, 7, 11]. Similarly, the difficulty threshold of a destabilizing event necessitating recruitment of cortical resources decreases with age. Evidence for overcompensation on cortical resources with age and motor impairment comes from dual-task studies [5]. The dual-task experimental protocol involves introducing a cognitive task to be performed in conjunction with a balance task. Cognitive tasks require cortical resources, so if the balance task is also utilizing cortical resources such that the total amount of resources required to complete both tasks exceeds the finite amount of resources available, performance on one or both tasks will suffer. HYA are usually unaffected by dual task paradigms, but in both impaired and unimpaired older adults, performance on the balance task decreases when the cognitive task is introduced [5]. Even before the onset of visible balance impairment, this shift toward increased reliance on cortical resources to maintain balance begins to surpass an individual's total amount of cortical resources. This shift then should be when motor rehabilitation begins as the onset of falls will not occur until long after the subcortical resources governing balance first became insufficient.

1.3 Electroencephalography (EEG)

We can use EEG to quantify cortical resources. EEG records electrical activity from electrodes placed on the scalp overlying the neocortex. Because the cortex is the brain structure closest to the scalp, electrical activity from neurons in the cortex can be picked up by EEG [12]. Conversely, activity from deeper brain structures like the cerebellum or basal ganglia likely will not be recorded.

In the time domain following a balance perturbation, a large negative peak (N1) can be recorded by EEG. The cortical N1 is localized to the supplementary motor area [13] and is thought to be an error assessment signal [14]. As an error assessment signal, N1 amplitude may be a biomarker reflective of increased reliance on cortical resources in difficult postural conditions. Previous work has shown the N1 scales with balance challenge and balance ability in healthy young adults (HYA)[14]. The N1 also is increased in conditions with increased threat and attenuated when balance perturbations are predictable [15], giving further evidence to the N1 as an error detection signal in addition to a sensory integration of environmental stimuli.

A reason for looking at cortical activity during balance beyond attempting to quantify ”cortical resources” broadly is to investigate the extent to which cortical sensorimotor integration during balance-correcting responses takes place. To do this, we can analyze synchrony of cortical oscillations over time. Oscillatory activity at distinct frequency ranges occurs during different cortical processes, so isolating raw EEG data into frequency bands gives us insight into the underlying cortical processes at play during and after a balance-correcting response [16].

Circuits of neurons fire in an oscillatory pattern, where an individual neuron will fire when neurons near it fire, triggering the neurons near that neuron to fire as well. As various cognitive processes take place in different neurons, these oscillations can range from 5-50 Hz [16] which are then binned into discreet frequency bands. In this study, we analyze cortical beta (13-30 Hz oscillatory activity) power.

Dance training targets the skill of sensorimotor integration, of which beta power has been identified as a cortical correlate [11, 17, 18]. Beta power specifically has also been associated with maintenance of a current posture and decreases before planned movement [16]. During involuntary, automatic movement, increased beta power is thought to be a biomarker of maintaining a sensorimotor set [16]. Resisting a change in posture following balance perturbation is an example of an attempt to maintain a sensorimotor set. Indeed previous work from the lab demonstrated that sensorimotor cortical beta activity evoked during reactive balance recovery reflects sensorimotor processing as it increased with perturbation magnitude and in individuals with lower balance ability [11, 17].

1.4 Cortical Activity in Expert Cohorts

Dancers are colloquially known and have been shown to have superior balance compared to non-dancers [19]. This is intuitive due to the intense training professional dancers receive related to honing good balance. In addition to practicing balance, modern dancers specifically also receive training in improvisational techniques, which require increased spatial awareness, body awareness, and response to environmental stimuli including other dancers [20]. Thus, modern dancers maintain balance by integrating sensory feedback and making adjustments, not necessarily by maintaining their original posture. Previous work in the lab with ballet dancers has shown that dancers' muscle activation patterns are more efficient than nondancers and can be transferrable across tasks [19]. Since motor cortex circuits important for a particular skill become more efficient with training, the transferability in muscle activation patterns across tasks may also be reflected neurophysiologically [21]. Additionally, dance therapies have been shown to increase balance ability in individuals with Parkinson's Disease, although the mechanism for this effect is unclear [22]. Identifying the neural correlates of dancers' balance may provide insight into the ways in which dance therapy can improve balance.

Modern dance training differs from other forms of dance in that it teaches both static and dynamic balance; there are times when dancers must hold a posture, but they also learn how to flow in and out of unstable positions. There is value in leaning into off-balance positions to get to a more stable place rather than training balance just by holding one position. Therefore, their approach to maintaining balance following a balance perturbation may not focus on keeping their current posture.

1.5 Hypothesis

I hypothesize that professional modern dancers require less cortical sensorimotor integration to maintain balance during difficult postural conditions than non-dancer HYA due to training that leads to better automatic sensorimotor integration maintained subcortically.

In testing this hypothesis, I predict that dancers will have less sensorimotor cortical beta power following a challenging balance perturbation, and that beta will scale with balance ability within both groups. During an unplanned, reactive step to recover balance, I predict that dancers, who are trained to recover from precarious balance conditions, will have a smaller change in beta than nondancers.

Chapter 2

Methods

2.1 Participants

We analyzed data from 6 HYA (19-38 years old, 2 female) recruited from Emory University and the surrounding Atlanta community to participate in this study as well as 5 professional modern dancers (20-26 years old, 5 female) who each have at least 10 years of dance training. Exclusion criteria included history of neurological, musculoskeletal, and/or visual impairments (Table 2.1).

Participant Demographics		
Variable	Dancers (n=5)	Controls (n=6)
Age, years, mean±SE	23.4±0.8	24.3±1.4
Sex, male/female	0/5	4/2
Height, cm, mean±SE	164.3±2.5	176.7±6.0
BEAM Score, ft, mean±SE	20.4±1.2	18.8±0.7
Step Threshold Magnitude, mean±SE	14±1.0	15±1.2

Table 2.1: Participant Demographics

2.2 Experimental Protocol

The experimental protocol for this project included two support surface perturbation series. Perturbations were generated by a custom perturbation platform that translates in the forward and backward direction in the horizontal plane (Fig. 2.1). Embedded in the platform is a force plate that records 3D ground-reaction forces used to classify steps [22, 23]. During both series, EEG was used to record cortical activity from the scalp.

Figure 2.1: A participant on the platform during data collection

2.2.1 The Step Threshold Series

First, participants underwent a series of forward support surface perturbations delivered at unpredictable timing at varying magnitudes either above or below individual's step threshold, the magnitude at which an individual will take an unplanned step to regain balance 50% of the time [24]. Due to physical constraints of the platform, only one direction of perturbations could be included in each block of trials. However, to limit habituation we included catch trials randomly interspersed throughout the perturbation block. These trials were small displacement with high acceleration and in the backward direction, opposite of the other perturbations. We chose the forward direction because forward perturbations are more difficult to recover from, and we wanted to avoid a ceiling effect by ensuring that we could deliver a perturbation large enough to exceed all participants' step thresholds. Additionally, because the response to forward perturbations is to lean backward and people have less range of motion in the hip strategy leaning backward compared to leaning forward, these perturbations resulted in less overall body movement, therefore reducing the motion artifact in our EEG data. It takes about 30 trials to identify the individual's step threshold.

2.2.2 Planned and Unplanned Stepping Perturbations

The step threshold from the first perturbation series informed perturbation magnitudes in the second series. Participants were told to either step or not step in response to perturbations, and perturbations above step threshold were only delivered when the instruction was not to step, ensuring trials with unplanned steps. Unplanned step trials were interspersed with no step trials, where the participant did not need to take a step to recover balance with the same instruction. This ensured that there were two difficulty levels: perturbations in which a step was not needed to maintain balance and perturbations in which a step was needed. The "small" perturbation magnitude was equivalent to 60% of the participant's step threshold to ensure that they could recover balance without taking a step and the "large" perturbation magnitude was 140% of participant's step threshold to ensure that participants would be forced to take a step to recover balance. This normalized perturbation challenge across participants with varying step thresholds. Both planned and unplanned steps were recorded by visual inspection and confirmed with the vertical ground reaction forces from the force plate, where a step is defined as a trial in which the vertical load force under a limb drops below 10N in the first 2000ms after perturbation onset. The same catch trials were included in this series as well.

2.3 Balance Ability

Before the perturbation series, participants completed a challenging beam walking task to assess balance ability [19].

Wearing standardized shoes and with their arms crossed, participants attempted to walk all the way across a narrow balance beam (3.66 m long, 1.27 cm wide, 2.56 cm tall) (Fig. 2.2). Each of 6 beam walking trials ended when the participant reached the end, stepped off the beam, or uncrossed their arms. Beam walking was performed at self-selected speeds, and the only objective participants were given was to try to reach the end of the beam. Balance ability could then be quantified as a BEAM score, or the mean distance traveled across 6 trials. Participants were not instrumented with EEG equipment or kinematic markers during this task.

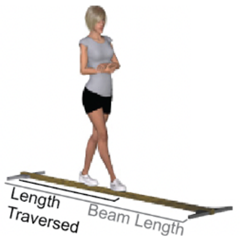

Figure 2.2: Beam walking setup

2.4 Data Analysis

2.4.1 EEG Pre-processing

A 64-channel set of actiCAP active recording electrodes (Brain Products GmbH, Munich, Germany) recorded brain activity continuously during the perturbation series. Electrode placement was in accordance with the international 10-20 system except for electrodes TP9 and TP10, which were placed directly on the skin over the mastoids. The Fz electrode was used as a reference. Recordings from the Cz electrode over primary motor and supplementary motor cortical regions were analyzed with custom MATLAB scripts and EEGLAB functions. To remove muscle activity and other artifacts from the EEG data, was run through a pre-processing pipeline following specifications outlined in Makoto's widely used pre-processing pipeline [25]. This removed artifacts from blinks, muscle activity, or bad electrodes to isolate brain activity components in the data.

This pipeline included first removing the epoch baseline, then decreasing the sampling rate of the data from 1000 Hz to 500 Hz and running the data through a 1Hz high-pass filter [26]. A channel was removed if it was flat for more than 5 seconds, had a high-frequency noise standard deviation of more than 4, or had less than a 0.6 correlation with nearby channels [27]. Removed channels were then interpolated and an empty Fz channel was added before computing an average reference. The Cleanline plugin was used to remove the 60 Hz line noise in each channel before extracting epochs for each trial and then running AMICA to identify independent components.

2.4.2 Quantifying N1 amplitudes and Beta Power

Once the data is pre-processed, N1 amplitudes can be quantified as the amplitude of the first negative peak following a perturbation at the central midline electrode (Cz), which overlies the sensorimotor area of the cerebral cortex (Fig. 2.3). These values were then compared to BEAM score and perturbation condition. Beta power was also measured at the Cz electrode. Changes in oscillatory power were quantified in single-trial epochs using wavelet time–frequency analyses in EEGLAB (pop_newtimef.m). A tapered Morlet wavelet with three oscillatory cycles at the lowest frequency (11 Hz), linearly increasing up to 6 cycles at the highest frequency (50 Hz) was used to measure power at each frequency in a sliding window of 256 ms. This transformation calculated the event-

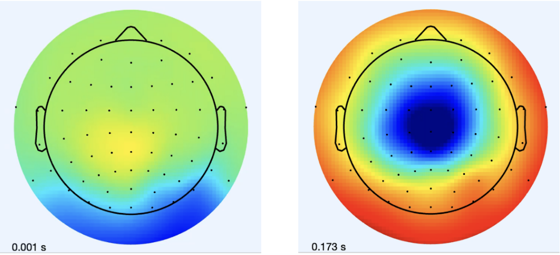

Figure 2.3: Compared to baseline at the time of perturbation onset (left) the greatest change in voltage recorded by EEG following a balance perturbation occurs at the cetral midline electrode (right). Black dots represent electrode placement. Green represents baseline voltage while blue is a decrease in voltage and red is an increase.

related spectral perturbation (ERSP), which represents changes in oscillatory power relative to perturbation onset in a defined set of frequencies [28]. Beta power was separated into 100 ms time bins and then compared against BEAM score and perturbation conditions. 50-150 ms post-perturbation was the time bin that captured the N1 response and 150-250 ms was the time bin that represented cortical activity after the initial N1 error assessment, while the participant is still executing a balance-correcting response.

Chapter 3

Results

3.0.1 N1 amplitudes scaled with perturbation difficulty and expectedness

Over all participants, N1 amplitudes were largest following unplanned steps and smallest following planned steps (p=0.0006) (Fig. 3.1A). N1 amplitudes following no step perturbations were larger than planned step perturbations (p=0.07) even though perturbation magnitudes were 60% of step threshold for both conditions. The difference between unplanned step and no step N1 amplitudes was near significant (p=0.08).

3.0.2 N1 amplitudes trend greater in dancers than nondancers

The trend of unplanned step N1 amplitude > no step > planned step holds when the participants are grouped into dancer and non-dancer categories. However, by visual inspection, the dancer N1 amplitudes appear greater in magnitude than their HYA counterparts in all perturbation conditions (Fig. 3.1B). This difference between dancer and HYA N1 amplitude was not statistically significant.

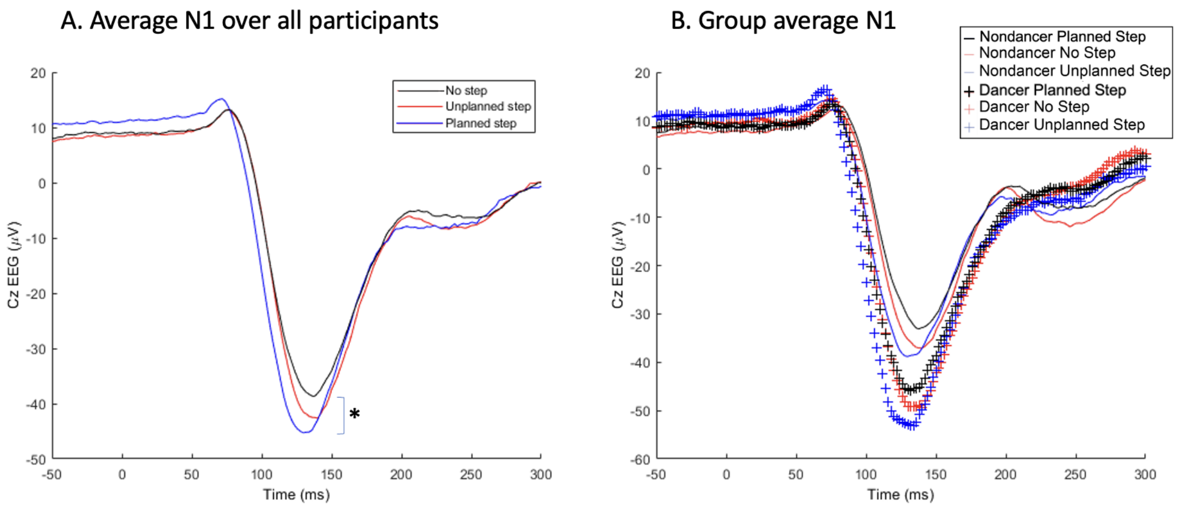

Figure 3.1: Cz electrode traces over time in each perturbation categories. The negative peak in each trace represents the N1. A) For all participants taken together, the mean N1 amplitude is greatest in magnitude following unplanned steps and least in amplitude following planned steps ($p<0.01$). B) When the data is split into dancers vs. non-dancers, the dancers have visually larger N1 amplitudes than non-dancers in each condition. This result was not statistically significant.

3.0.3 N1 amplitude did not scale with balance ability

Four of five dancers had a BEAM score at or above the median of 19.08 ft. Within either group and as a whole, BEAM score did not scale with N1 amplitude in any perturbation condition (Fig. 3.2).

3.0.4 In exemplar data, beta decreased less in dancers post-N1

In the time-frequency domain, differences in beta power between groups were not significant in the 50-150 ms post-perturbation time bin, which encompasses the N1 response. Differences in beta power in the 150-250 ms time bin were also not statistically significant. In this time bin, the peak of the N1 response has already occurred yet upon visual inspection, non-dancers showed sustained decrease in beta power that the dancers did not. In Fig. 3.3, the initial large red peak represents an increase in beta at the time of the N1 response and the subsequent blue sections are decreases in beta.

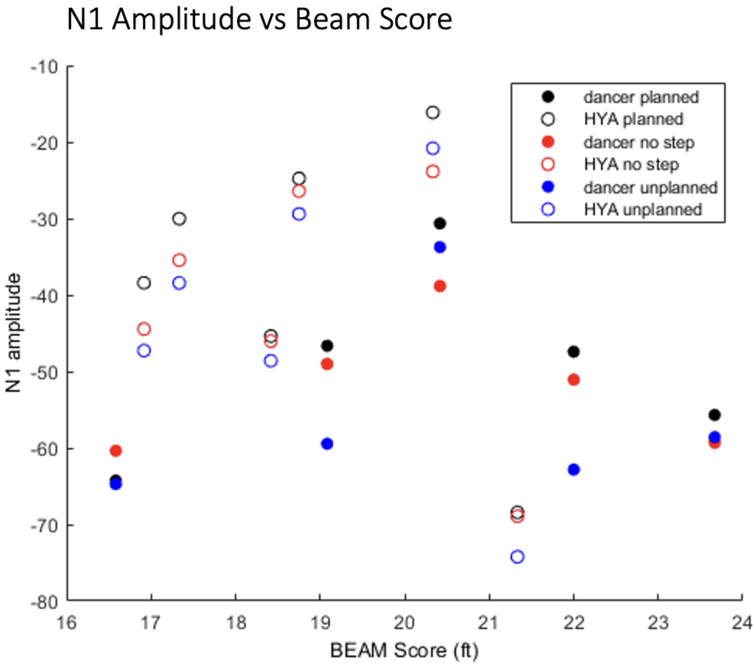

Figure 3.2: BEAM score vs. N1 amplitude for each individual and perturbation condition. Most dancer BEAM scores are at or above the median, yet do not show a positive correlation with N1 amplitude. Neither dancers' nor non-dancers' BEAM scores were correlated with N1 amplitude in this cohort.

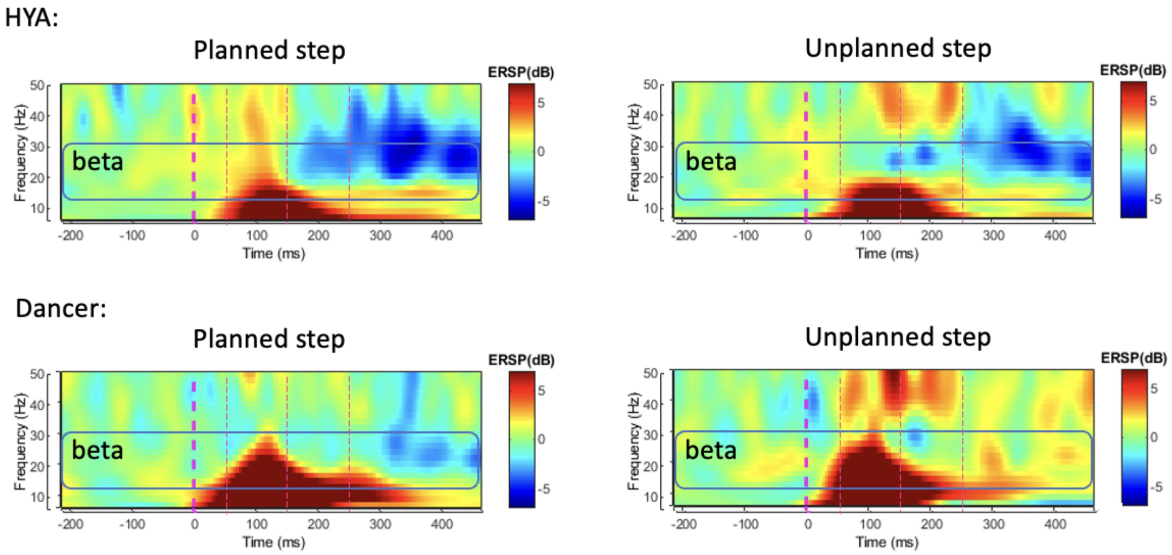

Figure 3.3: Event related spectral perturbations (ERSPs) for an exemplar control and dancer participant following planned and unplanned step perturbations at time 0. The beta band is 13-30 Hz, and colors represent increases or decreases in beta power from baseline according to the scale to the right of each ERSP. The 50-150 ms and 150-250 ms time bins are denoted with dotted lines.

Chapter 4

Discussion and Conclusions

4.1 Discussion

In this study, we investigated differences in cortical activity during reactive balance in professional modern dancers compared to non-dancer HYA. We used step thresholds to control for perturbation difficulty across participants undergoing support-surface perturbations. I hypothesized that dancers would require less cortically mediated sensorimotor integration to recover balance, predicting they would have smaller N1 amplitudes and a smaller change in beta post-perturbation, scaling with perturbation difficulty.

4.1.1 N1 amplitudes scaled with perturbation difficulty and expectedness

Across participants, N1 amplitudes were largest following unplanned step trials, followed by no step trials, and planned step N1s were smallest. This is in line with my predictions because planned step perturbations were smaller in magnitude than unplanned step perturbations. However, planned step perturbation trials were the only perturbation type administered when the participant was instructed to take a step regardless of step necessity. No step trials occurred when the instruction was not to take a step and thus were given in the same block as unplanned step trials. Even though the balance challenge for no step and planned step trials were the same, participant expectations regarding which type of perturbation they were going to receive next were more varied in the no step condition. While previous work has found that prior planning of a step does not have an effect on N1 amplitude [14], predictability may [15]. Since the N1 is an error assessment signal, the possibility for error in the expected outcome of a perturbation is much greater in blocks of trials where there is uncertainty regarding the magnitude and subsequent balance correcting response of a perturbation.

4.1.2 N1 amplitudes trend greater in dancers than nondancers

All dancers had larger N1 amplitudes than the average HYA N1 in each condition. Although not statistically significant potentially due to small group size, the observed trend differs from my prediction that N1 amplitudes would be smaller in dancers. Since each dancer had a BEAM score above the median score (Fig. 3.2) and N1 amplitude has been shown to be smaller in HYA with better balance [29], I expected dancer N1 amplitudes to be smaller than non-dancer HYA. The N1 response has been thought to be a potential biomarker of increased cortical engagement following balance perturbation, indicating worse balance ability. While dancers may not fear a fall, they may be more responsive to small threats to their balance and respond accordingly. Therefore, N1 amplitude may be an indicator of attentiveness to postural perturbation, and increased in both professional dancers and HYA with poor balance potentially, creating a U-shaped function across balance ability. In terms then of ”cortical resources” recruited to assist in balance recovery, dancers appear to have an initial cortical response greater than nondancers, differing from my hypothesis that dancer balance recovery would be more automatic than cortical.

4.1.3 N1 amplitude did not scale with balance ability

Dancers had generally better balance than nondancers as their BEAM scores were all at or above median. However, they did not follow the expected trend in HYA where balance ability scales with N1 amplitude [14]. Even though dancer N1 amplitudes were not statistically larger in magnitude than non-dancers', the lack of expected association between BEAM score and N1 demonstrates that dancer neurophysiological responses to balance perturbations may be different from other HYA due to factors other than better balance.

4.1.4 In exemplar data, beta decreased less in dancers post-N1

Changes in beta between dancers and nondancers were not significant in the time bin that encompassed the N1 peak, which is in line with the above lack of significant difference between N1 amplitudes between groups. In the time bin following the N1 peak, the between group difference was still not significant but showed a visually observed trend. Dancers trended toward having a smaller decrease in beta in the 150-250 ms time bin. This result matches my prediction that dancers would require less cortical sensorimotor integration to recover balance, manifesting neurophysiologically as a smaller change in beta power. The combination of trends toward larger N1 amplitudes and smaller changes in beta in dancers presents an interesting interpretation of dancer balance. The larger N1 may represent a fine-tuned sensitivity to perturbations to balance, but the smaller decrease in beta shows that once the balance error has been assessed, less cortical resources are recruited to correct balance. Dancers may have greater cortical error assessment than nondancers, but during their balance recovery utilizes less cortical resources. Their training may have an effect on attentiveness to changes in balance, resulting in greater cortical activity in the error assessment stage of the balance respond. Then, since their balance correcting actions are more automatic, the cortical response after their initial error assessment may be less cortically mediated.

4.1.5 Limitations of EEG

While EEG is a widely used technology for non-invasive data collection with human subjects, there are some key trade-offs to consider when drawing conclusions. EEG has high temporal resolution which allows us to analyze the time course of brain activity. However, each electrode can pick up activity from sources throughout the brain, so there is low spatial resolution [12]. To combat this, we localize our data to the Cz electrode which is located on top of the head, over the supplementary motor area. It is also where we see the N1 response most clearly. Since we see the cortical N1 at Cz, this electrode does pick up cortical activity following perturbations. However, brain activity from distant regions of the brain is also picked up by Cz, and conversely cortical activity is picked up by electrodes further from Cz. Analyzing EEG data in the source space rather than the electrode space will be helpful in further analysis. Using independent component analysis to group electrode activity into key sources that are comprised of electrical activity from multiple electrode channels can give results with reduced noise [30]. However, since the N1 response can still be split across several independent components while Cz is closest in physical proximity to the supplementary motor area, we chose to stay in electrode space for this N1 analysis.

4.1.6 Cohort Size

Due to the fixed timeline of this project, at the time of presentation, both experimental groups have less than 10 participants. This study relied on human participants who took time out of their day to participate in a 4 hour data collection in addition to at least two lab members not including myself who were crucial for experimental setup and execution. In addition to navigating the schedules of many to make this work possible, we also ran into a pre-processing pipeline issue that stalled data collection for one full semester until the bug was identified and resolved. All of this together contributed to the current study being underpowered. Small group size has implications for statistical significance. It is difficult to determine if data points are creating a trend or just outliers when there are so few. For example, the trend of dancers having larger N1 amplitudes than nondancers was not found to be statistically significant because the standard deviation of N1 amplitude for both groups was large enough to block out the observed difference. However, the trends we have observed thus far are still enough to prompt continued collection of dancer data to determine if trends hold in a larger dancer cohort.

4.1.7 Sex Matching

Due to availability of participants and time constraints, the current control and experimental groups are not adequately sex matched. Sex differences in EEG data have not been studied extensively. There is evidence that overall amplitude of beta oscillations is grater in HYA females at rest, but whether this affects the change in beta in other conditions is unknown [31]. In older and impaired adult cohorts, females have been associated with greater instance of falls [32, 33, 34]. Dual task error has also been able to distinguish fallers from non-fallers in females better than in males [34]. Since there is some evidence of EEG sex differences in HYA and more evidence that sex is a predictor of fall risk, especially when cortical resources are over-compensating to control balance, we will focus on recruiting sex-matched participants as we continue this study in order to adequately determine any effect of sex on outcome measures.

4.1.8 Potential Implications for Balance Rehabilitation

Older adults (OA) have also been shown to have larger N1 amplitudes than HYA [35], which has been interpreted as evidence of their lower balance ability and potential fall risk. If OA are at greater risk for falls, then it may be advantageous to be more sensitive to balance perturbations, resulting in a larger N1 amplitude. A sensitive error assessment might then be helpful for maintaining a balance rather than an indicator of poor balance. The N1 response alone may not be enough to assess balance ability or fall risk of an individual, since dancers trended toward a larger N1 but subsequent less recruitment of cortical resources governing the motor output of the balance-correcting response. Since dance therapies are being used for balance rehabilitation in motor impaired populations with success [36, 37, 38], knowing how dance affects neurophysiological responses is helpful for assessing individual progress during rehabilitation. Dance may cause cortical error assessment to be greater while simultaneously helping to improve balance. Therefore, aiming for individuals in rehabilitation to have an N1 response more similar to HYA may not be a useful or achievable goal. Instead, a greater N1 response coupled with balance rehabilitation may assist in improved balance because an individual is more sensitive to a potential balance perturbation and more equipped to counteract it successfully.

4.2 Conclusions

The aim of this project was to identify whether modern dancers recruit "cortical resources" less than nondancers to recover balance. Analyzing EEG data in the time domain and time-frequency domain at face value have answered this research question both no and yes. If considering the N1 error assessment signal as evidence of "cortical resources", then dancer balance correction may actually be more cortically mediated than nondancers'. If considering changes in beta power following the initial N1 response, dancers may have a more automatic than cortically mediated balance-correcting response. Taken together, these results may show that dancers have an increased sensitivity to changes in their balance due to a larger N1 amplitude, being able to detect slight changes in their current posture using cortical resources. After this initial error assessment, the balance-correcting muscle response itself may be more automatic due to training of sensorimotor integration skills that allow for subcortically-mediated balance correction. These findings may impact the way that we think about balance rehabilitation and our definitions of appropriate evaluations of balance improvement or fall risk.

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