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ORIGINAL ARTICLE

Older adults with a history of falling exhibit altered cortical oscillatory mechanisms during continuous postural maintenance

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ABSTRACT

Background and Aim: The significant risk of falling in older adults 65 years or older presents a substantial problem for these individuals, their caretakers, and the health-care system at large. As the proportion of older adults in the United States is only expected to grow over the next few decades, a better understanding of physiological and cortical changes that make an older adult more susceptible to a fall is crucial. Prior studies have displayed differences in postural dynamics and stability in older adults with a fall history (FH) and those who are non-fallers (NF), suggesting surplus alterations that occur in some older adults (i.e., FH group) in addition to the natural aging process.
Methods: The present study measured postural dynamics while the FH, NF, and young adult (YA) groups performed continuous postural maintenance. In addition, electroencephalography activity was recorded while participants performed upright postural stance to examine any group differences in cortical areas involved in postural control.
Results: As expected, older participants (FH and NF) exhibited worse postural stability, as evidenced by increased excursion, compared to the YA group. Further, while NF and YA show increased alpha activity in occipital areas during the most demanding postural task (eyes closed), the FH group did not show any differences in occipital alpha power between postural tasks.
Conclusions: As alpha activity reflects suppression of bottom-up processing and thus diversion of cognitive resources toward postural centers during more demanding postural maintenance, deficits in this regulatory function in the FH group are a possible impaired cortical mechanism putting these individuals at greater fall risk.
Relevance for Patients: Impaired inhibitory function in older adults may impact postural control and increase their risk of falling. Interventions that aim at addressing cortical processing deficits may improve postural stability and facilitate independent living in this population.

1. Introduction

Normal postural stability enables fine-tuned dynamic actions and movement in everyday settings [1]. An intact postural system can adeptly compensate for intrinsic and external perturbations for an individual to safely and efficiently navigate the natural world [2]. Therefore, not only does normal postural regulation promote dynamic and flexible control over muscular tone required for maintaining balance and upright stance, it is also imperative for reducing someone's risk of falling. Indeed, balance disorders (e.g., cerebellar ataxia and vestibular disorders) present with increased postural sway and uncoordinated movement that dramatically increase the prevalence of falls [3,4]. The natural aging process also significantly increases risk of falling as 40% of adults 65 and older report a fall incident at least once a year [5-7]. Increased fall risk in older adults is likely due to impoverished postural capacity

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as a consequence of the natural aging process [8,9]. However, the complexity and dependence on multiple systems (e.g., visual, proprioceptive, and vestibular) make it difficult to parse out the targeted mechanisms and/or regions of the brain responsible for that instability.

While older adults generally present with impoverished multisensory processing capacity, those with a fall history (FH) demonstrate more severe multisensory impairments [2,10] possibly suggesting a shared, global deficit leading to impoverished postural and multisensory processing capacity. For instance, older fallers display an increased susceptibility to the sound-induced flash illusion compared to older non-fallers [10-12]. When a single flash-beep pair is presented close in time (i.e., 70 ms) to a second beep, the perception of an illusory second flash can be elicited, also known as the fission illusion (the fission of 1 physical flash into 2 perceived flashes). The more temporally disparate the second beep is to the flash-beep pair (the greater the stimulus- onset asynchrony or SOA), the reduced likelihood of the illusion. However, while young adults (YAs) display a significant decrease of the illusion at larger SOAs (≥110 ms), the illusion is maintained in healthy older adults with older fallers exhibiting even greater illusion rates at this SOA (110 ms) and larger SOA values (270 ms) [10].

The enhanced illusion rates at larger SOA levels in older fallers reflect more severe deficits in their sensitivity of multisensory temporal processing compared to older non-fallers, possibly due to global changes in cortical processing influencing both the sensory and postural systems. Indeed, older adults with impaired multisensory processing present with abnormal limits of postural stability and use degraded strategies for postural stabilization (i.e., sway at hips rather than ankles) as compared to those with normal multisensory perception [2]. Further, while both older adults with and without a history of falling showed improved balance after an intensive balance training program, only older fallers also demonstrated improved sensitivity for multisensory integration [11]. These findings not only suggest a shared global deficit resulting in impaired multisensory processing and postural control but implicate reduced cortical capacity to handle postural and cognitive tasks simultaneously in older adults with a greater risk of falling. In another study that had participants perform the sound-induced flash illusion while maintaining a continuous, upright stance, older adults with a FH not only had increased illusion susceptibility but also showed increased postural sway during illusory (one flash and two beeps), but not during control (two flashes and two beeps) trials [12]. The more challenging illusory task thus detracted from these individuals' ability to exert appropriate postural control.

Studies investigating the cortical oscillations involved in these types of tasks help shed some light on the underlying drivers of such deficits. Alpha activity (8–13 Hz), normally thought to reflect top-down control of cortical resources and processing [13], was reduced in older adults compared to YAs during dual-tasking (maintaining an upright stance on an unstable surface while performing a 2-back visual working memory task) [14]. Alpha activity was also reduced in older fallers compared to non-fallers during postural perturbation tasks [15] suggesting that aging, and fall risk especially, degrades inhibitory control over occipital processing, a normal functional strategy to divert more cortical resources toward postural centers [14]. Bursts of gamma activity (30–50 Hz) at frontocentral and parietal sites have also been reported in YAs approximately 200 ms before maximal anterior sway at the ankle joint [16]. Therefore, gamma power can also be a useful indicator of active, bottom-up processing of somatosensory or vestibular inputs mediating postural stability.

In addition to examining the cortical oscillatory activity during these more dynamic types of postural tasks, center of pressure (CoP) can be collected while participants maintain a continuous (i.e., 60 s), balanced stance and provide a thorough depiction of postural stability and system dynamics. CoP is also one of the few reliable predictors available for predicting a fall in the older population [8,17-20]. For example, the degree of excursion (increased sway) is often greater in older adults with a history of falling compared to both YAs and to older non-fallers indicating impaired postural stabilization strategies [8,9,19]. The efficiency and overall stability of postural systems, measured by the root mean square (RMS) of excursion and velocity, can also distinguish between older adults without a FH and those with various balance disorders (for review [21]). Thus, it is a necessary next step to identify the related cortical processes and how they differ between older fallers and non-fallers.

The present study examines oscillatory dynamics in YAs, older adults without a history of falling (non-fallers; NF), and older adults with a recent history of falling (FH) while maintaining upright stance under varying levels of postural demand, such as eyes open (EO) versus eyes closed (EC). Specifically, we estimated alpha and gamma power across different cortical regions (frontal, motor, parietal, and occipital) and compared the strength of these oscillations between conditions and groups. As alpha activity is thought to reflect top-down control of sensory processing to direct resources toward reliable postural centers, we predicted reduced alpha power in the FH group. Further, reductions in gamma power over parietal areas, indicative of impaired processing of proprioceptive information, were expected in FH. As CoP estimates provide comprehensive information about a postural system's dynamics, this study was most interested in linking postural dynamics with these measures of cortical function. In line with prior findings, we expected FH group to have severe deficits in CoP data (i.e., increased excursion) and that reduced postural stability will be associated with decreased alpha activity across cortical areas compared to the NF and YA groups. In addition, we were interested in multisensory temporal processing in FH compared to NF and YA and the relationships between multisensory sensitivity and postural dynamics. While prior studies have used the sound-induced flash illusion to estimate multisensory processing in older fallers, we were more interested in examining the temporal binding window (TBW), a construct used to define the temporal limits within which two stimuli are likely to be perceptually bound. Healthy older adults exhibit broader TBWs [22,23], helping to explain the age-related increase in illusion susceptibility, however, TBWs of older fallers have not

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previously been reported. Due to their increased illusion rates at larger SOA levels, we expected broader TBWs in FH compared to NF and hypothesized this reduced multisensory efficiency to be associated with impoverished postural function in FH adults.

2. Materials and Methods

2.1. Participants

Twenty-four YAs (24.16 ± 3.86 years, 15 males), 24 older adults without any history of falling (non-fallers; NF) (69.93 ± 3.50 years, 10 males), and 16 older adults with a recent FH (73.20 ± 3.28 years, five males) participated in this study. One older NF did not participate in any of the postural experiments due to attrition. A FH was identified as the individual experiencing at least one fall in the 18 months preceding experimentation as determined by self-report. A fall is defined as having unintentionally coming to rest on the ground or another lower level not as the result of a major intrinsic event or overwhelming hazard [24]. To be included in the NF group, older adults could not have any history of a fall, regardless of the recency.

To ensure normal or corrected-to-normal sensory function, all participants were screened for normal hearing using AudioScope 3, a screening audiometer (Welch Allyn, Skaneateles Falls, NY, USA), and were required to have a pure tone threshold lower than 40 (for older adults) or 25 (for younger adults) dB for 1 and 2 kHz in both ears. Participants self-reported normal or corrected-to-normal vision and excluded from experimentation if they reported any major visual disorders (i.e., glaucoma). Other exclusion criteria included history of neurological disorders or disease, seizure disorder, brain injury, and use of antipsychotic medications. To account for any vestibular or musculoskeletal problems that could contribute to an individual's risk of falling, participants were asked to report any chronic pain, use of pain medications, recent musculoskeletal injuries, or any vestibular disorders. Finally, older adults were required to score ≥26 on the Montreal Cognitive Assessment to control for any potential cognitive decline [25].

Participants provided signed informed consent before any experimentation and were financially compensated for their time. The experimental protocol was reviewed and approved by the Institutional Review Board at the University of Nevada, Reno.

2.2. Temporal order judgment task

To estimate factors influencing participant's multisensory temporal processing (i.e., the TBW), participants performed the Temporal Order Judgment task. A black fixation cross was presented in the center of the screen throughout the entire experimental paradigm. Stimuli were generated using MATLAB (MathWorks, Natick, MA, USA) and Psychtoolbox extensions [26,27]. The auditory stimuli were pure tones of 1000 Hz created in MATLAB and presented binaurally at 70 dB (measured at the auditory source) through a speaker (Fantech HellScream GS 201, Nepal) directly under the center of the display to approximate the same spatial location as the visual signal. Visual and auditory stimuli were delivered through a Display ++ system with a refresh rate of 120 Hz and an AudioFile stimulus processor, respectively (Cambridge Research Systems, Rochester, UK).

On each trial, a 30 ms 1000 Hz beep and a 30 ms white circle with a diameter of 3.5° were presented at variable SOAs that varied from −420 to +420 ms in 30 ms steps where positive SOAs represent visual leading trials (top panel of Figure 1) and negative SOAs represent audio leading trials (bottom panel of Figure 1). Each SOA (29 total levels) was repeated 15 times for a total of 435 trials separated into five experimental blocks where the SOA order was randomized. Participants were required to take a 20 s break between experimental blocks and could take longer if they desired. At the end of each trial, participants responded as to which cue they perceived first with a "1" keyboard response indicating "flash first" and a "2" indicating "sound first." Trials were separated by a variable interval between 1200 and 1600 ms.

Before experimental blocks, participants performed a practice block to ensure they understood the task and responses. Four non-experimental SOAs were used in the practice block (400, −200, +200, and +400 ms) that was each repeated 3 times for a total of 12 practice trials.

For the audiovisual Temporal Order Judgment task, individual's proportion of "Flash First" responses was plotted as a function of SOA values and fit with a psychometric function. The mean of the fitted function was extracted as the point of subjective simultaneity, a measure of perceived synchrony or participant's bias in determining temporal order. In addition, left (audio-leading) and right (visual-leading) 75% thresholds were separately estimated and then summed together for a single, whole TBW value. Group average point of subjective simultaneity and TBW values were then calculated from individually analyzed data.

2.3. Functional and psychological assessments

To assess basic mobility and ambulation, participants performed the timed up and go test (TUG) before CoP collection [28]. Briefly, participants stood up from a chair, walked to and around a cone placed 3 m away, walked back to the chair, and sat down. Participants performed one familiarization TUG and one experimental TUG. Only their time from the experimental TUG was used for analysis. To quantify any psychological effects on an individual's postural control, the Falls Efficacy Scale-International was also completed before CoP collection to estimate the subject's fear of falling (FOF) [29]. Finally, participants indicated their current level of discomfort and pain before any postural maintenance experiment using the visual analog pain scale [30].

2.4. CoP collection

CoP data was collected in the Neuromechanics Lab at the University of Nevada, Reno, using an Advanced Mechanical Technology Inc. (AMTI) force platform (Watertown, MA, USA) and Qualisys Track Manager software (Qualisys Inc., Göteborg, Sweden). Participants were asked to stand with their hands by their sides wearing only socks and with their feet placed in a comfortable position (~2 inches between medial malleoli). Participants performed three 65 s trials for each upright stance

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condition, EO, EC, and verbal inhibition (VI), for a total of nine trials. These were separated into three experimental blocks so that each of the three conditions were performed in a random order before being repeated. At the start of each task, participants were asked to step to the center of the force platform and the recording began with the first 5 s being discarded to allow for postural adjustments and settling [21].

The VI task was a modified stop signal paradigm [31] in which a right or left arrow surrounded by a white circle was randomly presented on the display and participants were asked to verbalize the direction the arrow was pointing as quickly as possible (top panel of Figure 2). On 25% of trials, a stop signal was presented – the white circle surrounding the arrow would change to red – and participants had to inhibit their verbal response (bottom panel of Figure 2). At the start of each trial, a fixation cross was present on the center of the screen for 800 ms followed by the arrow within the white circle. On the stop signal trials, the circle changed to red 275 ms after the onset of the arrow. This stop signal delay of 275 ms was based on an average stop signal delay of young and old participants during a pilot phase. As verbal responses were given and not keyboard responses, the stop signal delay remained constant throughout the experiment unlike standard versions of a stop signal task where the stop signal delay is adjusted on each trial based on the participant's success or not of inhibiting their keyboard response.

2.5. CoP analysis

CoP data were preprocessed and analyzed using custom software in MATLAB R2019b. Data were preprocessed using the empirical mode decomposition method [32]. Briefly, empirical mode decomposition spectrally decomposes the time series signal into *n* intrinsic mode functions. Each intrinsic mode function is defined by a unique frequency reflecting distinct timescales within the CoP data allowing for filtering of information from some of these separable timescales [32]. Based on the intrinsic mode functions, a low-pass 10 Hz filter was initially applied to the data.

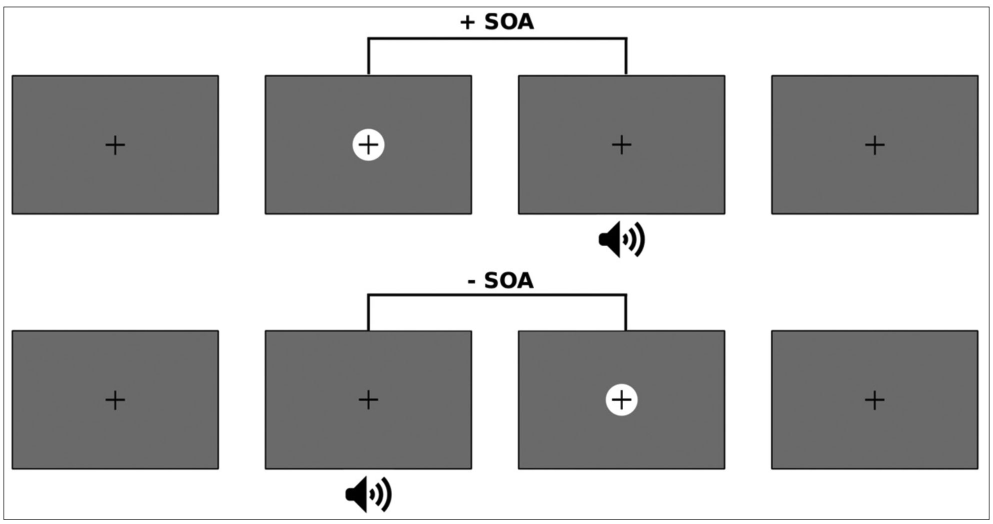

Figure 1. Experimental design of Temporal Order Judgment task. On each trial a visual stimulus (white circle) was presented in the center of the screen and a pure tone auditory stimulus was presented via speakers below the screen. The two signals were separated by variable stimulus onset asynchronies (SOAs). For trials with a positive SOA, the visual signal preceded the auditory signal (top panel) and for those trials with a negative SOA, the auditory signal preceded the visual signal (bottom panel). The fixation cross was presented in the center of the screen throughout the experiment.

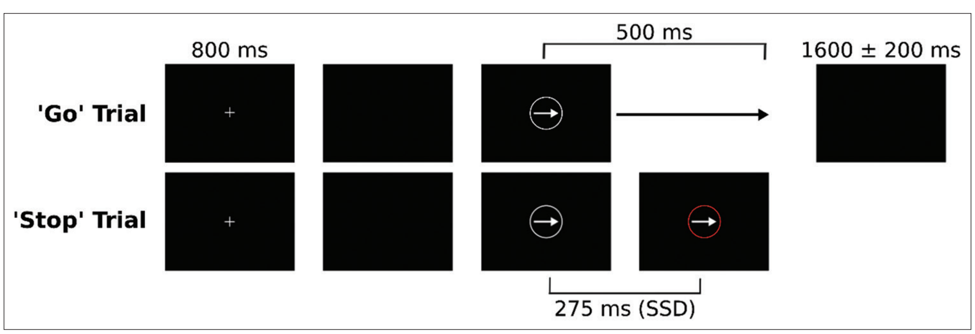

Figure 2. Experimental design for visual inhibition task. Each trial presented a white arrow inside of a white circle on the center of the screen for a maximum duration of 500 ms. Participants were instructed to respond as to the direction the arrow was facing as soon as the arrow was presented (Go trials; top panel), unless the circle surrounding the arrow changed to red (stop signal; bottom panel). This stop signal was always presented at a stop signal delay of 275 ms measured from the onset of the white arrow.

Then, the summed intrinsic mode functions were inspected for each 60 s trial for each individual and manually adjusted to remove electrical noise and non-stationarities at high frequencies from the resulting time series. This procedure was performed separately for anteroposterior (AP) and mediolateral (ML) CoP data. An average of 5.35 ± 0.66 and an average of 5.63 ± 0.77 intrinsic mode functions were retained in the AP and ML directions, respectively.

Custom software in MATLAB R2019b was used to quantify postural stability from preprocessed CoP data using excursion amplitude, the RMS velocity, and a complexity index (CI) estimated from multiscale entropy. These measures were computed for each task for each individual so that for one postural stability estimate, one participant had nine CoP estimates (3 estimates × 3 tasks). Excursion amplitude was calculated as the absolute average of the differences between two consecutive points on CoP path ([33]; Eq. 1), while RMS velocity was calculated from the square root of the average squared velocity across a 26-element range and then rectified by square rooting the data (Eq. 2). Note that these values were calculated separately for the AP and ML directions; Eq. 1. depicts excursion derived from CoP data in the AP direction only. The differences between two consecutive points on CoP path

$$Excursion = \frac{1}{N}\sum_{i=1}^{N}\left|COP(i+1) - COP(i)\right| \quad \text{Eq. 1}$$

$$RMS = \left(\frac{1}{N}\sum_{n=1}^{N}COP_n^2\right)^{1/2} \quad \text{Eq. 2}$$

Multiscale entropy method was used to determine a mean CI for each postural task for each participant [34]. Multiscale entropy was calculated on the excursion amplitudes. Briefly, sample entropy using a similarity criterion of 0.2 was computed for 20 separate and equivalently spaced timescales within the CoP data. The summed integral for each sample entropy value was then used as the final, CI value.

2.6. Electroencephalography (EEG) acquisition

To understand cortical dynamics associated with postural maintenance, the same postural experiment was repeated while high- density EEG data were recorded. Due to physical and system restrictions, the EEG version of the experiment had to be conducted separately from the CoP experiment in a separate building on the University of Nevada, Reno campus.

Participants performed the same three postural tasks while EEG data were continuously recorded from a 128 channel BioSemi Active 2 system (BioSemi, Amsterdam, The Netherlands). Note that this was separate from the CoP collection and thus CoP data from force plate were not recorded simultaneously with EEG data. In addition to the standard 10–20 electrode locations, this system included intermediate positions. Default electrode labels were renamed to approximate the more conventional 10–20 system (see supplementary Figure S1 in [35]). Four additional channels recorded electrooculography signals, two channels on the lateral sides of each eye to detect horizontal movement and two channels above and below the right eye to detect vertical movement (i.e., blinks). EEG was sampled at a rate of 512 Hz and processed offline using EEGLAB (v.14_0_0b) and ERPLAB (v.6.1.3) with MATLAB R2013b (MathWorks, Natick, MA, United States).

2.7. EEG pre-processing

EEG data were initially preprocessed with a high-pass 0.5 Hz and low-pass 125 Hz Butterworth filter and a narrowband (59.5–60.5 Hz) filter was applied to remove line noise. Next, peripheral channels (TP8, TP8h, C6, TH8, T8, FT8, FT8h, F6, F8, F5, F7, FT7, FT7h, C5, T7, T7h, TP7, and TP7h) that are prone to facial and cranial muscle activity contamination were removed to reduce high- frequency noise before analysis [36]. Remaining channels were further inspected for artifacts using a threshold of ±200 μV and the TrimOutlier plugin (v.o.17). An average of 1.35 (±2.14) channels was removed after this artifact detection and spherically interpolated across all participants. Next, EEG data were re-referenced to the common average and segmented into 1s non-overlapping trials within each of the three postural conditions. Epochs were visually inspected for artifacts and an average of 22.37 (±17.58) trials was selected for rejection across participants. Blinks, eye movements, and cranial artifacts were corrected in the epoched data using independent component analysis. Extremely poor signal- to- noise ratio occurred for a number of participants due to the highly sensitive nature of the high-density 128 channel system, and the specific experimental set up requiring participants to stand throughout data collection. This resulted in EEG data from 19 YA, 19 NF, and 13 FH being included for further analysis.

2.8. Power spectrum analysis

Power spectra were estimated separately for each condition using a short time Fourier transform on discrete temporal windows sliding in 25 ms steps. This procedure was performed separately for two frequency ranges: A low- frequency band (3–35 Hz) and a high- frequency band (35–90 Hz) [37]. For the low- frequency band, data were initially tapered with a Hanning window and the power spectra were computed with a time window of 400 ms [37,38]. The high- frequency band was computed using a multi-taper method in which we applied 8 Slepian tapers and used a 200 ms window [38]. Power spectra were estimated separately for each tapered temporal segment for each trial and then averaged.

Subsequent analysis computed the average alpha (7–13 Hz) and gamma (35–60 Hz) for each individual within four distinct ROIs: Frontal, motor, parietal, and occipital. Individual alpha power was averaged across channels that comprised each ROI. Specifically, the frontal ROI was comprised of 13 channels (FFC2, F2, AFF2, AFF4h, AFz, AFFz, Fz, FFCz, FCz, FFC1, F1, AFF1, and AFF3h) [14,36], the motor ROI was comprised of 16 channels (Cz, C2h, C2, C4h, C4, FC4, FC4h, FCC2h, FCC2, FCC1h, FCC1, FC3, FC3h, C1, C1h, and C3h) [39,40], the parietal ROI was comprised of 21 channels (P1, CPP3, PPO3, PPO5, PO3h, PPO1, Pz, PPOz, POz, PO4h, PPO2, P2, CPP4, PPO4, PPO6, P8, P6, CPP6h, CPP5h, P5, and P7) [14,36], and the occipital ROI was comprised of nine channels (PO11, PO1, POO5, POOz, Oz,

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OIz, POI2, O2, and POO6) [41,42]. ROIs and their corresponding channels are illustrated in Figure 3.

2.9. Statistical analysis

As data from the audiovisual Temporal Order Judgment task violated parametric assumptions, Kruskal–Wallis tests and *post hoc* Wilcoxon rank sum tests were performed to determine group differences in TBW and PSS estimates.

One-way ANOVAs were conducted on TUG and FOF measures to estimate group differences. To analyze CoP data, excursion, RMS velocity, and CI values from each trial (3 trials/condition) for each individual were inspected and outliers that were ± 1.5x the interquartile range were removed. The remaining data were averaged across trials from each condition for each subject, resulting in a single excursion value for each condition/subject. The remaining data were as follows: EO condition – 22 YA, 22 NF, and 16 FH; VI condition – 23 YA, 22 NF, and 15 FH; and EC condition – 22 YA, 21 NF, and 13 FH.

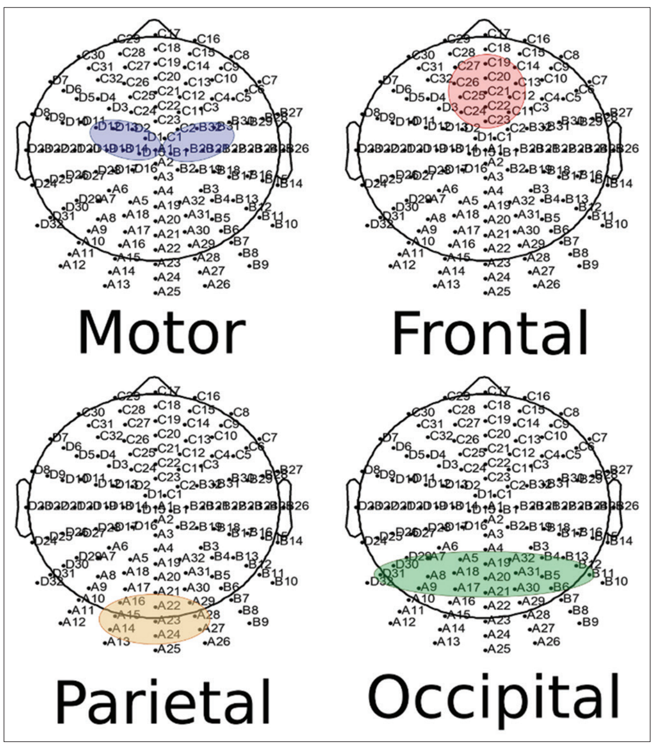

Figure 3. ROIs and their corresponding channels.

Linear mixed models were then conducted as this type of analysis is robust to incomplete cases due to missing conditions that occur following outlier removal. Three separate models were performed using excursion, RMS velocity, and CI values as dependent variables of the models. Group (NF, FH, and YA), direction (AP vs. ML), and postural task condition (EO, EC, and VI) were used as fixed effects consistently across the three models while participants were set as a random effect to account for variability across subjects. As the power spectrum data violated the normality assumption, separate Kruskal–Wallis tests with Holm-Bonferroni corrected *P*-values were performed to examine group differences within each ROI, for each postural task. Within each group, spearman rho correlation (r_s) analysis was conducted to examine relationships between CoP and gamma/alpha power. Statistical tests were performed in R statistical software and the linear mixed model analysis was conducted using the nlme package [43].

2.9.1. Correlation analysis

To determine associations between the multisensory and postural behavioral and oscillatory measures recorded from this project, we generated correlation matrices for each group using Spearman rho correlation coefficients. As some individuals were missing data from some, but not all variables, we employed pairwise deletion. Correlation analysis was performed in R version 4.0.3 statistical software using the *rcorr* function from the Hmisc package [44].

3. Results

3.1. Multisensory temporal processing deteriorates with age

Figure 4 displays the group- averaged data fit with psychometric functions (left panel) along with the estimated TBW (middle panel) and point of subjective simultaneity (right panel) measures. Kruskal–Wallis tests showed that there was no group difference in point of subjective simultaneity estimates (X^2 (2) = 0.31, P = 0.86); however, there was a significant main effect of group on TBW measures (X^2 (2) = 10.02, P < 0.01). Follow- up pairwise comparisons using Wilcoxon rank- sum tests showed that the FH and NF groups had significantly wider TBWs than the YA group

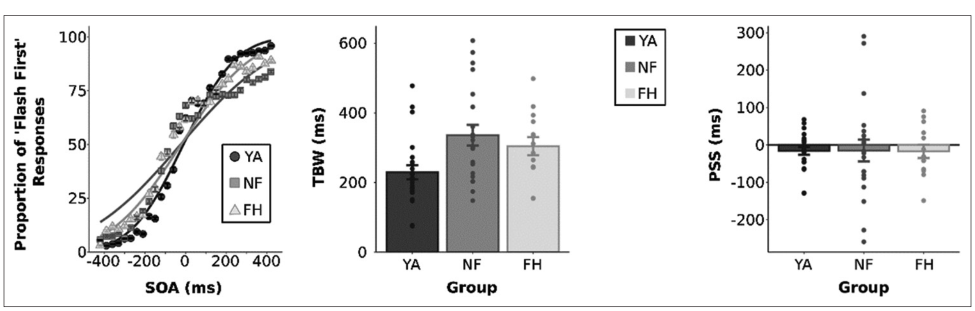

Figure 4. Group averaged data fit with psychometric functions (left panel) along with the estimated TBW (middle panel) and point of subjective simultaneity (right panel) measures.

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(all $P < 0.05$) while there was no difference between the FH and NF groups ($P = 0.75$).

3.2. FH demonstrate reduced mobility and enhanced FoF

Next, we wanted to quantify any group differences between general mobility and FOF. There was a significant effect of group on times to complete the TUG test (F [2,60] = 7.84, $P < 0.001$) with *post hoc* pairwise comparisons revealing that the FH group had significantly longer TUG times (10.4 ± 1.9 s) compared to both the YA (8.8 ± 1.1 s) and NF groups (8.9 ± 1.3 s) (*adjusted* $P < 0.01$) while there was no significant difference between YA and NF (*adjusted* $P = 0.98$). Similarly, there was a significant effect of group on FOF scores (F [2,59] = 4.52, $P < 0.05$) with the FH group showing significantly larger scores than both the NF and YA groups (*adjusted* $P < 0.05$).

3.3. Increased excursion in the AP direction for the FH and NF groups

There was also no significant difference between groups (F [2,60] = 1.01, $P = 0.37$) on visual analog pain scale [30] scores indicating that pain did not influence postural stabilization. The trajectory of CoP data from a representative participant for each group is plotted for all three postural conditions (EO – top panel; VI – middle panel; and EC – bottom panel) in Figure 5. From initial visual inspection, it is apparent that postural sway increases as the task demand increases. In addition, the amount of postural sway increases from YA to both the NF and FH groups.

Figure 6 shows the average excursion for each group and for each condition. A linear mixed model using group, task and direction as fixed effects, and a 3-way interaction revealed that excursion values did not significantly differ for FH compared to YA ($b = 0.001$, $t(59) = 0.40$, $P = 0.69$) or to NF groups ($b = 0.003$, $t(59) = 1.48$, $P = 0.15$), nor was there any significant difference for NF compared to YA ($b = 0.002$, $t(59) = 1.18$, $P = 0.24$). There was a significant main effect of task with excursion significantly increasing in EC condition relative to EO condition ($b = 0.004$, $t(277) = 3.50$, $P < 0.001$), and significantly increasing in EC relative to VI condition ($b = 0.003$, $t(277) = 2.62$, $P < 0.01$). However, there was no significant difference between VI and EO task ($b = 0.001$, $t(277) = 1.04$, $P = 0.30$). While the excursion was not significantly affected when moving in the ML compared to the AP direction ($b = 0.0005$, t (277) = 0.38, $P = 0.70$), there were significant interactions between direction and group. Specifically, in the AP direction only, excursion values were larger for FH compared to YA ($b = 0.004$, $t(277) = 1.99$, $P < 0.05$) and larger for NF compared to YA ($b = 0.005$, $t(277) = 2.55$, $P < 0.05$) groups. Overall, this model revealed that the EC condition had the largest excursion values regardless of group and there was significantly increased sway by the older adults (FH and NF groups) in the AP direction.

3.4. RMS velocity increases with difficulty of postural task

Next, a linear mixed model was performed to understand how group, task, and direction affected RMS velocity (Figure 7).

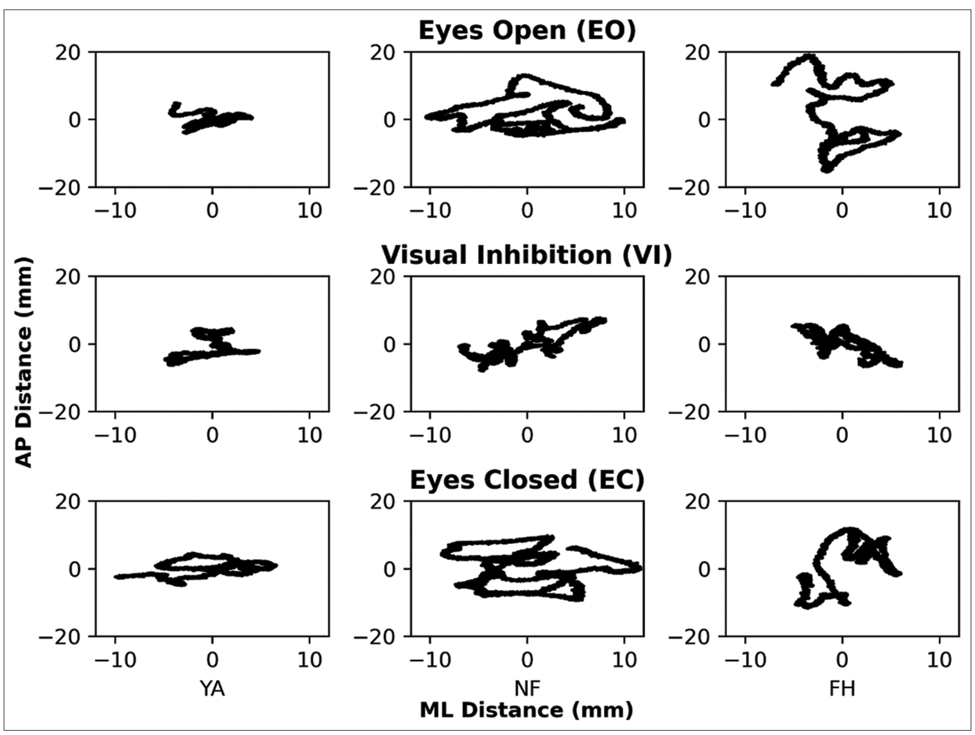

Figure 5. Stabilograms across conditions. The time series of CoP data are plotted for the eyes open (EO; top row), visual inhibition (VI, middle row) and eyes closed (EC, bottom row) conditions for a representative participant from the young adult (YA, left column), non-faller (middle column) and fall history (right column) groups. Anterior-posterior (AP) sway is shown along the y-axis while medio-lateral (ML) sway is shown along the x-axis.

The final model included all three main effects and a significant interaction between group and direction. The effect of group was not significant across any comparisons (all *b values* ≥0.002, *all t values* (59) ≤1.15, all *p values* >0.26). However, both the VI (b = 0.19, t (277) = 2.10, P < 0.05) and EC (b = 0.40, t (277) = 4.36, P < 0.001) conditions significantly increased RMS velocity estimates compared to EO and the EC condition significantly increased RMS velocity compared to VI task (b = 0.21, t (277) = 2.31, P < 0.05). Further, the AP direction significantly increased RMS velocity compared to ML direction (b = 0.23, t (277) = 2.51, P < 0.05). The only significant interactions present revealed that the EC condition significantly increased RMS velocity relative to EO (b = 0.37, t (277) = 2.84, P < 0.01) and to VI (b = 0.33, t (277) = 2.57, P < 0.05) tasks during the AP direction, not ML. While this model did not show any significant difference between groups in RMS velocity measures, there was an expected increase in RMS velocity with increasing task difficulty in the AP direction.

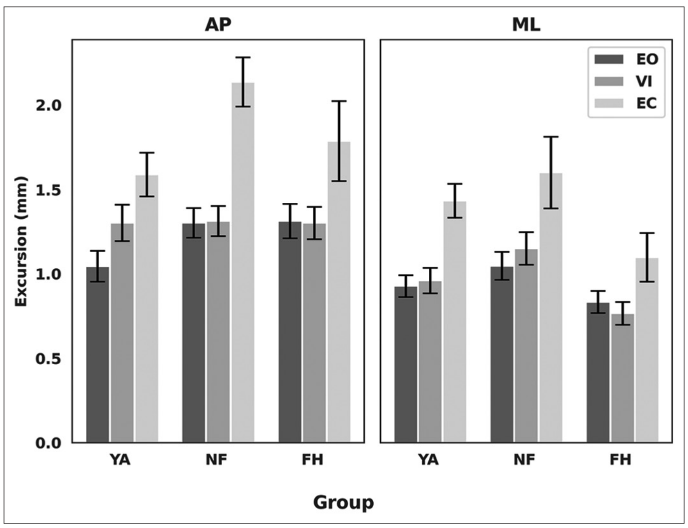

Figure 6. Average excursion values. Group-averaged excursion values are presented for the eyes open (EO; black bars), visual inhibition (VI; dark grey bars) and eyes closed (EC; light grey bars) conditions in the anterior-posterior (AP; left panel) and medio-lateral (ML; right panel) directions. ** Error bars represent SE values.

3.5. FH group did not show a significant impairment in postural complexity

The final postural stability measure, CI (data not shown), was best fit with a model that only included the three main effects, no interactions. The NF group had significantly larger CI values compared to the YA (b = 0.48, t (59) = 2.63, P < 0.05) while the FH group was not significantly different compared to YA (b = 0.25, t (59) = 1.24, P = 0.22) or NF (b = 0.23, t (59) = 1.13, P = 0.26). There were no significant effects of VI condition (b = 0.11, t (289) = 1.22, P = 0.22) or EC condition (b = 0.10, t (289) = 1.08, P = 0.28) relative to EO or of VI relative to EC (b = 0.01, t (289) = 0.11, P = 0.91). However, CI values were significantly greater in the AP versus the ML direction (b = 0.57, t (289) = 8.03, P < 0.001). Contrary to our prediction, the NF group had significantly larger CI values relative to YA while CI values from the FH group did not differ from either YA or NF.

3.6. Increased occipital alpha power for NF and YA, not FH, during EC posture

In addition to collecting CoP data while participants performed postural tasks, a separate experiment collected EEG data while participants repeated the same postural conditions. Average alpha power (7–13 Hz) was computed within the 4 ROIs for each participant, separately for each task. Separate Kruskal–Wallis tests with Holm-Bonferroni correction were performed to examine the effect of group within each ROI, for each condition. There were no significant differences between groups for any condition in occipital, motor, and parietal ROIs (X^2 [2] ≤3.64, *adjusted P-values* >0.9). While the group effect did not survive multiple comparisons correction (X^2 [2] ≥6.98, adjusted P-values ≤0.3), it is visible from Figure 8 that FH showed relatively lower alpha

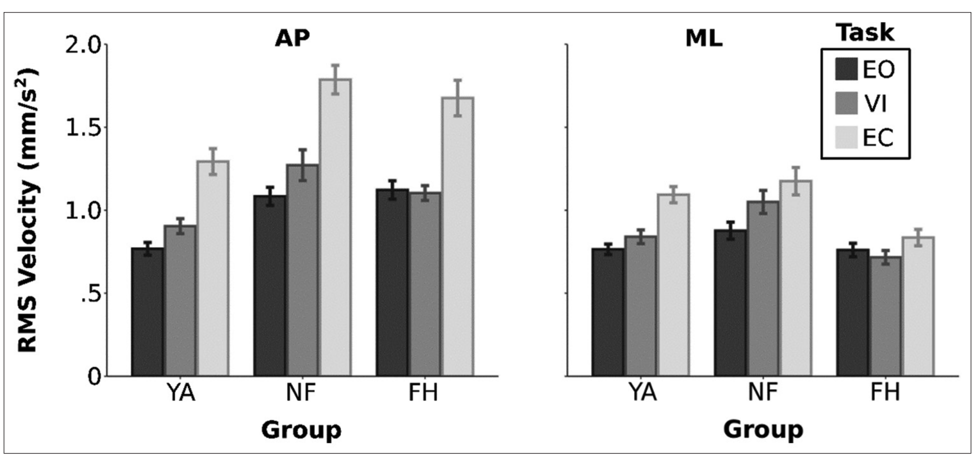

Figure 7. Average RMS velocity. Group-averaged RMS velocity values are presented for the eyes open (EO; black bars), visual inhibition (VI; dark grey bars) and eyes closed (EC; light grey bars) conditions in the anterior-posterior (AP; left panel) and medio-lateral (ML; right panel) directions. ** Error bars represent SE values.

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power in frontal ROI compared to NF or YA across conditions. The same analysis was conducted for average gamma power (35–60 Hz, data not shown) and there was no significant effect of group for any condition within any ROI (X^2 [2] ≥3.91, all adjusted *P-values* >0.9).

We also examined the effect of postural task on alpha power within each group and within each ROI using Holm-Bonferroni adjusted *p* values. As shown in Figure 8, in the occipital ROI, there was a significant effect of postural task for YA (X^2 [2] = 21.64, adjusted $P < 0.001$) and for NF (X^2 [2] = 11.30, adjusted $P < 0.05$), but not for the FH group (X^2 [2] = 9.10, adjusted $P = 0.11$). There was no significant effect of task on alpha power estimates in frontal, motor, or parietal ROIs for any group (X^2 [2] ≤7.84, all adjusted *P-values* 0.18).

3.7. No significant correlations between oscillatory activity and CoP measures in the AP direction

To assess relationships between oscillatory activity and CoP estimates, we computed Spearman's rho correlation coefficients between alpha power in each ROI and the CoP measures estimated from the force plate procedure for each group and each task. We restricted correlations to the AP direction as CoP measures only showed significant group and task effects in this direction. A Bonferroni adjusted *P*-value of 0.006 (0.05/9; 9 dependent comparisons for each group) resulted in the loss of significance and no significant correlations were found between alpha power and CoP measures. Similar null results were found when the same correlation analysis was performed to examine the relationship between gamma power and CoP measures. Note that in the present study, EEG data were collected separately from the force plate data due to physical and system restrictions. This limitation might have contributed to our failure of finding significant correlation between oscillatory activity and CoP measures.

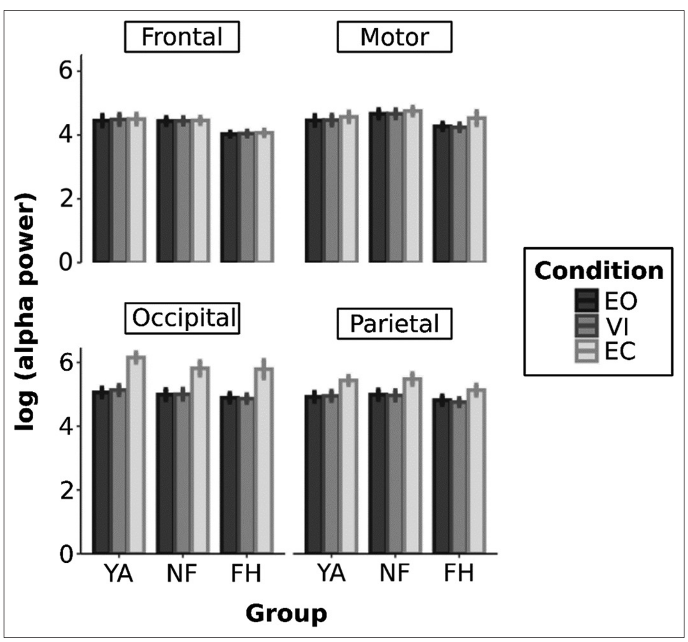

Figure 8. Group-averaged alpha power across conditions and ROIs. The group-averaged alpha power estimates are presented for the eyes open (EO; black bars), visual inhibition (VI; dark grey bars) and eyes closed (EC; light grey bars) conditions. Power values were extracted from the frontal (left top panel), motor (right top panel), occipital (left bottom panel) and parietal (right bottom panel) ROIs. ** Error bars represent SE values.

3.8. Relationships between postural control and multisensory temporal processing

To assess relationship between multisensory temporal processing and postural control, correlation between TBW and TUG was computed for each group. There was no significant correlation found in YA (r_s = 0.13, P = 0.54) or NF (r_s = 0.32, P = 0.15) group. Only the FH group showed a trend toward significance (r_s = 0.5, $P < 0.09$), worse mobility (i.e., larger TUG scores) was associated with wider TBWs (Figure 9).

4. Discussion

Majority of independent, community-dwelling older adults who have experienced a fall do not present with comorbid histories of cognitive, balance, or musculoskeletal problems [45,46]. Therefore, what distinguishes a healthy older adult who has had a history of falling from one that does not? Understanding and identifying these underlying factors will increase our knowledge of cortical and/or physiological changes that increase fall risk and may also provide useful predictive measures for quantifying a non-faller's likelihood of falling in the future. The present study examined three separate CoP measures used to quantify postural sway as well as estimates of multisensory temporal processing and if these two systems were related. Finally, we present cortical oscillatory activity recorded during continuous postural stance under variable conditions.

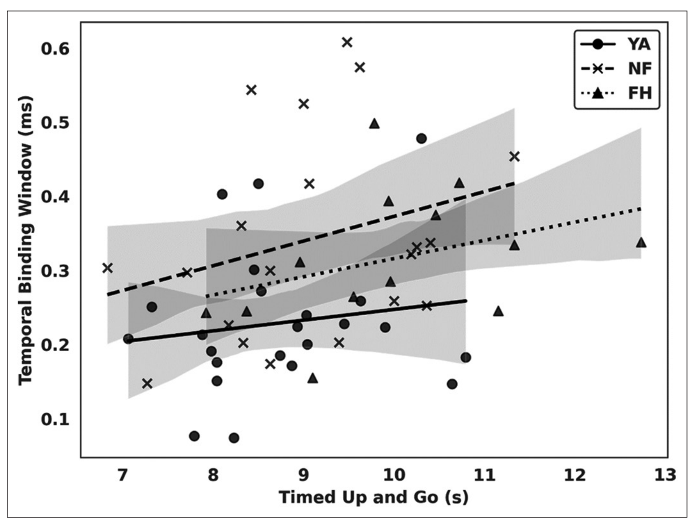

Figure 9. Relationship between timed get up and go test (TUG) and temporal binding window (TBW). Individual data with lines of best fit are shown for young adult (YA; circles with solid line), non-faller (NF; cross with dashed line) and fall-history (FH; triangles with dotted line) groups for TUG (x-axis) and TBW (y-axis). 95% confidence interval for the lines of best fit are shown as shaded regions surrounding the lines.

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Both FH and NF had significantly wider TBWs compared to YA indicative of more flexible, less precise perceptual binding capacity as a result of natural aging. However, both older fallers and non-fallers present with retained perceptual synchrony as compared to YA similar to prior findings suggesting compensatory age-related strategies to maintain perceptual constancy [47]. While these results indicate that sensitivity to temporal relationships between multimodal cues is degraded as a natural consequence of the aging process, similar to prior findings [22,47-49], older individual's presenting with a FH do not necessarily have enhanced multisensory impairments. However, the link between postural stability (TUG) and multisensory processing (TBW) was evident in the FH group, in line with prior findings [10-12]. For instance, worse general mobility (larger TUG scores) was associated with wider TBWs, suggesting that deficits in mobility are related to a general reduction in multisensory temporal precision. Overall, the relationships between impaired postural stability and declines in multisensory processing found in the FH group may implicate a common mechanism in these individuals that result in these extensive deficiencies.

The three separate postural conditions tested increased in difficulty level from EO to VI to EC. We expected the EC condition to be the most difficult, particularly for all older adults as this population displays increased reliance on the visual system for maintaining posture [50]. The VI condition was chosen as an intermediary difficulty level since performing a cognitive task while maintaining posture diverts cognitive resources away from postural centers increasing the amount of sway and variability in postural stabilization [21,50-52]. However, the present results show that task difficulty only affected RMS velocity during the EC condition eliciting the largest variability in velocity followed by VI and then EO, similar to prior findings [53].

The "Loss of Complexity" hypothesis describes reduced system adaptability and reactivity to external perturbations as a function of aging [54]. The group-averaged CI values ranged from 2.61 to 3.74, in line with a prior report using similar analysis parameters in YAs (CI: 1.26 ± 0.26) [55]. However, another study reported much higher CI values estimated from healthy older adults (CI: 9.15 ± 1.2) [53]. Multiscale entropy was performed in this project using scales 1–20 compared to the 2–8 used in Manor *et al*. [53] which could explain this variability. Regardless, results from the present study seem to contradict what we know of complexity as the NF group had significantly larger CI values than YA and there was no difference between the FH and YA or between the FH and NF groups. However, many of the older adults in both the NF and FH groups found a narrow stance too difficult to maintain, similar to prior reports [56]. Therefore, participants adopted more of a comfortable, relative to a narrow, stance which may have contributed to increased sway in the AP versus ML direction [57,58] and helps to explain the unexpected CI results. For instance, a comfortable stance in YAs induces such a degree of ease that complexity of their postural systems does not vary greatly across tasks. Conversely, older non-fallers generally have a more difficult time maintaining posture compared to YAs and as their postural system is relatively intact, an increase in their CI reflects the increased difficulty of postural maintenance. Finally, the more comfortable stance adopted by older adults with a recent FH clearly did not allow for impairments in their system complexity to be revealed as compared to the other groups.

The common pattern of task and/or group effects found only in the AP direction is consistent with some prior findings [33,53,59] but contrary to others that showed sway in the ML direction as reliable in distinguishing fallers from non-fallers during EC and dual-task conditions [9]. Sway in the AP direction reflects ankle plantar flexion while sway in the ML direction reflects ankle eversion range of motion and hip abduction/adduction [2,59,60]. As enhanced postural difficulty recruits higher levels within the postural system and requires proper coordination between these levels, differences between the FH and NF groups in ML direction may only exist at an upper level in the hierarchy. Therefore, to elucidate ML CoP differences between FH and NF as previously reported, increased task demand may be required such as one-leg balance [59,61] or induced postural sway (moving supporting platform in AP or ML direction) [62].

While excursion amplitude was the only CoP measure that was significantly different for the FH and NF groups compared to the YA group, only the YA and NF groups showed an effect of postural task on alpha power in the occipital ROI. Within the occipital area, the previous studies have reported a significant trend of increased alpha power with increased postural task difficulty, likely for the suppression of irrelevant activity outside of postural centers [15,40,63-65]. When participants are asked to maintain posture with their EC, visual input is absent and thus processing within occipital areas (i.e., mental imagery) should be minimal so that cortical resources can be engaged by other sensory or postural areas systems. Therefore, enhanced alpha power in the occipital ROI suggests suppression of visual areas, likely working to shift cognitive resources toward subcortical centers that are critical in precise postural adjustments and stabilization. The reduced effect of postural task on occipital alpha power in the FH group suggests a potential deficit in top-down, inhibitory control required to limit bottom-up cortical processing for better balance maintenance. This may be an underlying driver of the increased likelihood of falling in the FH group, although future studies that use a large sample size and a more demanding task (i.e., balance on one leg) that induces a greater response on these higher order, regulatory centers may help parse out these oscillatory group differences.

5. Conclusions

The present findings are an excellent first step in deciphering cortical processing differences between the FH and NF/YA groups during maintained postural stance to further the development and understanding of cognitive impairments that put older adults at a greater risk for experiencing a fall. Follow- up studies that increase the sample size of our FH group may improve the statistical robustness of our findings. Future studies that record EEG and posture data simultaneously may facilitate the discovery of relationship between oscillatory activity and CoP measures.

Additional analyses could also look at oscillatory activity in different frequency bands within and across various cortical areas to understand how long and local range connectivity may be affected in FH versus NF group. However, the current results do indicate that those older adults with a recent FH likely suffer from limited cortical capacity to suppress bottom-up processing during more demanding postural conditions. This altered inhibitory function may also drive more global deficits in this group, such as worse multisensory temporal sensitivity, a conclusion supported by the correlation between wider TBWs and impaired mobility. General cortical deficits are a useful and likely target for future interventions that could not only improve postural stability but enhance global function and daily, independent living in this population.

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Conflicts of Interest

There are no conflicts of interest to disclose.

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