

Beyond Visual Feedback: Isolating Vestibular Error-Related Potentials During Wheelchair Operations

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ABSTRACT: The Error-Related Potential (ErrP) is a well-established neurophysiological marker of error perception, historically elicited through visual feedback in Brain-Machine Interfaces (BMIs). However, to our knowledge, no study has yet investigated whether such a response can be elicited in the complete absence of visual input, relying solely on vestibular feedback generated by physical movement. This study addresses this question by recording electroencephalographic (EEG) signals from participants during a powered wheelchair motion task, under two conditions: one with visual feedback available, and one with visual occlusion imposed through blindfolding. Our findings demonstrate that a recognizable ErrP waveform is elicited under full visual occlusion, exhibiting a morphology consistent with that observed in the sighted condition. Observed differences in latency and amplitude between conditions are discussed in relation to their experimental contexts. These results provide preliminary evidence that vestibular feedback alone constitutes a sufficient substrate for ErrP generation, opening a promising avenue for BMI systems not relying on visual input for implicit error monitoring.

INTRODUCTION

Error-related Potentials (ErrPs) are a class of event-related potentials (ERPs) that arise as an endogenous neurophysiological response to the perception of erroneous outcomes [1]. First characterized in the context of response monitoring, these signals are generated whenever a mismatch is detected between an expected and an observed outcome, independently of whether the error was self-generated or externally observed [1].

In the specific context of assistive devices, where the error originates from a misrecognition of the user's intent by the interface, the elicited signal is referred to as interaction ErrP [1]. At the waveform level, this subtype is characterized by a fronto-central waveform comprising a negative peak (ERN/Ne) at around 250 ms post-feedback, followed by a positive peak (Pe) at around 320 ms, and a subsequent negative peak at around 450 ms [1][2]. The consistency and robustness of this morphology across subjects and paradigms have positioned the ErrP as a particularly tractable target for single-trial decoding in brain-machine interface (BMI) research [3].

Early investigations into the functional properties of ErrPs were conducted almost exclusively within visual paradigms, in which participants monitored screen-rendered interfaces and evaluated the correctness of cursor trajectories, agent movements, or graphical decision outputs [4][5][6]. While these studies established the decodability of the signal under controlled laboratory conditions, they were structurally incapable of dissociating the role of visual processing from other potential contributors to error representation. Subsequent research extended the applicability of ErrPs to more naturalistic and physically grounded settings: reliable ErrP generation was demonstrated during the observation of robotic arm trajectories, and the signal was shown to be sufficiently stable for integration into closed-loop reinforcement learning architectures, enabling autonomous agents to update their control policies in response to implicitly decoded human evaluative feedback [7][8]. Across all of these paradigms, however, visual feedback remained the primary—and often exclusive—sensory channel through which error information was delivered to the participant. A distinct but related body of work has examined the utility of ErrPs in assistive technology (AT) applications, with a particular focus on physical human-robot interaction. Notably, it was demonstrated that ErrP responses are preserved even when the interface between user and robot involves tactile rather than purely visual coupling, as in the case of exoskeleton-based paradigms [9], providing evidence that non-visual afferent feedback can independently contribute to error-related cortical activity. Other findings enabled the construction of multi-component BMI systems in which ErrP-based error monitoring operates in concert with motor imagery-based command generation, forming a complete supervisory loop for rehabilitation robotics [10][11]. Nevertheless, a fundamental limitation persisted across this literature: in the overwhelming majority of studies, participants retained full visual access to the task environment, rendering it impossible to determine the extent to which the observed ErrP reflected visual error processing as opposed to proprioceptive, somatosensory or vestibular feedback. The application of ErrP-based BMIs to powered wheelchair navigation has received comparatively limited attention. An early investigation in this domain employed a teleoperation paradigm, in which the user re-

mained physically decoupled from the robot and evaluated its behavior through a visual display [12]—a configuration that, despite its applied motivation, did not substantively depart from the visual interface paradigms that preceded it. A more recent study proposed a complete ErrP-driven navigation pipeline; however, its validation was restricted to simulated environments and provided no evidence regarding the integrity of the signal during real wheelchair locomotion [13]. In neither case was it possible to assess whether ErrP generation would be sustained in the absence of visual feedback, or whether the physical dynamics of wheelchair movement could themselves serve as a sufficient substrate for error elicitation.

The present study was designed to address these open questions directly. Participants were asked to perform a wheelchair navigation task under two conditions: one in which visual feedback was available, and one in which full visual occlusion was imposed through blindfolding. This design allowed, for the first time, the recording of ErrPs from participants physically situated within a moving powered wheelchair, while simultaneously enabling a systematic dissociation of visual and vestibular contributions to the error-related response. Any error-related neural activity recorded under the blindfolded condition could thus be attributed unambiguously to non-visual, movement-derived afferent feedback, providing the first characterization of an ErrP elicited exclusively through vestibular channels during active self-motion in a real navigation context.

Beyond resolving this methodological ambiguity, the present investigation carries broader implications for the design of BMI systems intended for naturalistic use. If error-related cortical activity can be elicited independently of visual input — on the sole basis of movement-derived afferent feedback — then error-correction paradigms need not be restricted to interfaces providing explicit visual feedback. Such a finding would suggest that vestibular signals may constitute a sufficient, or at minimum complementary, substrate for implicit error monitoring, substantially expanding the range of contexts in which ErrP-based BMIs could be deployed.

MATERIALS AND METHODS

Participants: Five healthy male volunteers (mean age 27.8 years, range 26-31) took part in the experiment, one of whom had prior experience with ErrP-based paradigms. Informed consent was obtained from all participants prior to the study, which was conducted in compliance with the ethics committee "Comitato etico della ricerca psicologica", University of Padova (1575-b).

Wheelchair setup: A commercial powered wheelchair (18.68E HI-LO Electric, Vassilli s.r.l., Padua, Italy) was modified to meet the requirements of this study. Wheelchair kinematics were tracked by two Kübler KIS40 encoders attached to the front wheels via a belt, and odometry was derived using the method in [14]. User interaction was mediated by a Logitech F710 joystick, while a screen mounted in front of the participant pro-

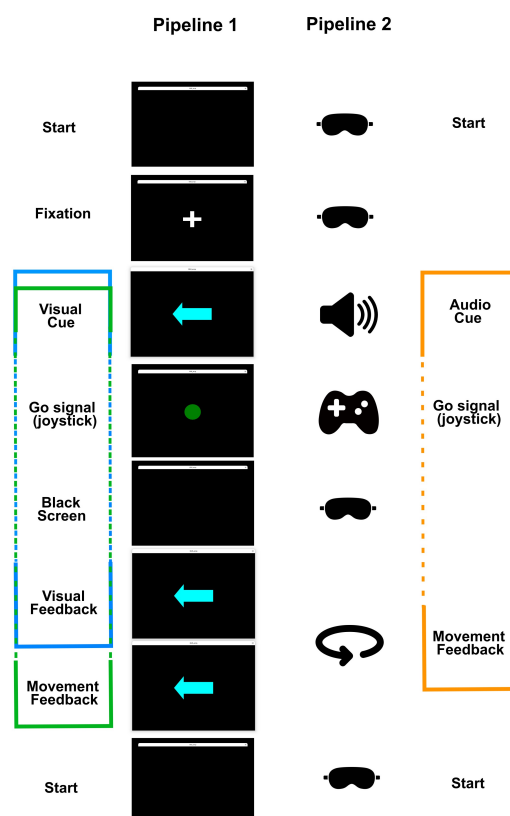


Figure 1: Schematic representation of the two experimental pipelines employed for ErrP elicitation. Pipeline 1 (left) follows a fully visual paradigm, in which each depicted symbol on the display marks a specific event. Pipeline 2 (right) preserves the same temporal structure but replaces visual stimuli with audio cues, with participants remaining blindfolded throughout. Colored boxes highlight the three experimental contrasts of interest: the blue dashed box isolates the visual ErrP epoch, when the mismatch is between the visual cue and the visual feedback; the green solid box captures vestibular ErrPs in the presence of vision, when the mismatch is between the visual cue originally given and the rotational feedback; and the orange dashed box identifies the purely vestibular ErrP under blindfolded conditions, when the mismatch is between the audio cue and the rotational feedback.

vided the visual feedback integrated into the experimental pipeline. The central computing unit was a Lenovo ThinkPad T15g Gen 2 laptop (11th Gen Intel Core i9-11950H CPU, NVIDIA GeForce RTX 3080 GPU, 32 GB RAM), to which all components were connected. An R-Net chipset (Curtiss-Wright) bridged the wheelchair's CAN-bus to the laptop via a USB isolator, minimizing electrical interference and ground-loop artifacts induced by the motor system. The software architecture relied on ROS Noetic (v1.17.3) as the communication backbone, with ROS-Neuro middleware [15] ensuring synchronization between the BMI, GUI, and robotic components. Post-hoc data analysis was conducted on the same platform using MATLAB 2024a (MathWorks, Natick, MA, USA).

Signal acquisition: The EEG acquisition system consisted of a 64-channel amplifier (AntNeuro eego 64, The Netherlands), with the following 32 channels retained for

analysis: AF3, AF4, F3, F1, FZ, F2, F4, FC3, FC1, FCZ, FC2, FC4, C3, C1, CZ, C2, C4, CP3, CP1, CPZ, CP2, CP4, P3, P1, PZ, P2, P4, PO3, POZ, PO4, O1, O2. The reference electrode was placed at Iz and the ground electrode at AFz. Ocular artifacts were monitored via electrooculography (EOG), recorded through an external bipolar box connected by adapter cable to adhesive pads positioned around the right eye. Both EEG and EOG signals were sampled at 512 Hz and filtered using a 4th-order zero-phase Butterworth bandpass filter with a passband of 2–10 Hz. Spatial resolution of the EEG was further enhanced using a small Laplacian spatial filter computed from the four nearest neighboring electrodes for each channel.

Experimental setup: To isolate the vestibular error potential and establish a robust basis for comparison, data were collected across two distinct experimental pipelines (Fig. 1). In both conditions, each participant was seated in the powered wheelchair and instructed to follow a pre-defined sequence of cues.

In the first pipeline, each trial consisted of two successive phases: a visual-feedback phase and a movement-feedback phase. Trials began with a black screen (1.5 s), followed by a fixation cross (1.5 s) and a directional cue (1.5 s). Upon the appearance of a green dot, the participant provided input via the joystick. Following an inter-stimulus interval (2.0 s), a visual feedback arrow (2.0 s) indicated the direction of the upcoming movement. The wheelchair remained stationary throughout these steps. After a delay of 1.5 s from the onset of the visual cue, the wheelchair performed an in-place rotation at a constant angular velocity for 2.0 s, constituting the movement feedback consistent with the previously displayed arrow. Each trial concluded with a final black screen (1.5 s). ErrPs were elicited by introducing unexpected events with a probability of 30%: in these trials, both the visual feedback arrow and the subsequent physical rotation were presented in the direction opposite to the participant's joystick input. In the remaining 70% of trials, the system's response was congruent with the participant's intended command.

In the second pipeline, designed to isolate the purely vestibular error potential, participants were blindfolded to eliminate any visual information. The temporal structure of the first pipeline was preserved to allow direct comparison; however, directional instructions were delivered via audio cues reproducing the words 'Left' or 'Right'. In the absence of visual feedback, participants were able to respond with the corresponding joystick command immediately after the audio cue. After a delay of 1.5 s from joystick input, the wheelchair executed an in-place rotation at a constant angular velocity for 2.0 s, at a lower speed than that employed in the first pipeline, as a precautionary measure to minimize vestibular discomfort in participants during the blindfolded condition. The direction of the rotational feedback was incongruent with the audio cue with a probability between 30% and 40%.

To ensure consistency across participants, a full writ-

ten description of the task was provided before each session for both pipelines. Each participant completed ten runs per pipeline, each consisting of ten consecutive trials. Prior to the recording phase, a brief familiarization period with the wheelchair was provided to ensure comfort and procedural understanding. All sessions were conducted with motors continuously active, a design choice motivated by the need to evaluate EEG signal quality under realistic electromagnetic conditions and to avoid artifacts associated with motor switching transients. To further ensure reproducibility, participants were instructed to limit head movements and maintain a consistent posture throughout each trial. Trial rejection was performed by visual inspection: only trials exhibiting the expected EEG frequency spectrum were retained, while those showing evident noise in the EOG channel or EEG amplitudes exceeding 35 μ V were discarded (12.9% of total trials).

RESULTS

The grand-average ErrPs recorded at FCz are shown in Figure 2 for three experimental conditions: a) Visual ErrP, b) Vestibular ErrP in the presence of vision and c) Vestibular ErrP in the absence of vision. Each subplot displays the difference waveform obtained by subtracting the mean correct-trial signal from the mean error-trial signal, averaged across participants; the shaded area represents ± 1 standard error of the mean (SEM). In all three conditions, the ErrP morphology appears clearly identifiable: a prominent negative deflection (Ne/ERN-like component) followed by a positive peak (Pe-like component), consistent with the canonical error-related potential described in the literature. The topographic maps computed within a ± 30 ms window centered on each relevant peak suggest a fronto-central scalp distribution focused on FCz and Cz, in agreement with the expected neural generators of the ErrP [1].

Condition 1 – Visual ErrP: The first row shows the potential evoked by a visual mismatch, in which the visual cue and the visual feedback were incongruent. The negative peak is sharp and well-defined and occurs at a relatively short latency (mean latency = 0.302s) with respect to feedback onset ($t = 0$). The topographic maps at the peaks reveal a highly focal activation restricted to the FCz–Cz region.

Condition 2 – Vestibular ErrP in the presence of vision: The second row depicts the response to a mismatch between the visual cue and the wheelchair rotation feedback, delivered while participants retained full visual access. The characteristic negative-positive complex is preserved. A slight increase in peak latency (mean negative peak latency = 0.446s) relative to Condition 1 is observed, which may reflect the additional processing time required to perceive the direction of rotation. The topographic distribution at the peaks shows both some focal activity over FCz–Cz and also a broader activation. The dotted and dashed traces overlaid on the subplot represent, respectively, the delta in angular velocity with

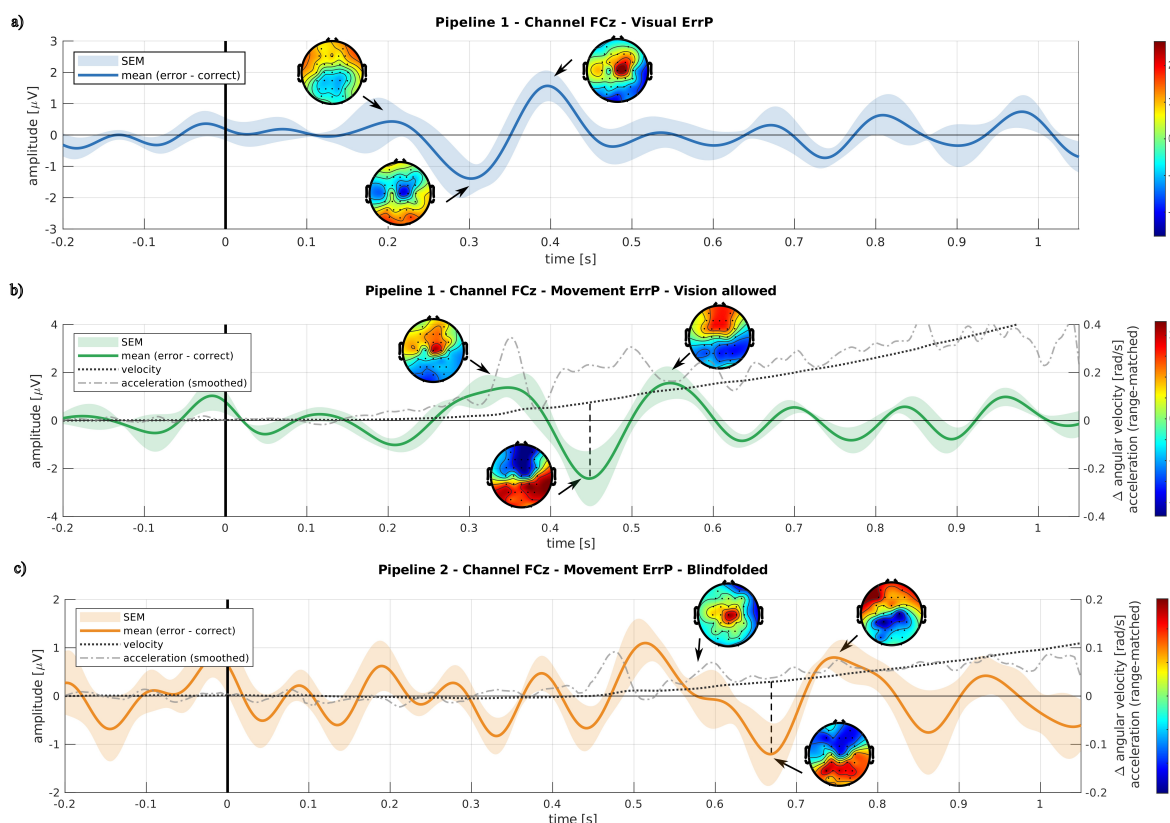


Figure 2: Grand-average ErrP difference waveforms (error minus correct) recorded at FCz across the three experimental conditions, averaged across participants (shaded area: ± 1 SEM). **a)** *Condition 1*: visually-evoked ErrP, time-locked to the onset of the visual feedback stimulus. **b)** *Condition 2*: vestibular ErrP recorded in the presence of vision, time-locked to wheelchair rotation onset; **c)** *Condition 3*: vestibular ErrP recorded under full visual occlusion, time-locked to wheelchair rotation onset. Note that amplitude scales differ across panels to enhance morphological comparability. For each condition, topographic maps computed within a ± 30 ms window centered on the most prominent peaks illustrate the fronto-central scalp distribution of the response. For the second and third condition, the dotted trace represents the delta in angular velocity of the wheelchair with respect to its rest state, while the dashed trace represents the corresponding angular acceleration, scaled to match the velocity axis range to facilitate visual comparison of peak timing. The vertical line at instant 0 represents the feedback onset, visual or rotational. Vertical dashed lines show the velocity value corresponding to the negative peak of ErrP.

respect to its rest state and the angular acceleration of the wheelchair, scaled to match the velocity axis range to facilitate visual comparison of peak timing. A vertical dashed line connects the ErrP negative peak to the corresponding velocity value. Inspection of these kinematic signals suggests that the positive peak preceding the negative ErrP peak occurs in temporal proximity to, or slightly after, the peak in angular acceleration, raising the possibility that the transient acceleration event may constitute the primary vestibular cue through which the direction of the erroneous movement is first detected.

Condition 3 – Vestibular ErrP in the absence of vision: The third row presents the ErrP elicited under purely vestibular conditions: participants were blindfolded and the mismatch occurred between an auditory cue and the wheelchair movement. The negative deflection is present but exhibits reduced amplitude and a longer latency compared to the two preceding conditions (mean negative peak latency = 0.667s). The SEM band is slightly wider, suggesting greater inter-subject variability. Neverthe-

less, the first topographic map still displays a fronto-central focus qualitatively comparable to that of Condition 1, which may support the presence of a genuine error-related signal. The second and third topoplots show a more diffuse distribution, broadly similar to that observed in Condition 2, possibly suggesting a shared neural substrate across the two motor conditions regardless of visual availability. A qualitatively similar temporal relationship between the positive peak preceding ErrP negativity and the angular acceleration peak is observed under blindfolded conditions, suggesting that this kinematic feature may retain its role as a salient afferent trigger for error detection even in the absence of visual input.

DISCUSSION

A central motivation for investigating vestibular ErrPs in assistive navigation contexts stems from the practical limitations of display-based feedback paradigms in such scenarios. In real-world wheelchair use, sustained visual attention to a monitor is neither practical nor desirable,

as dynamic environments demand that the user's gaze be directed toward the surroundings. The possibility of eliciting an ErrP on the basis of the physical movement itself, with feedback delivered through vestibular channels rather than through a visual display, would therefore eliminate this gaze-allocation constraint and constitute a meaningful step toward more ecologically valid and natural BMI-driven navigation.

The present results show that an error-related potential can be elicited across three distinct sensory conditions — visual, visuo-vestibular, and purely vestibular — in the context of wheelchair-based BCI control. The consistent ERN–Pe morphology and the fronto-central topography observed across all conditions are in line with the established ErrP literature, and are compatible with the interpretation that the recorded signals reflect error monitoring activity involving the anterior cingulate cortex and supplementary motor area [1].

However, some considerations need to be made regarding the amplitude differences between the three conditions, since Conditions 1 and 2 yielded larger ErrP amplitudes than Condition 3. This pattern may be attributable to at least two concurrent factors. First, the error stimulus probability in Conditions 1 and 2 was fixed at 30%, whereas in Condition 3 it ranged between 30% and 40% depending on the run. Given the well-established inverse relationship between error probability and ErrP amplitude, the slightly elevated error rate in Condition 3 may have contributed to the observed amplitude reduction. Second, it has been reported that ErrP amplitude may co-vary with the perceived salience or urgency associated with an erroneous movement, which may be approximated by the velocity of the unintended motion [16]. The lower wheelchair rotation speed employed in Condition 3 relative to Condition 2 would predict a smaller error signal under this account. These two explanations are not mutually exclusive and may have acted in concert to attenuate the vestibular ErrP; future work could focus on disentangling their respective contributions.

Moreover, blindfolding in Condition 3 may have introduced alpha-band contamination, as eye closure is known to increase posterior alpha power [17], potentially contributing to amplitude differences between Conditions 2 and 3 beyond those attributable to the sensory manipulation itself. However, the available evidence argues against alpha contamination as the primary explanation for the observed ErrP. In fact, the 2–10 Hz bandpass filter partially attenuates the alpha band, and more importantly, the fronto-central topography and time-locked ERN–Pe morphology of the signal are inconsistent with the posterior, sustained fluctuations characteristic of alpha activity [17]. Taken together, these observations support the interpretation that the signal in Condition 3 reflects genuine error-related activity, even if alpha modulation may have contributed to the increased inter-subject variability observed in that condition.

The latency differences across conditions also deserve consideration. The visual ErrP in Condition 1, being

time-locked to the discrete onset of a visual stimulus, would be expected to exhibit the shortest and most consistent latency. The vestibular ErrPs in Conditions 2 and 3, by contrast, are contingent on the participant's ability to perceive the direction of wheelchair rotation — a process that does not begin instantaneously at command delivery but rather unfolds over a temporal window shaped by the dynamics of the rotation itself. The longer latency observed in the motor conditions is therefore consistent with the additional sensorimotor processing required to infer movement direction from vestibular or combined visuo-vestibular signals. The further latency prolongation in Condition 3 may suggest that the absence of visual information and the reduced rotational speed jointly delay the detection of the erroneous movement direction, thereby shifting the ErrP peak and increasing inter-subject variability — the latter also reflected in the wider SEM band.

A further kinematic observation is consistent with this latency interpretation. Visual inspection of the grand-average waveforms suggests that, in both motor conditions, the positive peak preceding the negative ErrP peak coincides with, or follows slightly after, the peak in wheelchair angular acceleration. This pattern is compatible with the hypothesis that the transient acceleration event, rather than sustained rotational velocity per se, may represent the critical afferent trigger for error detection in this paradigm. This interpretation aligns with known properties of vestibular and somatosensory encoding, in which acceleration-sensitive afferents respond preferentially to changes in motion state rather than to sustained movement [18]. Whether acceleration magnitude modulates ErrP amplitude in a manner analogous to the velocity effects discussed above remains an open question that warrants systematic investigation in future work.

The topographic similarity between the second and third topoplots of Conditions 2 and 3 is particularly noteworthy. Despite the fundamentally different sensory context, the spatial pattern of cortical activation appears strikingly comparable across the two conditions, which may imply that a common neural mechanism is engaged when monitoring errors in wheelchair rotation, irrespective of whether this monitoring is supported by concurrent visual input. This observation is compatible with the hypothesis that the signal recorded in Condition 3 belongs to the same functional class as the ErrP in Condition 2, rather than reflecting an unrelated cognitive process; nonetheless, further evidence from single-trial analyses and larger samples would be needed to confirm this interpretation.

Taken together, these preliminary findings suggest the feasibility of eliciting a vestibular ErrP in a real-world assistive device scenario, and point to movement velocity and acceleration and sensory modality as potential factors shaping its amplitude and latency. Notably, the present results provide evidence that a recognizable ErrP waveform can be elicited under purely vestibular condi-

tions, indicating that error-related signal generation is not contingent upon the availability of visual feedback, but instead reflects a more general error-monitoring mechanism capable of operating across different sensory modalities. As a consequence, visual and vestibular error signals may represent independent contributions to a common monitoring process, which could in principle be engaged either in isolation or in combination depending on the sensory context. Future work should systematically manipulate rotational speed in controlled designs to establish a more quantitative characterization of the velocity-amplitude and acceleration-amplitude relationships, and investigate whether the signal parameters observed here are compatible with reliable single-trial classification for closed-loop BCI error correction.

CONCLUSION

The present study, by identifying an ErrP under purely vestibular conditions, provides evidence that error-related neural activity is not exclusive to visually-driven paradigms, but may instead reflect an error-monitoring mechanism that operates across sensory modalities. From an applied perspective, these findings suggest that BMI error-correction paradigms need not be restricted to interfaces providing explicit visual feedback, opening the way toward more naturalistic, attention-preserving systems in which movement-contingent vestibular signals serve as a sufficient, or complementary, substrate for implicit error monitoring in real-world wheelchair navigation.

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