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Motion P3 Demonstrates Neural Nature of Motion ERPs

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***Abstract*—The technical challenges of recording electroencephalographic (EEG) data during motion are considerable, but would enable the possibility of investigating neural function associated with balance, motor function and motion perception. The challenges include finding a reliable method of motion stimulus reproduction, removing artifacts, and ensuring that the recordings retain sufficient EEG signal for proper interpretation. This study details the use of the P3 waveform to validate the concept of motion-based EEG data, and discusses some potential future uses in experimental and clinical settings.**

I. Introduction

The proper perception of self-motion is important in everyday function – walking, standing and sitting rely on knowing whether and how much the body is in motion. The vestibular system is very important in the perception of self-motion, as it detects the body's acceleration. Heading, or direction, is a fundamental feature of motion and recent studies have shown that the vestibular system plays a pivotal role in the perception of self-motion heading in humans (e.g. [1, 2]). There have been a number of advances in the understanding of the neuronal processing of visual and vestibular signals for self-motion perception in non-human primates [3-6]. However to date, there has only been very few studies investigating the neural correlates of self motion perception in humans. Electroencephalography (EEG) is the most suitable candidate for these purposes as invasive recordings are often impossible, and other non-invasive methods such as function magnetic resonance imaging (fMRI) or positron emission tomography (PET) require bulky equipment which does not lend itself to motion studies.

These studies have primarily investigated rotational self-motion [7]-[10]. These have been seen to have some potential for clinical research [11] but problems such as electrooculogram (EOG) and electromyogram (EMG) artifacts from rotational stimuli are frequently reported [8, 9, 11]. Advances in signal processing methodologies such as independent component analysis (ICA) now allow the attenuation of such artifacts. Previously we have shown that it is feasible to acquire EEG responses to auditory stimuli during continuous linear self-motion on a Stewart platform without noticeable electromagnetic or EMG interference [12]. Building upon these results, here we investigate the possibility of recording EEG responses to a linear motion stimulus.

We have chosen heading as the motion feature to manipulate, and the P3 event-related potential (ERP) component as the EEG feature to evoke. To do this we have adapted the classical two-stimulus oddball paradigm [13], in which participants are presented a stream of frequent standard stimuli (80%) and infrequent target stimuli (20%) and are instructed to respond to the target stimuli. The target stimulus elicits the P3 component, which is a response to a new stimulus. The P3 component is a well-studied component evoked by observing change in the experimental environment; it has been observed in multiple stimulus modalities: visual, auditory, somatosensory [14] and olfactory [15].

In this study, participants were seated upon a Stewart motion platform while EEG was acquired using a 128-channel system. The standard and target heading stimuli were forward translations at a 45° angle to the left or right of straight ahead. The goal of our study was to investigate the neural correlates of self-motion perception using high density EEG to characterize vestibular processing of heading change.

II. Methods

A. Subjects

Sixteen subjects with normal or corrected-to-normal vision participated in the experiment. The age range was 22 – 35 (mean 28.1 ± standard deviation 3.9). Subjects gave their informed consent before taking part in the experiment, which was performed in accordance with the ethical standards laid down in the 1964 Declaration of Helsinki. One subject's data

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was particularly noise-contaminated and was removed from the study.

B. Apparatus

A Maxcue 600 platform manufactured by Motion-Base PLC [16] was employed. This is a 6-legged Stewart platform with 6 degrees of freedom, which are rotation and translation about 3 axes. A fixation cross was displayed on a projection screen, with a field of view of 86°×65°, a resolution of 1400×1050 and a refresh rate of 60 frames per second. Subjects wore Sennheiser HD600 headphones with one-way communication capability while white noise was played to mask the sound of the platform. EEG data were recorded at 512Hz using a Biosemi ActiveTwo™ 128 channel EEG system with 7 supplementary electrodes recording EOG and reference channels. Data were referenced to the average reference.

C. Motion Paradigm

The classical oddball paradigm, in which an infrequent target is identified in a sequence of frequent standard stimuli [17], was adapted for a bi-directional motion stimulus.

Subjects were presented with sequential angled forwards motions; the angles were either 45 degrees left or 45 degrees right. Left and right were presented in an oddball manner, with random presentation probabilities of 0.8 for standard and 0.2 for target. The use of left and right for the standard and target was balanced across runs. The return to center after each stimulus was of sub-threshold amplitude. There was a random interval of 1.5 – 2.5 s after the return. There were a total of 8 blocks, each consisting of 50 trials – 40 standards and 10 targets – yielding a total of 320 standard stimuli and 80 target stimuli per subject. Each block lasted approximately 5 minutes. All subjects were given two breaks lasting 5-10 minutes.

The ideal motion displacement profile was:

$$s(t) = 0.49\frac{(2\pi t - \sin(2\pi t))}{4\pi^2} \qquad 0 < t \le 1s$$

where t is time. It had a maximum displacement, velocity and acceleration of 0.078 m, 0.156 ms^{-1} and 0.49 ms^{-2} respectively, lasting 1s, which are above the detection thresholds reported in [18].

D. EEG Processing

All data were high-pass filtered at 1 Hz and low-pass filtered at 95 Hz for processing, with a bandstop filter at 47 - 53 Hz. The FASTER processing method [19] was used to preprocess data and remove artifacts. Epochs of 1000ms with 500ms pre-stimulus baseline were extracted from the continuous data; there was a mechanical delay of approximately 200ms from the trigger onset to the onset of the motion.

E. Analysis

The goal of this study was to determine whether motion-based EEG responses were valid. The P3 is known to be centered on the midline between electrode sites Cz and Pz, depending on the modality. It is known to occur approximately 300 – 400 ms after stimulus onset in healthy adults. Thus the spatial regions of interest were taken as electrode sites Pz, Cz and Fz along the midline. Temporal regions of interest were pre-identified as 300ms – 800ms after stimulus onset, to allow for the non-discrete nature of the motion stimulus. Due to the wide peak observed from the ERPs, the target ERP was also epoched relative to the button press to remove some of the inter-subject temporal variability [20]. This is seen in figures 2 and 3, where there are 2 target ERPs – epoching relative to motion onset means that the 0ms mark denotes the start of motion, while in the button-press epoched target, 0ms denotes the moment the button was pressed.

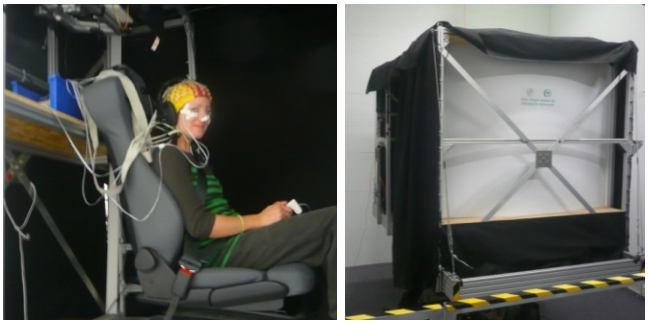

Figure 1. (Left) The outside of the Stewart platform. (Right) A subject with EEG electrodes in place, ready to perform the task. During this study, a cloth sheet covered the platform to mask visual motion cues, and a Biosemi ActiveTwo EEG system was installed to record EEG data.

III. Results

A. Behavioral Results

Reaction times to the target stimulus were measured, along with hit rates.

TABLE I
Reaction times and hit rate for the target stimulus, and statistical tests of left (L) vs. right (R) directions.

Measure	Value (mean ± SD)	Statistical test, L vs. R
Reaction time (ms)	876 ± 214	t_{14}=1.12, p=0.14
Hit rate (%)	93.91 ± 9.36	t_{14}=0.31, p=0.38
Reaction time intra-participant SD (ms)	57 ± 24	-

B. EEG Results

Amplitude values were compared between standard and target conditions at midline electrodes.

TABLE II
Mean values at the 600ms peak (100ms average) at midline electrodes for standard (STD) and target (TGT) conditions, and statistical test values.

Electrode	Amplitude, standard (95% CI, μV)	Amplitude, target (95% CI, μV)	Statistical test STD vs. TGT
Fz	-0.798±0.45	-2.162±0.92	t_{14}=3.24, p=5.9x10^{-3}
Cz	0.306±0.40	1.950±0.96	t_{14}=4.40, p= 5.99x10^{-4}
Pz	0.646±0.41	2.910±0.84	t_{14}=7.56, p=2.62x10^{-6}

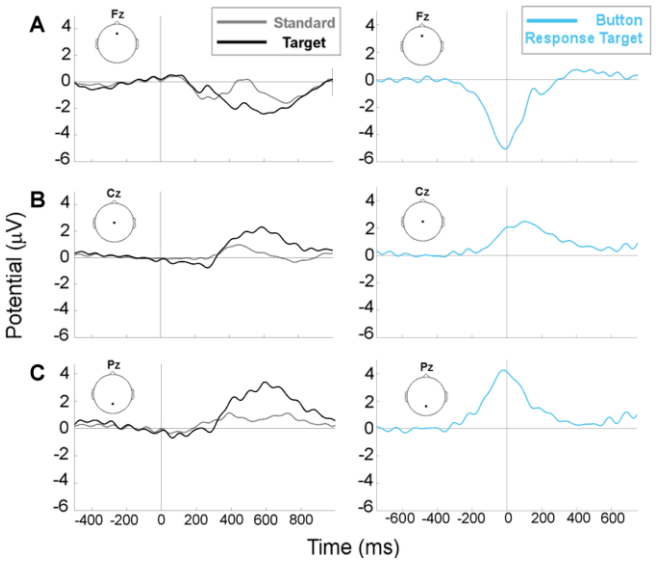

Figure 2. (Left) Group average standard (gray) and target (black) ERPs from (A) Fz, (B) Cz and (C) Pz after averaging relative to **motion onset** (0ms denotes the motion onset). (Right) Group average target ERPs from (A) Fz, (B) Cz and (C) Pz after averaging relative to the **button response** (0ms denotes the button response)**.**

Group average ERPs computed from the standard (grey) and target (black) conditions with respect to motion onset (MO) and button press (BP) are shown in Figure 2. The target averaged with respect to the button press is also shown (blue). As participants did not respond during the standard condition, there is no standard condition for the button press ERPs. Inspection of the group average ERP-waveforms, in Figure 3, showed a clear dipolar response peaking at 600ms, with a topographic maximum at the parietal midline site (Pz). Pz is the typical site at which the P3 is measured. Table II details statistical tests performed on the peaks of the MO standard vs. target ERPs at these electrodes.

Figure 3 shows butterfly plots of the group average MO standard (A), MO target (B) and BP target (C) waveforms. These figures display each channel's ERP overlaid, and scalp topographies at three time periods, including the largest positive peak. The topographies in A and B represent two time periods of similarity between standard and target waveforms at 300 ms and 350 ms, and the P3 peak at 600ms which is seen to differ between conditions, while the topographies in C represent 200ms before button press, 0ms, i.e. the time of button press, and 200ms after.

The results indicate that the elicited response contains the P3 waveform.

IV. Discussion

In this study we investigated the feasibility of recording motion-based EEG responses using a basic oddball paradigm. The results display a typical P3 topography (see [21]) with significant peak amplitude increase at 600ms. This provides strong evidence that we have successfully recorded a P3 response to a motion stimulus without significant EMG or EOG interference. As we are investigating the sensory

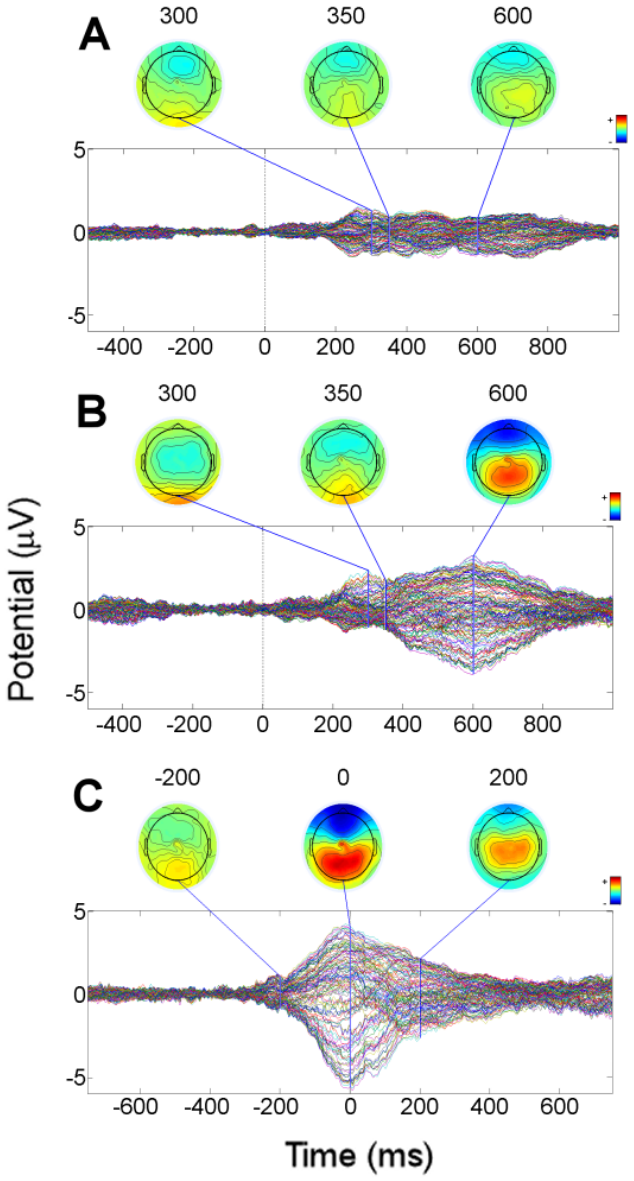

Figure 3. Butterfly plots depicting (A) MO Standard, (B) MO Target and (C) BP Target.

processing of motion using a similar paradigm (for preliminary results see Nolan et. al, 2011), the aim of this P3 experiment was a validation of the protocol to ensure that the EEG data obtained during motion was comparable to data obtained during static EEG sessions. The P3 is a well-known ERP component which has been seen to be present across multiple sensory modalities and so was chosen as the ERP feature to investigate to validate the paradigm.

While this positive result may initially seem straight forward, it should be emphasized that there were a number of potential factors which may have interfered with proper EEG recording. The foremost of these is EMG interference – the possibility that the ERPs recorded were in fact capturing muscle movement of the neck to stabilize head in response to the full-body motion via the vestibulo-collic reflex. However, the P3 component occurs only in the target stimuli which are directionally different but of the same amplitude,

and so the increase in amplitude cannot be considered to be related to neck muscle movement. Furthermore, frequency analysis showed no increase in high-frequency activity which would be characteristic of EMG responses. Topographically, such stabilizing EMG responses would be likely to occur temporally, while here the P3 is seen to be active in the parieto-central regions. For these reasons, we conclude that the ERPs are not driven by EMG activity.

Due to the vestibular-ocular reflexes, the possibility of contamination of the EEG by EOG signals was also a possibility. Such signals are characterized by high-amplitude frontal topographies, which are not seen the ERP responses recorded here. Furthermore, it has been seen that FASTER, the artifact reduction method employed in this study, is highly effective in removing ocular contamination [19].

A further result of the study arises when considering the differences between the MO target ERP and the BP target ERP. Due to the sinusoidal nature of the stimulus and the inherent ramp-up of intensity that comes with it, detection of motion onset is not instant. The wide peak seen in the P3 in the motion-onset epoched ERP is considerably sharper in the button-press epoched ERP. This implies that the inter-subject variability of motion detection threshold is high, as the sharper P3 peak from each subject is spread across time when averaged, resulting in the wide peak. This concept is supported by the behavioral data presented in Table I, where the inter-subject variability in response time is 214ms, while the intra-subject variability is only 57ms. This observation should be taken into account in further motion-based EEG studies.

The consequences of these results are far-ranging. As the vestibular system contributes strongly to motion perception [1], the possibility of testing vestibular function experimentally, to enhance understanding of the neural correlates of vestibular function, and clinically, for aiding diagnosis in vestibular dysfunctions. Furthermore, the investigation of the multisensory contribution of visual, somatosensory and auditory senses to motion perception – which is currently poorly understood – is also a possibility. This may be particularly important in research in aging – falls are the most common injury for elderly people [22], and as multisensory integration has been seen to become inefficient with age and may be implicated in falls [23], investigating the multisensory elements of motion perception may lead to better prevention of falls.

In conclusion, this study has demonstrated a P3 component elicited using a motion-based stimulus, without interference from reflexive EMG or EOG. This provides a strong grounding for further EEG-based studies into the processing of full-body motion.

REFERENCES

[1] J. S. Butler, S. T. Smith, J. L. Campos, H. H. Bülthoff, "Bayesian integration of visual and vestibular signals for heading," *J Vis,* vol. 10, p. 23, 2010.

[2] C. R. Fetsch, A. H. Turnned, G. C. DeAngelis, D. E. Angelaki, "Dynamic Reweighting of Visual and Vestibular Cues during Self-Motion Perception," *J. Neurosci.,* vol. 29, pp. 15601-15612, December 9, 2009 2009.

[3] Y. Gu, G. C. DeAngelis, D. E. Angelaki, "A functional link between area MSTd and heading perception based on vestibular signals," *Nat Neurosci,* vol. 10, pp. 1038-1047, 2007.

[4] Y. Gu, D. E. Angelaki, G. C. DeAngelis, "Neural correlates of multisensory cue integration in macaque MSTd," 2008.

[5] Y. Gu, C. R. Fetsch, B. Adeyemo, G. C. DeAngelis, D. E. Angelaki, "Decoding of MSTd Population Activity Accounts for Variations in the Precision of Heading Perception," *Neuron,* vol. 66, pp. 596-609, 2010.

[6] C. R. Fetsch, A. H. Turner, G. C. DeAngelis, D. E. Angelaki, "Reliability-based cue re-weighting in rhesus monkeys: behavior and neural correlates.," *IMRF,* 2009.

[7] J. D. Hood and A. Kayan, "Observations upon the evoked responses to natural vestibular stimulation," *Electroencephalography and Clinical Neurophysiology/Evoked Potentials Section,* vol. 62, pp. 266-276, 1985.

[8] C. Kolchev, "Vestibular Late Evoked Potentials (VbEP) Processed by Means of Brain Electrical Activity Mapping (BEAM)," *Acta Oto-laryngologica,* vol. 115, pp. 130-133, 1995.

[9] V. Rodionov, J. Elidan, H. Sohmer, "Analysis of the middle latency evoked potentials to angular acceleration impulses in man," *Electroencephalography and Clinical Neurophysiology/Evoked Potentials Section,* vol. 100, pp. 354-361, 1996.

[10] R. Loose*, et al.*, "Vestibular evoked potentials from the vertical semicircular canals in humans evoked by roll-axis rotation in microgravity and under 1-G," *Behavioural Brain Research,* vol. 134, pp. 131-137, 2002.

[11] B. Hofferberth, "The Clinical Significance of Vestibular Evoked Potentials (REP)," *Acta Oto-laryngologica,* vol. 115, pp. 124-125, 1995.

[12] H. Nolan, R. Whelan, R. B. Reilly, H. H. Bülthoff, J. Butler, "Acquisition of Human EEG Data during linear Self-Motion on a Stewart Platform," in *Proceedings of the 10th Annual Meeting of International Multisensory Research Forum*, New York, USA, 2009, p. 363.

[13] J. Polich, "Cognitive Brain Potentials," *Current Directions in Psychological Science,* vol. 2, pp. 175-179, 1993.

[14] K. H. Chiappa, *Evoked Potentials in Clinical Medicine*. New York: Lippincroft-Raven, 1997.

[15] C. D. Morgan, M. W. Geisler, J. W Covington, J. Polich, C. Murphy, "Olfactory P3 in young and older adults," *Psychophysiology,* vol. 36, pp. 281-287, 1999.

[16] M. von der Heyde, "A Distributed Virtual Reality System for Spatial Updating: Concepts, Implementation, and Experiments," Technische Fakultät, Universität Bielefeld, 2001.

[17] J. i. Katayama and J. Polich, "Auditory and visual P300 topography from a 3 stimulus paradigm," *Clinical Neurophysiology,* vol. 110, pp. 463-468, 1999.

[18] A. J. Benson, M. B. Spencer, J. R. Stott, "Thresholds for the detection of the direction of whole-body, linear movement in the horizontal plane," *Aviat Space Environ Med,* vol. 57, pp. 1088-96, Nov 1986.

[19] H. Nolan, R. Whelan, R. B. Reilly, "FASTER: Fully Automated Statistical Thresholding for EEG artifact Rejection," *Journal of Neuroscience Methods,* vol. 192, pp. 152-162, 2010.

[20] H. Gibbons and J. Stahl, "Response-time corrected averaging of event-related potentials," *Clinical Neurophysiology,* vol. 118, pp. 197-208, 2007.

[21] J. Polich and M. R. Heine, "P300 topography and modality effects from a single-stimulus paradigm," *Psychophysiology,* vol. 33, pp. 747-52, Nov 1996.

[22] CDC, "Fatalities and Injuries from Falls Among Older Adults --- United States, 1993--2003 and 2001--2005," *Morbidity and Mortality Weekly Report,* vol. 55, pp. 1221-1224, 2006.

[23] A. Setti, K. Burke, R. Kenny, F. Newell, "Is inefficient multisensory processing associated with falls in older people?," *Experimental Brain Research,* vol. 209, pp. 375-384, 2011.