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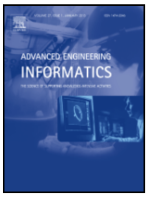

# Beyond BCI—Validating a wireless, consumer-grade EEG headset against a medical-grade system for evaluating EEG effects of a test anxiety intervention in school

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## ABSTRACT

Educational neuroscience is an emerging interdisciplinary field. However, the use of neuroimaging techniques and tools such as electroencephalography (EEG) in school-based interventions and research is limited, largely due to the high costs and physical constraints of conventional research- or medical-grade equipment. Neuroimaging and electrophysiological data can provide useful evidence to validate the efficacy of interventions. The present study explores the utility of lightweight, affordable, and easy-to-set-up EEG systems for use in school-based research with children. Specifically, we examine the effects of a deep-breathing-for-test-anxiety intervention on brain electrical activity during a flanker distractor interference task in eleven-year olds, comparing the pattern of results observed using a consumer-grade EEG system (Emotiv EPOC+) against that obtained using a medical-grade EEG system (Neurostyle). Behavioral, EEG, and respiratory data was obtained from Primary 5 students ($N = 45$; $M_{age} = 10.88$, $SD = 0.33$), split into Emotiv and Neurostyle groups. The aim of the study was two-fold: to examine the effects of deep breathing on neurophysiological and behavioral correlates of inhibitory control of attention in children, and to understand the affordances and limitations of the Emotiv EPOC+ system for school-based research with children. Results from power spectral analyses suggest that deep breathing may enhance attentional control on a neural level by modulating brain electrical activity on several frequencies. Despite limitations, the consumer-grade EEG system appears to be capable of detecting some degree of power spectral differences associated with intervention effects.

## 1. Introduction

The recent years have seen increasing interest in applying findings and techniques from neuroscience to the field of education, including validating cognitive theories or educationally relevant interventions with neuroscientific evidence. Nonetheless, the gap between neuroscience and education is often called “a bridge too far” [1]. For instance, a source of neurophysiological data extensively used in the study of cognitive and affective processes is EEG—non-invasive recordings of brain activity measured by electrodes placed along the scalp. Neurofunctional data measured by techniques such as EEG provide insights on covert cognitive processes that may not always be apparent on a behavioral level. However, the use of EEG outside of laboratory settings—such as, in validating or monitoring school-based interventions—has been limited by the costs and physical constraints imposed by conventional clinical systems. Technological advancements in recent years have, however, resulted in new possibilities and opportunities to “shorten the bridge”, even if only incrementally. Lightweight, mobile, and simplified versions of clinical EEG systems in the form of wireless EEG headsets are now commercially available at relatively low cost and are easy to set up. Though originally developed for brain-computer interface (BCI) gaming applications, some of these consumer-grade headsets also provide EEG-based and raw EEG data, making them a potentially valuable tool for wide scale studies of covert cognitive processes in ecologically valid settings, such as during a classroom lesson or collaborative activity.

Simplified, low-resolution systems, however, come with limitations to the quality and extent of the data that can be collected. To date, much of the research conducted with these consumer-grade EEG headsets focused on BCI applications. A handful of studies have attempted to validate some of the more popular commercial systems for use in recording research-quality EEG data [2–4]. For example, Badcock and colleagues compared auditory ERPs recorded using the Emotiv to those simultaneously collected by a conventional research system and found Emotiv to be a valid alternative for recording reliable late auditory ERPs—though results may be constrained by noise [3]. Most validation studies have focused on examining signal quality recorded with a few electrodes over specific regions; none has yet examined how consumer-grade systems may fare against clinical-grade systems in evaluating intervention effects. More validation research is still wanting. The current study seeks to add to the growing knowledge base by comparing the full pattern of EEG results observed on a consumer-grade system with that observed on a medical-

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grade system. At the same time, it seeks to contribute to existing understanding on the effects of a deep breathing intervention—demonstrated to significantly reduce anxiety and improve test performance in children [5]—by examining its neurophysiological effects on attentional control.

### *1.1. Deep breathing intervention for test anxiety*

Test anxiety is a prevalent problem for school-going children around the world, especially in societies with high stakes examinations [6–8]. Test anxiety can be a detriment to children’s psychological well-being and performance and can unnecessarily limit children’s test performance and hinder future educational and career progression. Associated with negative outcomes ranging from low achievement to depression and suicide ideation [9,10], early intervention in schools to equip students with skills to ameliorate the adverse effects of test anxiety is necessary.

One intervention technique found to help children with test anxiety is deep breathing. In a controlled experimental study, Khng [5] found that 5th-graders who spent a minute practicing deep breathing before a timed mathematics test reported a significantly greater reduction in state anxiety and increase in test performance, compared to children in the control group. It was hypothesized that the intervention might improve performance by enhancing attentional focus—in the form of inhibitory-control-of-attention to distractors assessed on a flanker task [11]. Test-anxiety has been hypothesized to affect an individual’s inhibitory-control-of-attention to distractors, such as worrying thoughts and visual distractors, interfering with the appropriate allocation of cognitive resources to task-relevant processing [12]. The flanker task requires one to focus one’s attention on a central target and suppress attention allocated to processing flanking distractors, and is commonly used to measure inhibitory-control-of-attention, in very young children to elderly adults. Flanker interference is typically reflected by the difference in performance between congruent/non-inhibitory and incongruent/inhibitory trials, where the distractors are respectively congruent and incongruent with the target (e.g., “< < < < <” vs. “< < > < <”). A larger behavioral flanker interference (representing reaction time or accuracy costs) reflects poorer attentional control to distractors. Deep breathing is expected to reduce test anxiety’s effect on inhibitory-control-of-attention, which can be measured as better performance (i.e., smaller interference) on the flanker task. However, Khng’s study did not find a significant reduction in behavioral measures of flanker interference following the intervention.

### *1.2. Using EEG data to validate interventions*

Enhancement in brain functions arising from cognitive interventions such as attention training may be reflected on a neurophysiological level without direct improvements in behavioral performance. In one study for example, although a group of preschool children who underwent attention training did not show significantly improved behavioral performance on an attention task, the EEG data suggested enhanced efficiency in executive attention [13]. It is possible that the deep breathing intervention in [5] similarly facilitated attentional focus at a neurophysiological level without overt behavioral effects. This enhanced neural efficiency might have then facilitated better math test performance.

Neurophysiological data such as EEG are thus important sources of information for validating the efficacy or effects of interventions. Recording EEG data from children in school settings is however, rarely done. Conventional high-resolution, clinical/research-grade EEG equipment tend to involve high costs and long preparation times, limiting their utility in large-scale data collection in school-based research. Furthermore, the fitting of conventional gel-based electrodes may cause some discomfort and diminish children’s willingness to participate in EEG studies [14]. In recent years however, low-cost and lightweight alternatives have become available. For example, the Emotiv EPOC + wireless EEG headset (Emotiv Systems Inc., San Francisco, CA, USA) can be easily set up in a short time and worn without significant discomfort. Although they tend to have a lower resolution and signal-to-noise ratio than conventional research-grade systems, EEG data recorded on consumer systems such as the Emotiv EPOC + have been used to successfully generate event-related potentials (ERPs) and power spectra measures [3,15,16], renewing the possibility of collecting EEG data in ecologically valid settings during large-scale, school-based interventions.

### *1.3. The present study*

The current study explores the feasibility of using the Emotiv EPOC + system to collect research-grade EEG data in interventions for children conducted in schools. At the same time, we explore the behavioral and neurophysiological effects of the deep breathing intervention on inhibitory control of attention in a flanker task. Two parallel within-subject experiments are conducted on separate groups of participants, differing only in the EEG system used to record EEG data: Emotiv EPOC + or Neurostyle NS-EEG-D1 (Neurostyle Pte Ltd, Singapore). Of interest is whether the Emotiv system is able to detect effects of a deep breathing intervention for children on a flanker attention task, on the level of brain electrical activity. We also compare, descriptively, the pattern of EEG results obtained across the two systems to better understand the limitations of the consumer-grade headsets. Results from both sets of data will add to current understanding of the effects of taking deep breaths before a task on one’s neurophysiological state during task performance.

## **2. Method**

### *2.1. Participants*

Participants were 45 right-handed Primary Five students (∼Grade 5) from five local schools, randomly assigned to the Neurostyle group ($N = 22$; $M_{age} = 11.67$; $SD = 0.36$; 15 males) or the Emotiv group ($N = 23$; $M_{age} = 11.82$; $SD = 0.27$; 11 males). Children with known developmental or neurological disorders, or without normal or corrected-to-normal vision, were excluded. Informed parental consent and child assent were obtained from all participants. The conduct of the study was approved by the institutional review board of the authors’ university. Data from participants were excluded for insufficient data due to recording errors, unreliable task performance (close to chance level accuracy), or extreme noise (for EEG). Usable EEG, behavioral, and respiratory data were respectively available from 32, 37, and 25 participants. Details of the sample contributing to each valid dataset and their overlaps are presented in Table 1, together with study parameters. It should be noted that the respiratory data is analyzed separately on its own (see *Statistical analyses* and *Results and Discussion*), with the final analysis combined over the two EEG groups (differences in respiratory data is not expected to and did not differ between EEG groups). The overlap between the EEG and behavioral datasets within each group (∼90%) is more important for interpreting the current results as correlations between EEG and behavioral indices are examined.

### *2.2. Materials*

A computerized, modified flanker task [11,17] written with E-Prime was administered. Stimuli were centrally presented in white, 32-point Arial font on a black background. Each trial began with a fixation cross lasting 500 ms, followed by a blank slide of 500ms, and then a central arrowhead flanked by two arrowheads on either side. The flankers pointed in the same direction as the central target in the congruent condition (e.g., “< < < < <”), and in the opposite direction in the incongruent condition (e.g., “> > < > >”). Participants were required to focus their attention on the central target and to respond as quickly as they accurately can by pressing a left or right response key to indicate the direction the arrowhead is pointing. The stimulus display was terminated on response or after 1500 ms. An inter-stimulus interval (ISI) blank of 1100 ms, 1300 ms, or 1500 ms followed (Fig. 1). A total of 150 congruent trials and 150 incongruent trials were administered in equal proportions over 6 blocks. Behavioral data on reaction time (RT) and accuracy were collected.

**Table 1**
Study Parameters: Sample Distribution, EEG System Information, Conditions and Congruency.

| | | Neurostyle | | | | Emotiv | | | | Total |
|---|---|---|---|---|---|---|---|---|---|---|
| Sample distribution in valid datasets | Group *N (Male N)* | 22 (15) | | | | 23 (11) | | | | 45 |
| | EEG data | 17 (12) | | | | 15 (7) | | | | 32 |
| | Behavioral data | 17 (12) | | | | 20 (9) | | | | 37 |
| | Respiratory data | 18 (11) | | | | 7 (2) | | | | 25 |
| | Overlap *N (% of max possible)* | | | | | | | | | |
| | EEG & Behavioral | 15 (88%) | | | | 14 (93%) | | | | |
| | EEG & Respiratory | 15 (88%) | | | | 6 (86%) | | | | |
| | Behavioral & Respiratory | 15 (88%) | | | | 4 (57%) | | | | |
| | EEG & Behavioral & Respiratory | 13 (76%) | | | | 4 (57%) | | | | |
| EEG system parameters | Channels | 24 | | | | 14 | | | | |
| | Electrodes | FP1, FP2, AF9, AF10, F3, F4, F7, F8, Fz, C3, C4, Cz, T7, T8, P3, P4, P7, P8, Pz, O1, O2, Oz | | | | AF3, AF4, F3, F4, F7, F8, FC5, FC6, T7, T8, P7, P8, O1, O2 | | | | |
| | Sensor/ conductance | Ag-AgCl/gel | | | | Gold-plated/saline | | | | |
| | Sampling rate | 256 Hz | | | | 128 Hz | | | | |
| | Fit | Stretch mesh cap | | | | Rigid headset | | | | |
| | Set-up time | 30–45 min | | | | 5–10 min | | | | |
| Within-subject | Conditions | Deep breathing | | Control | | Deep breathing | | Control | | |
| | Congruency | Cong. | Incong. | Cong. | Incong. | Cong. | Incong. | Cong. | Incong. | |

Note. Cong., congruent; Incong., incongruent. All participants experience 150 congruent and 150 incongruent trials, in 6 blocks of 50 mixed trials, under both deep breathing and control conditions.

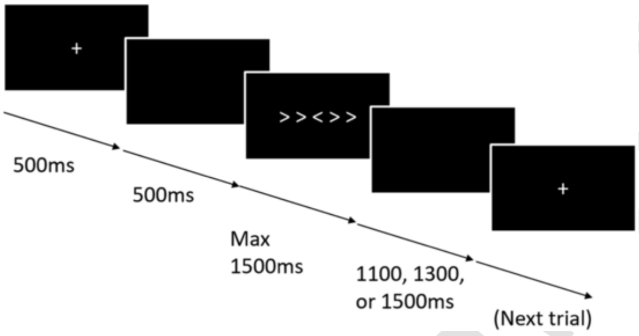

**Fig. 1.** Flanker task.

### 2.3. Procedure

We tested participants individually in their respective schools, in a quiet room away from sources of major electrical interference (e.g., elevators). Each participant was fitted with the respective EEG headset (Emotiv) or mesh cap (Neurostyle). Four minutes of resting state EEG (eyes-open) was recorded before participants were fitted with a respiration belt. Participants were then instructed on the flanker task and completed a practice block followed by the six experimental blocks with a one-minute rest between blocks. Respiratory and EEG activity were recorded throughout the task. The flanker task was administered to each participant twice, under a deep breathing condition and a control condition, in counterbalanced task order across participants. In the deep breathing condition, participants were taught to breathe slowly, deeply, and gently into their bellies. They were instructed to place their palms on their lower abdomens and to watch/feel the rise-and-fall with each inhalation/exhalation, as they engaged in a brief practice. To encourage slow, deep breathing, participants were instructed to count their breaths to see if they could reduce the number of breaths they needed to take. After observing that the deep breathing could be executed as required, participants were given a one-minute practice of deep breathing followed by the flanker task. Participants practiced deep breathing during the one-minute rest breaks in between blocks. No visual nor other forms of feedback on their respiration (e.g., real time graphical visualization of respiratory recording) were given to the participants during the deep breathing practice. In the control condition, the one-minute-deep-breathing segments were replaced with one-minute-rest segments. A 10-minute rest break was given between the two conditions.

### 2.4. Respiratory recording and data processing

Respiratory abdominal movement during the flanker task was measured with a Thought Technology respiration sensor belt (Thought Technology, Ltd., Quebec, Canada) and the accompanying BioGraph Infinity software. The IBM SPSS Statistics for Windows, version 24 software (IBM Corp., Armonk, NY, USA) was used to further compute, after discarding the first and last minute of recording, the respiratory rate (average number of breaths per minute) and respiratory amplitude (average amplitude per minute) during the flanker task in each condition.

### 2.5. EEG recording and data processing

For the Emotiv group, EEG data was collected on the Emotiv EPOC+ (Emotiv Systems Inc., San Francisco, CA, USA). The 14-channel rigid headset held gold-plated electrodes at the international 10–20 positions (see Table 1 and Fig. 2), referenced to the mastoids. Data was acquired wirelessly at a sampling frequency of 128 Hz. The sensors were moistened with a saline solution just before the headsets were fitted on the participants. Sensor contact quality was monitored with Emotiv's inbuilt program. For the Neurostyle group, EEG data was acquired using the 24-channel Neurostyle NS-EEG-D1 system (Neurostyle Pte Ltd, Singapore) at a sampling frequency of 256 Hz. An Easycap (EASYCAP GmbH, Germany) stretch mesh cap customized to fit children's heads was used to position the gel electrodes according to the modified combinatorial nomenclature extended from the 10–10 system [18]. Signal quality was assessed with the system's inbuilt impedance testing before recording commenced.

EEG data processing and artefact rejection was performed offline on MATLAB (version R2017a, The MathWorks, Inc., Natick, MA, USA). First, the continuous EEG data recorded during the flanker task was bandpass filtered between 1 and 30 Hz using zero-phase FIR filter with a Hamming window func-

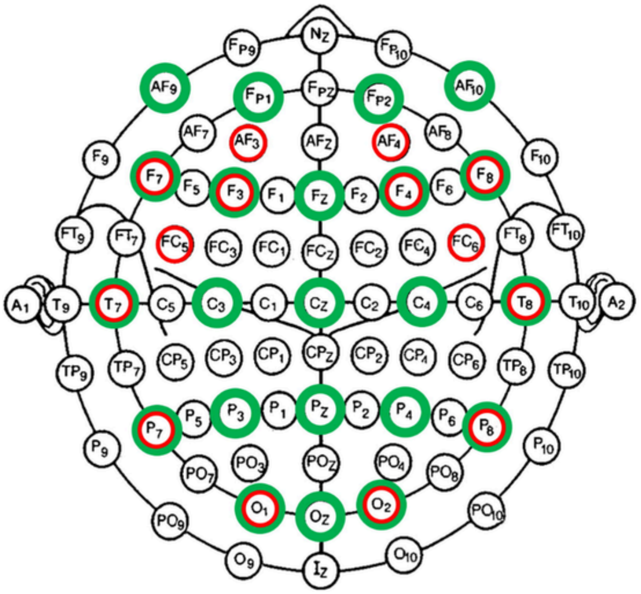

**Fig. 2.** Montage of Emotiv (red rings) and Neurostyle (green rings) electrode placement superimposed on the modified combinatorial nomenclature of the 10–10-system. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

tion to remove the DC drift and power line noise. Next, single-trial data extracted from 0.5s pre-stimulus to 1s post-stimulus (−0.5s to 1s) was extracted for further processing. The single trial data was then scanned for artifactual trials and the noise-affected trials were identified and discarded using a fully automated statistical thresholding method (FASTER toolbox). The clean trials were then inspected for noisy channels using a PREP toolbox [19] and the noisy channels were reconstructed using a spherical interpolation. Next, the data was re-referenced to a common average referencing to reduce muscle related artifacts. Finally, to fully remove eye blink and muscle-related artifacts, the data was decomposed using Independent Component Analysis (ICA). From the decomposed components, the artifactual components corresponding to the eye-blinks and muscle movements were identified using an automatic EEG artifact detector based on the joint use of spatial and temporal features (ADJUST toolbox) [20]. The artifact free data was then reconstructed from ICA components by excluding the artifactual components.

#### *2.5.1. Power spectral analysis*

The Welch's periodogram was used to calculate power spectral density (PSD) from 0 to 1 s single trial data. Average trial PSD was computed separately for accurate congruent and incongruent trials and respectively summed over the 1.00–4.00 Hz, 4.01–7.50 Hz, 7.51–12.50 Hz, and 12.51–30.00 Hz bands to derive absolute band power in the respective delta (δ), theta (θ), alpha (α), and beta (β) frequency bands. Absolute power in each frequency band was divided by the total power over 1–30 Hz to derive relative band power for each channel. Relative power represents the percentage of the amplitude in a respective frequency band against the total amplitude across all frequency bands and thus represents how much of the overall signal is accounted for by a particular band.

#### *2.5.2. Topological maps*

The topological maps of group-averaged relative power in each frequency band (by condition and congruency type) obtained on the two systems can be found in Fig. 3. As might be expected, the Neurostyle system had better coverage and was more sensitive at picking up EEG activity than the Emotiv system. The signal to noise ratio was generally poorer in the Emotiv system, contributing towards a smaller proportion of usable data compared to the Neurostyle (see *Participants*). Nevertheless, a visual comparison of the topological maps suggests substantial overlap between the two sets of recordings, particularly for the theta, and to some extent, beta frequency bands. Most markedly missed from the Emotiv seems to be fronto-central alpha, and anterior-frontal delta.

### *2.6. Statistical analysis*

Statistical analyses were performed on SPSS. Behavioral and respiratory data were combined across Neurostyle and Emotiv groups; EEG data was analyzed separately. Behavioral data was first checked for differences between the two groups by conducting the three-way, Group (Neurostyle vs. Emotiv) × Condition (deep breathing vs. control) × Congruency (congruent vs. incongruent) analyses of variance (ANOVAs) separately on mean RT and accuracy. If no interactions by Group were found, the two-way Condition × Congruency ANOVAs of interest were performed and interpreted. Inaccurate trials were excluded from the calculation of RT. Similarly, respiratory data was first examined with Group × Condition ANOVAs on respiratory rate and respiratory amplitude. Single-factor (Condition) ANOVAs were performed and interpreted following the absence of Group effects. We examined the EEG data on accurate flanker trials in three-way, fully-within repeated measures ANOVAs (Condition × Congruency × Electrode), separately on absolute and relative power from each frequency band. The Greenhouse-Geiser epsilon correction was used when assumptions of sphericity were violated. Missing data was excluded in a pairwise fashion. Unless otherwise stated, statistical significance was evaluated at $\alpha = 0.05$. Marginal significance refer to $0.05 < p < 0.10$.

## 3. Results and discussion

### *3.1. Behavioral measures*

No significant differences were found between the two groups on RT or accuracy. Consistent with the typical Flanker interference effect, significant main effects of Congruency were found on RT [$F_{1, 36} = 190.16$, $p = 0.000$, $\eta\rho^2 = 0.84$] and accuracy [$F_{1, 36} = 60.48$, $p = 0.000$, $\eta\rho^2 = 0.63$]. In both conditions (deep breathing and control), RTs were significantly faster and accuracy higher for congruent than incongruent trials. No significant intervention effects were observed on RT or accuracy: main and interaction effects involving Condition did not reach statistical significance (Table 2). The deep breathing intervention did not seem to have reduced flanker interference on the behavioural level. It also did not seem to have enhanced general performance on the task (i.e., greater speed or accuracy).

### *3.2. Respiratory measures*

No significant Group × Condition interaction was observed on respiratory rate or amplitude. Single-factor ANOVAs showed a main effect of Condition on respiratory rate [$F_{1, 24} = 6.19$, $p = 0.020$, $\eta\rho^2 = 0.21$]. The breathing intervention significantly slowed down breathing pace, with the average respiratory rate lower in the breathing ($M = 21.72$; $SD = 3.41$) compared to the control condition ($M = 23.60$; $SD = 4.01$).

### *3.3. EEG measures*

In the interest of space, we focus on effects and comparisons of interest between the systems. Of particular interest is the two-way interaction between Congruency × Condition (i.e., how does spectra power during congruent and incongruent flanker trials change after deep breathing), and its three-way interaction with Electrode (i.e., which recording sites are sensitive to these intervention effects). Significant higher-order (e.g., three-way) interactions qualify lower-order (e.g., two-way) interactions.

#### *3.3.1. Absolute power*

In the Neurostyle group, the three-way Condition × Congruency × Electrode ANOVAs showed a marginally significant Condition × Congruency interaction effect in the beta band [$F_{1, 16} = 3.85$, $p = 0.067$, $\eta\rho^2 = 0.19$], with significantly higher power for congruent compared to incongruent trials in the Control condition only. There were no significant three-way interactions.

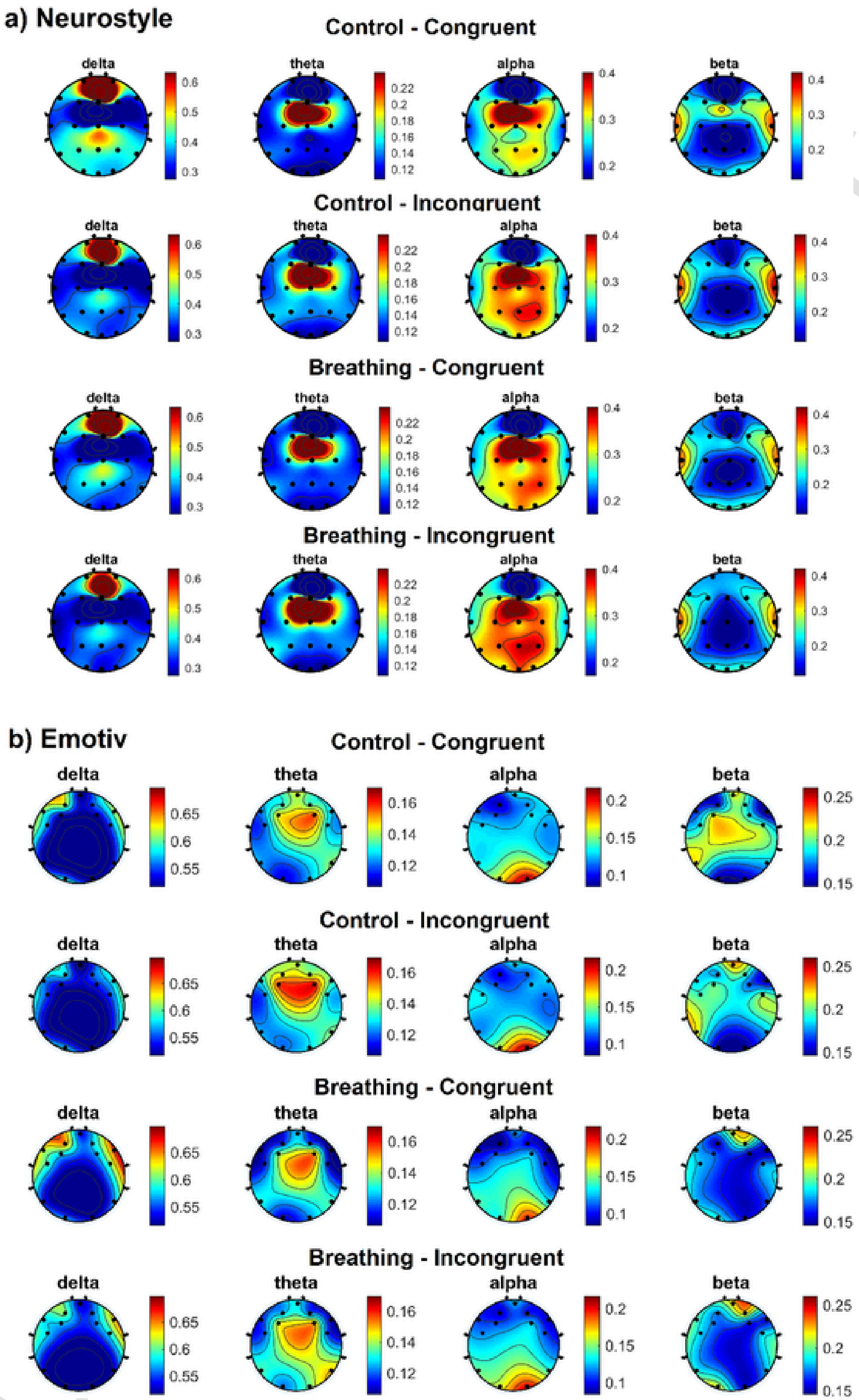

**Fig. 3.** Topological maps of relative power in (a) Neurostyle and (b) Emotiv systems.

**Table 2**
Behavioral Measures of Flanker Task Performance.

| | Control | | Breathing | |
|---|---|---|---|---|
| | Congruent Mean (SD) | Incongruent Mean (SD) | Congruent Mean (SD) | Incongruent Mean (SD) |
| RT (ms) | 584.12 (53.44) | 738.32 (92.80) | 598.07 (60.43) | 746.79 (95.90) |
| Acc (%) | 96.56 (6.10) | 88.86 (8.42) | 96.34 (5.02) | 87.77 (9.96) |

In the Emotiv group, there were no significant main or interaction effects in the three-way ANOVAs.

### 3.3.2. Relative power

In the Neurostyle group, the Condition × Congruency interaction effect was marginally significant in the beta [$F_{1,\,16} = 4.00$, $p = 0.063$, $\eta\rho^2 = 0.20$] band, with marginally higher power during incongruent compared to congruent trials in the Control condition only. This was however qualified by a Condition × Congruency × Electrode interaction [$F_{5.60,\,89.52} = 2.31$, $p = 0.044$, $\eta\rho^2 = 0.13$], also significant in the alpha [$F_{5.03,\,80.44} = 2.62$, $p = 0.030$, $\eta\rho^2 = 0.14$], and delta [$F_{5.66,\,90.57} = 3.38$, $p = 0.006$, $\eta\rho^2 = 0.17$] bands, and marginally significant in theta [$F_{6.26,\,100.23} = 1.94$, $p = 0.079$, $\eta\rho^2 = 0.11$]. We probed these three-way interactions by examining the two-way Condition × Congruency interactions paneled by Electrode to understand which recording sites reflected modulation of flanker interference (Congruency effects) by Condition, and in what way (Table 3a).

Significant modulation of alpha was seen over bilateral frontal (F3, F8), central (C3, C4) and right parietal (P4) sites. Alpha power during congruent trials seemed more amenable to the effects of deep breathing than during incongruent trials, with increased congruent-power over C3, and P8 and a frontal power reduction (F3). Power increase during incongruent trials were small and non-significant. Apart from over frontal sites, power differences between congruent and incongruent trials seemed generally reduced in the breathing condition: significant differences in power between trial types (incongruent > congruent; C3, Cz, P4) became non-significant—likely due to the increased power during congruent trials. The congruent-power reduction over frontal sites, conversely, led to significant power differences in the breathing condition (incongruent > congruent; significant in F8). We correlated these power changes with corresponding behavioral measures of flanker performance (e.g., change in congruent power with congruent RT and accuracy; change in congruency power difference (i.e., congruent vs. incongruent trials) with interference RT and accuracy (i.e., Incongruent RT or Acc – Congruent RT or Acc)) The change in congruent-power at F3 was significantly correlated ($r = 0.58$, $p < .05$) with change in congruent RT from control to breathing condition.

Significant modulation of beta was seen over the right frontal (F8), temporal (T8), and parietal (P4, P8), and left occipital (O1) sites. Beta power seemed more amenable to the effects of deep breathing during congruent than incongruent trials, with increased congruent-power over most sites (significant at P4, O1, Oz). Breathing effects on incongruent trials were apparent over T8 and P8: incongruent-power was reduced significantly at T8 and marginally at P8. Congruency differences seemed generally reduced in the breathing condition: significant power differences between trial types became non-significant (F8, T8, P4, F3). The congruent-power change at O1 was correlated ($r = 0.59$, $p < .05$) with the corresponding change in congruent accuracy in the breathing condition. Changes in congruency power differences from control to breathing condition at F3 ($r = -0.54$, $p < .05$) and Oz ($r = -0.54$, $p < .05$) were correlated with corresponding changes in RT interference; at Oz ($r = -0.67$, $p < .01$) with changes in accuracy interference.

Significant modulation of delta was seen over left-mid frontal (F3, Fz), bilateral central (C3, C4), right temporal (T8), mid-right parietal (P4, P8, Pz), and right occipital sites (O2). Delta power seemed more amenable to the effects of deep breathing during congruent than incongruent trials, but associated with

**Table 3**
Relative Power Pairwise Comparisons in Condition × Congruency ANOVAs panelled by Electrode: a) Neurostyle system; b) Emotiv system. Elec., Electrode; a, alpha frequency band; b, beta frequency band; d, delta frequency band; t, theta frequency band. Only pairwise contrasts at least marginally significant at (Bonferroni-corrected) $p < .10$ reported. Boldface frequencies are $p < .05$. Superscript on electrodes show frequencies with significant (boldface) or marginally significant Condition × Congruency interaction effects.

| | a) Neurostyle | | | | | | | | | b) Emotiv | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | Congruent (C) | | Incongruent (I) | | Control (N) | | Breathing (B) | | | Congruent (C) | | Incongruent (I) | | Control (N) | | Breathing (B) | |
| Elec. | B > N | N > B | B > N | N > B | I > C | C > I | I > C | C > I | Elec. | B > N | N > B | B > N | N > B | I > C | C > I | I > C | C > I |
| AF9$^{\mathbf{t}}$ | | | | | a, b, t | d | a, b | d | – | – | – | – | – | – | – | – | – |
| AF10 | | | | t | a, b, t | d | a, b | d | – | – | – | – | – | – | – | – | – |
| FP1$^{\mathbf{t}}$ | b | d | | | a, b, t | d | a | d | AF3 | | | | | b, t | d | a, b | d |
| FP2 | | | | | **a**, t | d | a, b, t | d | AF4 | | | | | | | b | |
| F3$^{\mathbf{a}\ b\ \mathbf{d}}$ | d | a | | | | b | t | d | F3 | | | | | t | | | |
| F4 | | | | | | | t | | F4 | | | | | | | | |
| F7 | | | | | t | | a | d | F7 | | a | | a | b, **t** | d | a | d |
| F8$^{a\ b\ t}$ | | t | | | b | | a, t | b, **d** | F8 | | a | | | t | | **a**, b, t | d |
| Fz$^{\mathbf{d}}$ | d | b | | b | | b | t | | – | – | – | – | – | – | – | – | – |
| C3$^{\mathbf{a}\ \mathbf{d}}$ | a | d | | | a, t | d | t | | FC5 | | | | | | | | |
| C4$^{\mathbf{a}\ \mathbf{d}}$ | a | | | | a, t | d | t | | FC6 | | | | | | | | |
| Cz$^{a}$ | a | | | | a, t | d | t | d | – | – | – | – | – | – | – | – | – |
| T7 | | | a | | a, t | d | a | d | T7 | | | | | b | | | |
| T8$^{\mathbf{b}\ \mathbf{d}}$ | a | | | b | b | d | | | T8 | | | | | | | | |
| P3 | | | | | a, t | d | a | d | – | – | – | – | – | – | – | – | – |
| P4$^{a\ b\ d\ t}$ | a, b | d | | | a, b, t | d | | | – | – | – | – | – | – | – | – | – |
| P7 | a | d | a | d | a | d | a | d | P7$^{\mathbf{a}}$ | | | | | | | | |
| P8$^{a\ \mathbf{b}\ \mathbf{d}}$ | a, t | d | | b | t | d | | | P8$^{\mathbf{b}\ \mathbf{d}\ t}$ | | | | | | | b, **t** | d |
| Pz$^{\mathbf{d}}$ | b | d | | | a, t | d | a | d | | | | | | | | | |

generally reduced congruent-power particularly over the more posterior regions (P4, P8, O1, Oz). Similar to that in alpha, a reverse pattern was observed over the over left-mid frontal sites (F3), where breathing increased congruent-power. Power reduction during incongruent trials was observed over Oz. Congruency differences seemed reduced in the breathing condition in almost all but the frontal sites: significant differences between trial types (congruent > incongruent, C3, C4, T8, P4, Pz) became non-significant or reduced (Pz). Congruent-power became significantly higher than incongruent-power at F3. The congruent-power change in at F3 correlated ($r = -0.58$, $p < .05$) with congruent RT change; at O1 ($r = -0.54$, $p < .05$), and P8 ($r = -0.52$, $p < .05$) with congruent accuracy change. Changes in congruency power differences from control to breathing condition at F3 ($r = -0.54$, $p < .05$) and Oz ($r = -0.54$, $p < .05$) were correlated with corresponding changes in RT interference; at Oz ($r = -0.67$, $p < .01$) with changes in accuracy interference.

Limited modulation of the theta band was found over AF9, FP1, F8, and O2. Breathing significantly reduced theta in congruent trials over F8. A significant congruency difference (incongruent > congruent) was eliminated in the breathing condition over left anterior-frontal sites (AF9, FP1). At O2, breathing affected congruent (increase) and incongruent (decrease) trials in different directions though the simple effects were not significant. These theta power changes were not correlated with performance change.

In the Emotiv group, the Condition × Congruency × Electrode ANOVAs showed significant Condition × Congruency interaction effects only in the alpha band [$F_{1,\,14} = 6.92$, $p = 0.020$, $\eta\rho^2 = 0.33$]: power was significantly higher during incongruent compared to congruent trials in the breathing condition, but not in the control condition. Although there were no significant three-way interactions, considering the small sample size (and large number of electrodes), we proceeded with exploring the paneled two-way interactions. Compared to the Neurostyle data, there were much fewer pairs of significant contrasts. For relative alpha, most of the breathing effects observed involved significantly higher power for incongruent compared to congruent trials during the breathing condition only (AF3, F8, O1), although the Condition × Congruency interaction effect was significant only at O1 (also at P7 but simple effects were not significant) (see Table 3b). A similar modulation at P8 was significant in beta (incongruent > congruent marginally significant only in breathing condition) and marginally significant in theta. Delta modulation was significant at P8 and marginally significant at O1, with higher power for congruent than incongruent trials during the breathing condition. Changes in theta congruency power differences from control to breathing condition at P8 correlated with changes in RT interference ($r = 0.53$, $p < .05$).

Taken together with the topographical findings, it seems that the Emotiv system may not be able to capture effects particularly around the fronto-central, temporal, and parietal regions—at least for the effects of interest in the current paradigm. As can be seen in Table 2, a much more limited subset of effects could be observed with the Emotiv—though on a broad level, there are similarities/overlaps. For instance, the pattern of contrasts in the anterior-frontal region recorded by Emotiv's AF3 was quite similar to the general pattern reflected by the anterior-frontal set of recordings on the Neurostyle (AF9 – FP2). To a lesser extent, there were also some overlaps between the homologous pairs of inferior frontal electrodes (F7, F8), especially pertaining to breathing effects on congruency differences, which are of particular interest in this study. The other regions seemed more problematic. As previously described, there were almost no effects picked up by Emotiv's electrodes in the fronto-central (FC5, FC6), temporal (T7, T8), and parietal (especially P7) regions, while significant modulation effects of interest were picked up by Neurostyle's electrodes in these areas. Although Emotiv's homologous electrodes at P8, O1 and O2 did pick up some effects like the Neurostyle, the pattern of results differed substantially.

## 4. General discussion

As per the previous behavioral study [5], effects of the deep breathing intervention were not observable in behavioral performance on reducing the rate of respiration (i.e., slowing down breathing). At the same time, we found some evidence of the intervention having an effect on a neurofunctional level, based on data collected by both a high-resolution medical-grade EEG system and a low-resolution consumer-grade system.

The high-res system provided stronger evidence of intervention effects. Overall, results suggest that deep breathing tended to increase relative alpha and beta power and decrease delta and theta power during baseline flanker task performance (congruent trials), and reduce relative power differences between these baseline, non-inhibitory (congruent) trials and trials requiring inhibitory control of attention (inhibitory, incongruent trials). Some intervention effects on incongruent trials were observed in the form of reduced relative beta power over the right temporal-parietal region, and in reduced delta over the mid-occipital site. However, these were not correlated with changes in behavioral performance.

Although intervention effects on congruent trials were of less interest, an interesting exception to the congruent-power change pattern in the alpha and delta bands was observed over the left superior frontal area where deep breathing decreased alpha and increased delta on congruent trials—which were strongly associated with a reduction in RT during congruent trials. The congruent-power increase in the beta band observed over the left occipital area was also strongly associated with an increase in congruent-trial accuracy. Alpha oscillations have been described to play an active inhibitory role [21,22]. Delta—especially lower delta (1–2 Hz)—activity has been suggested to serve a more general action-monitoring role in the flanker task, towards the prevention of errors and more accurate task performance [23]. Beta activity has been associated with alertness and task engagement; occipital beta activity especially plays an important role in visual discrimination and attention [24]. Our current finding (and the related correlation between changes in congruency power differences and behavioral interference) is consistent with an account of deep breathing modulating the relative power distribution in these frequency bands to enhance performance on congruent trials.

Although significant results from the lo-res system were limited, current findings showed significantly higher relative alpha power during incongruent compared to congruent trials only in the breathing condition. Based on the view that alpha oscillations play an inhibitory role, it has been posited that elevated alpha during incongruent trials of the flanker task reflect enhanced top-down cognitive control [21,22]. Deep breathing might have increased attentional focus in the face of distractors, by powering up alpha. Results of the Condition × Congruency interactions paneled by Electrode suggest that the effects of the intervention on relative alpha power are most likely to be observed at the left anterior frontal, bilateral frontal/fronto-central, and left occipital sites. Although this last set of results followed a Condition × Congruency × Electrode interaction that failed to reach statistical significance, this might also have been partly due to the limited power in the current study to detect significant effects with the small sample size, number of sites examined, and lower power of the consumer-grade system. The three-way interaction may emerge significant with a larger sample size. Nonetheless, though this set of results may be tentative, they provide some insights that will be useful for future reference or validation.

We note that most of our effects were found with relative band power, while marginal or insignificant effects were found with absolute power. Studies of cognitive processes commonly examine both, but rarely discuss underlying differences. Frequencies in absolute band power are independent of each other whereas relative power takes into account the concurrent power in other bands. An increase in band power relative to other bands, may thus reflect changes in dynamic neural resource allocation [25], which is relevant for the context of the current study. During tasks such as the flanker, neural activity in different frequency bands are proposed to be implicated in different phases/stages of interference processing and response [23,26]. It is suggested that oscillations in each frequency range may serve multiple functions, and that specific configurations of oscillations across various frequency bands in selectively distributed networks give rise to complex, integrative brain functions [27,28]. Fndings from the present study are broadly in line with the literature on the role of top-down executive power is consistent with cognitive resource allocation theories of test anxiety (as earlier described) [12,29].

The current findings should be interpreted in light of some limitations. As previously discussed, the small sample size may have limited the power the current study held to detect significant effects; similar studies should be conducted with larger samples in the future, especially if low-powered commercial devices are to be used. Due to the context of the intervention under investigation [5], the current sample comprised a mix-gender, developmental population (~11 year-olds), including girls around the age of puberty. Despite efforts to balance the sample distribution on gender, our final datasets were not completely balanced. Brain electrical activity can differ with brain maturation and results may differ if varying proportions of children at different stages of maturation are included. At least on the level of behavioral measures, we did not find differences in flanker interference when we checked for gender effects in our sample.

## 5. Conclusion

The current study sought to explore effects of a deep-breathing intervention for test-anxiety on brain electrical activity and behavior performance during inhibitory-control-of-attention. Findings suggest that deep breathing may enhance attentional focus by modulating the allocation of neural resources for task-relevant/irrelevant processing. We also sought to examine the usability of a consumer-grade EEG system for school-based intervention research. Findings are consistent with previous studies that have examined EEG data collected on the Emotiv EPOC+, suggesting that, although there are marked limitations, the low cost, consumer-grade systems may still be able to provide some amount of useful data during large scale, school-based interventions.

## Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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