HAL open science

Human Postural Responses to High Vestibular Specific Extremely Low-Frequency Magnetic Stimulations

Nicolas Bouisset, Sebastien Villard, A. Legros

▶ **To cite this version:**

Nicolas Bouisset, Sebastien Villard, A. Legros. Human Postural Responses to High Vestibular Specific Extremely Low-Frequency Magnetic Stimulations. IEEE Access, 2020, 8, pp.165387-165395. 10.1109/access.2020.3022554 . hal-03499662

HAL Id: hal-03499662

https://hal.science/hal-03499662v1

Submitted on 5 Jan 2022

HAL is a multi-disciplinary open access archive for the deposit and dissemination of scientific research documents, whether they are published or not. The documents may come from teaching and research institutions in France or abroad, or from public or private research centers.

L'archive ouverte pluridisciplinaire **HAL**, est destinée au dépôt et à la diffusion de documents scientifiques de niveau recherche, publiés ou non, émanant des établissements d'enseignement et de recherche français ou étrangers, des laboratoires publics ou privés.

Distributed under a Creative Commons Attribution 4.0 International License

IEEE Access
Multidisciplinary ⋮ Rapid Review ⋮ Open Access Journal

Received August 7, 2020, accepted August 30, 2020, date of publication September 7, 2020, date of current version September 22, 2020.

Digital Object Identifier 10.1109/ACCESS.2020.3022554

Human Postural Responses to High Vestibular Specific Extremely Low-Frequency Magnetic Stimulations

NICOLAS BOUISSET[1,2], SÉBASTIEN VILLARD[1,2], AND ALEXANDRE LEGROS[1,2,3,4,5]

[1]Human Threshold Research Group, Imaging, Lawson Health Research Institute, London, ON N6A 4V2, Canada
[2]Department of Kinesiology, Western University, London, ON N6A 3K7, Canada
[3]Department of Medical Biophysics, Western University, London, ON N6A 3K7, Canada
[4]Department of Medical Imaging, Western University, London, ON N6A 3K7, Canada
[5]EuroMov, Université de Montpellier, 34090 Montpellier, France

Corresponding author: Alexandre Legros (alegros@lawsonimaging.ca)

This work was supported in part by the Hydro-Québec, Canada, in part by the Electricité De France, France, in part by the Réseau de Transport d'Électricité, France, in part by the Lawson Internal Research funding, in part by the Mathematics of Information Technology and Complex Systems through the Mitacs-Accelerate Program, Canada, and in part by the Electric Power Research Institute (EPRI).

ABSTRACT ***Background***: International agencies recognize the lack of knowledge to further establish standards and guidelines to protect the workers and the public from extremely low-frequency magnetic fields (ELF-MF). In that regard, postural control has been proposed as a biomarker of potential adverse effects in humans. Considering its crucial role in postural control and its specific neurophysiological characteristics, the vestibular system emerges as an ELF-MF likely target. However, postural modulation to vestibular ELF-MF exposure remains inconclusive. Previous studies led us to investigate stimulation orientation and point of application to clarify the ELF-MF impact on balance in humans. ***Objectives:*** This research aimed to investigate the acute postural impact of lateral vestibular-specific ELF-MF stimulations. ***Methods*:** Postural control of thirty eight healthy participants was analyzed with lateral vestibular-specific ELF-MF stimulations ranging from 20 Hz to 160 Hz, up to 142 T/s and vestibular electrical stimulations at the same frequencies. Both spatial orientation and quantity of movement variables were used to investigate postural modulations. ***Results*:** Despite a conclusive positive control effect, no significant effects of ELF-MF and alternating current stimulation exposures were found regardless of frequency conditions. ***Conclusions***: Although important electric fields were generated, no postural modulation was found. However, at these frequencies, the potential vestibular activation did not translate into functional postural sway but might be observed with reflexive vestibular outcomes.

INDEX TERMS Electromagnetic induction, extremely low frequency magnetic fields, human vestibular system, postural control.

I. INTRODUCTION

The generation, distribution, and use of alternating current (AC) are ubiquitous in modern societies, exposing the public to 50/60 Hz Extremely Low-Frequency Magnetic fields (ELF-MF < 300 Hz). According to Faraday's law of induction, changing magnetic flux density over time (dB/dt, measured in T/s) induces Electric Fields (E-Fields) and currents within conductors such as the human body. In this context, answering health and safety concerns to protect workers and the public is crucial. In that regard, international agencies such as the International Commission for Non-Ionizing Radiation Protection (ICNIRP) and the International Committee on Electromagnetic Safety from the Institute of Electrical and Electronics Engineers (IEEE-ICES) review scientific data to establish guidelines and standards enacted at national levels [1]–[3].

The main experimental paradigm to investigate the acute consequences of electrostimulation emerging from induction in humans is the perception of magnetophosphenes. Magnetophosphenes are flickering visual manifestations perceived when exposed to sufficiently strong time-varying

The associate editor coordinating the review of this manuscript and approving it for publication was Bo Pu.

This work is licensed under a Creative Commons Attribution 4.0 License. For more information, see https://creativecommons.org/licenses/by/4.0/

MF [4]. The main hypothesis regarding magnetophosphenes is that they result from membrane potential modulations of graded potential retinal cells, impacting in cascade the continuous release of neurotransmitters through their ribbon synapses [5]. Interestingly, the retinal cells share common neurophysiological properties with the vestibular hair cells. Indeed, both types of cells use graded potential for signal processing [6] both releasing glutamate gradually from ribbon synapses [7]–[11].

Vestibular hair cells are found in both canals and otoliths (composed of the utricle and the saccule), responsible for detecting head rotational and linear accelerations respectively. Vestibular hair cells transduce mechanical information (i.e. head movements) into an electric signal treated by the central nervous system (CNS) [12]. Compellingly, as for the retinal cells [13], small intensity current stimulations easily trigger the vestibular hair cells [14]–[18] making them likely susceptible to ELF-MF induced currents.

Since vestibular, visual, and proprioceptive inputs [19], are integrated to manage balance through postural control [20] it was suggested that ELF-MF could impact postural sway. However, our previous investigations of vestibular ELF-MF stimulations did not show acute postural outcomes [21], [22], questioning the assumption that similar neurophysiological systems should respond equivalently to ELF-MF stimulations. We previously argued that the top-down orientation of our fields in regards to hair cells' orientation was not optimal for their modulation and that lateral field orientation could be better [22]. The impact of field orientation had also been demonstrated to be crucial in the case of magnetophosphene perception [23] which prompted further investigation of postural outcomes under lateral stimulation of the vestibular system. We previously attempted to address this question [21], however, clear methodological biases had to be answered to reach relevant conclusions.

Therefore, the main objective of the current work is to further investigate a potential acute vestibular impact of lateral ELF-MF stimulations at powerline frequencies (i.e 60 Hz in North America). To do so, we improved our previous study [21].

Given the close neurophysiological similarities between the retinal cells and the vestibular hair cells and the fact that both are triggered by electrical stimulations, we hypothesize that ELF-MF impact the vestibular hair cells modulating postural sway. Since greater currents cause greater vestibular outcomes [24], [25], and induced currents' strength proportionally increase with dB/dt [26], we hypothesized that higher dB/dt values yield larger postural modulations.

Lövsund *et al.* [27], illustrated the effect of dB/dt on magnetophosphenes' perception by comparing electro- and magneto-stimulations. Indeed, in the case of an electric stimulation the current intensity delivered is not changed by an increase of the stimulation frequency, whereas the increase in frequency for an ELF-MF stimulation proportionally increases the induced current intensity. Following the same paradigm, we compared vestibular specific ELF-MF and AC stimulations over increasing frequencies expecting to find different frequency effects in the postural responses.

II. METHODS

A. PARTICIPANTS

Thirty-eight healthy participants (16 females-22 males, 24.3 ± 3.51 years old) were recruited for the study and tested in the Human Threshold Research Facility at St. Joseph's Hospital in London, Ontario, Canada. Were excluded volunteers with a history of any vestibular-related pathology or dysfunction, chronic illnesses, neurological diseases that affect normal body movement, and participants having permanent metal devices above the neck. Participants had to refrain from exercise and alcohol, caffeine, or nicotine intake 24 hours before the study.

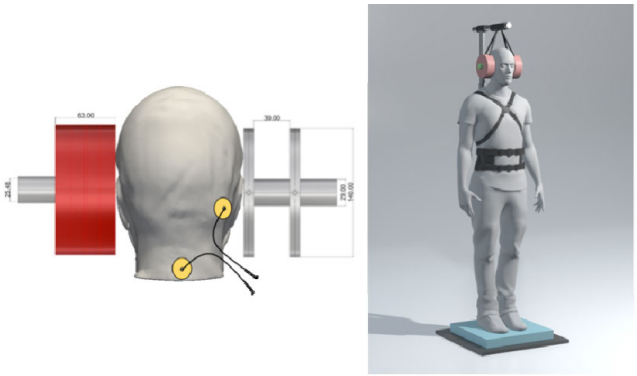

FIGURE 1. Experimental exposition apparatus. The left panel shows a diagram of the custom coils system centered over the mastoid process and the monaural electrode montage (yellow circles) delivering the DC and AC currents. The right panel shows a volunteer wearing the vest self-sustaining the MF headset device unloading the weight of the coils.

B. EXPERIMENTAL DEVICES

We delivered the MF vestibular specific stimulations to the subjects' right vestibular system via a customized headset coil exposure system (6.70 kg). It consisted of two 570 turn-coils of 5.9 cm of mean diameter, with a 2.5-cm diameter core of Permendur-49 (The Goodfellow Group, Coraopolis, PA, USA- see Fig.1 left panel) inserted within each coil. We chose a Permendur-49 core for its high and stable permeability properties enabling us to better confine and guide the MF towards the vestibular system. Also, as in our previous work [21], we used a Permendur-49 core to increase the flux density developed by the coil to reach 100 mT_{rms} (141.42 mT peak) at 3 cm from the coils where the vestibular system approximately lies [28]. The inductance of the coil was 26 mH. The two coils were bound together to a custom adjustable headset to better fit participants' heads (Fig. 1 right panel). Although we only stimulated the right vestibular system in this study, we kept both coils not to introduce any postural bias due to asymmetrical load. The whole headset was suspended by a rod system, designed to support up to 10.5 kg,

tied to a vest worn around participants' chests (Atlas Camera Support, Los Angeles, Ca, USA), to unload the weight of the coils as they were maintained on the participants' head (see Fig. 1 right panel).

We controlled the system and collected data using a custom LabVIEW™ script (LabVIEW 2014 version 14.0.1 (32 bit)) through a 16-bit National Instruments A/D Card output channel (National Instruments, Austin, TX), driving an MTS™ Magnetic Resonance Imaging gradient amplifier capable of delivering up to 200 A_{rms} at ± 345 V (MTS Automation, Horsham, PA, USA). We delivered Direct Current (DC) and AC stimulations using a transcranial current stimulation device (StarStim, Neuroelectrics, Spain) driven by the NIC software (Neuroelectrics Instrument Controller, version 1.4.1 Rev.2014-12-01) via Bluetooth. We used a force plate (OR6-7-1000, AMTI, USA) to collect participant's body sway at 1 kHz according to 6 degrees of freedom: forces and moments data each in the 3 dimensions. Data was saved in a single measurement file, along with the MTS™ amplifier's current time series, used to synchronize all measurements with MF exposures, for later analysis. The Center of Pressure (COP) trajectory was calculated post-recording using a calibration matrix provided by the manufacturer. No hardware filtering was applied.

C. PROTOCOL

We fully equipped the participants after they gave their written informed consent. We used the same monaural montage for both DC and AC stimulations (Fig. 1 left panel). DC was only used as a positive control condition to validate the choice of our dependent variables. For DC stimulation, we placed the cathode behind the right mastoid process and the return electrode at the C7 spinal process (see Fig. 1 left panel). To improve impedance, we rubbed the right mastoid and C7 spinal processes with alcohol wipes (Mooremedical, USA). To provide proper conduction between the electrodes and the skin, we saturated the circular 25 cm^2 Ag/AgCl electrodes (StarStim, Neuroelectrics, Spain) with 8 mL of saline solution. We then secured the electrodes using the StarStim exposure cap and tape. To ensure appropriate stimulations, we maintained electrodes' impedances below 10 kΩ throughout the experiment, as recommended by the manufacturer. Before starting the testing, we exposed the participants to 5 seconds DC (2 mA) and AC (peak ± 2 mA at 20 Hz) trials as stimulation samples. The MF headset exposure system was then set over the StarStim exposure cap. To ensure careful headset placement, we centered the coils at the mastoid processes level. For consistency, we kept both the StarStim cap and the MF exposure device on the head during all testing conditions.

We tested participants in periods of 20 seconds. We asked them to stand on a 6-cm thick foam pad (Airex AG, Switzerland) placed on the force plate, with the eyes closed, arms resting at their side, and feet together to maximize vestibular contribution [29]. A second investigator, blinded to the type of stimulation applied to the participants, was present to prevent potential falls and for safety purposes. Exposure conditions consisted of five seconds of MF (100 mT_{rms}), DC (2mA), AC (peak ± 2 mA), or no stimulation (CTRL). As in Villard *et al.* [21], we delivered MF and AC stimulations at five different frequencies (20 Hz, 60 Hz, 90 Hz, 120 Hz, and 160 Hz). All trials were randomly distributed. In a post-experiment analysis, we randomly assigned the CTRL trials to an experimental condition and we respectively tagged them as "CTRL DC", "CTRL AC" and "CTRL MF" for DC, AC and MF conditions. To avoid participant fatigue, dissipate the stimulation effects and allow the vestibular system to reach its normal resting firing rate between blocks, we gave 30 s of rest between trials [30]. Also, to avoid cerebrovascular alterations that could bias postural outcomes after standing back up, participants could relax but could not sit during the resting periods [31].

To analyze the effects of the ELF-MF stimulation device, we recorded the participant's postural control with and without the coils system. To keep the device and the electrodes consistently aligned with the mastoid processes throughout the experiment, these two final conditions were not randomized and were recorded at the end of the trials.

Subjects wore earplugs throughout the experiment to conceal the noise generated by the coils. The Health Sciences Research Ethics Board at Western University approved this protocol (#106122) performed following the Declaration of Helsinki.

D. DATA ANALYSIS

The COP time series were filtered with a low pass bidirectional 4th order Butterworth zero-phase digital filter with a cutoff frequency of 5 Hz. The use of a residual analysis with a customized Matlab program (MatLab version 9.3 – The MathWorks Inc., USA) determined the cutoff frequency. We computed sway characteristics using a customized Matlab program. Classically, sway variables are analyzed on anteroposterior (AP) and mediolateral (ML) axes separately, but here we favored planar analyzes over one-dimensional analyses for mainly two reasons listed thereafter. First, balance is best controlled by coordinating the body in space in both dimensions simultaneously [32]. Second, biomechanical factors bias AP-ML analyzes [33], [34], which may be particularly impacted in a protocol sensitizing the vestibular function and involving a heavy customized headset coil exposure system. Finally, monaural electrical stimulations were used (different from the binaural bipolar montage in [21], which induce oblique deviations in the AP and ML plane [35]. Therefore, we favored planar analyzes over one-dimensional analyses.

Electrical stimulations of the vestibular system impact both the quality (sway spatial orientation) and the quantity of movement (sway size) [29]. Both were therefore considered. To investigate the acute stimulation effects, we measured the sway differences between the 5 seconds period before stimulation onset (PRE-STIM) and the 5 second stimulation period (STIM) for all our analyzes.

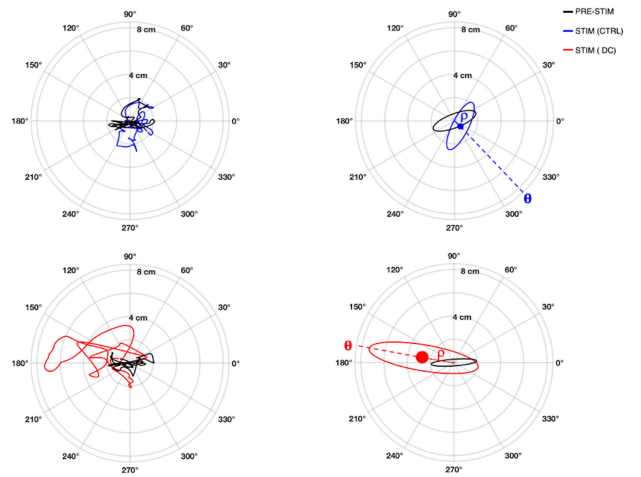

FIGURE 2. Postural sway for one participant. Left panels show the Center of Pressure (COP) before (black) and during the exposure (blue: CTRL, red: DC). The right panels show the 95% confidence ellipses encompassing these COPs. The displacement from the ellipses' barycenter from pre-stimulation to during-stimulation provides distance ρ and the angle θ. Finally, the size of the dot characterizes the difference of sway movement (calculated as COP average velocity) between pre- and during stimulation.

Spatial orientation was estimated by first conducting a Principal Component Analyses (PCA) on COP datasets to compute PRE-STIM and STIM 95% confidence interval ellipses [36] for each trial (Fig. 2 right panels). Then the barycenter was found at the intersection of the major and minor axes of each ellipse. To facilitate the analysis, the mean of the PRE-STIM COP dataset was subtracted from both the PRE-STIM and STIM datasets, centering all the PRE-STIM barycenters on zero. To estimate the spatial direction of sway we found the angle theta ($\boldsymbol{\theta}$) between 0 degrees and STIM barycenters (Fig. 2 right panels).

Two analyses were done for sway size. First, we calculated the distance rho (ρ) between PRE-STIM and STIM barycenters (Fig. 2 right panels). Then, among classical sway variables, the pathlength (the total length of COP excursion) has proved to be the more sensitive as well as the more reliable [37], [38]. Therefore, we computed both PRE-STIM and STIM pathlengths as the total sum of the distances between each point in the AP-ML plane (Fig. 2 Left panels). However, because pathlength varies with recording time it is often hard to compare results from one study to another. For this reason, we retained mean velocity (Pathlength over time). With transcranial electrical stimulations, great E-Field variability exists between participants [39], leading to great postural outcomes variability. Therefore, we calculated the difference between the STIM and PRE-STIM mean velocities (Δ speed) for each trial in order to individualize the analysis of the stimulations' impact.

E. STATISTICAL ANALYSIS

We performed all linear statistical analyses using R version 3.3.2 [40] and all circular statistics using the CircStat toolbox in Matlab [41]. A level of significance of $\alpha = 0.05$ was adopted throughout data analysis.

Differences in all three CTRL conditions were analyzed with a one-way repeated measure analysis of variance (ANOVA). To investigate the effect of wearing the stimulation device (ON vs OFF) as well as the effect of our positive control (DC vs CTRL), we implemented paired t-tests to analyze ρ and Δ speed. Two-way ANOVAs (2 stimulation modalities (AC / MF) $\times$ 6 conditions (CTRL plus five frequencies)) for repeated measures were used to test the effect of frequency of the time-varying exposure types on ρ and Δ speed. Rao's spacing test for circular uniformity was used to determine whether $\boldsymbol{\theta}$ was distributed uniformly. If not, the mean $\boldsymbol{\theta}$ and angular deviation ($\pm$ AD), as well as the mean resultant vector length ($||\vec{\boldsymbol{r}}||$), were implemented to describe the main direction of sways from PRE-STIM to STIM barycenters. $||\vec{\boldsymbol{r}}||$ is a measure of angular dispersion around the mean ranging from 0 to 1. The closer $||\vec{\boldsymbol{r}}||$, gets to 1, the more the angles are concentrated around the angular mean thus describing one specific direction [41].

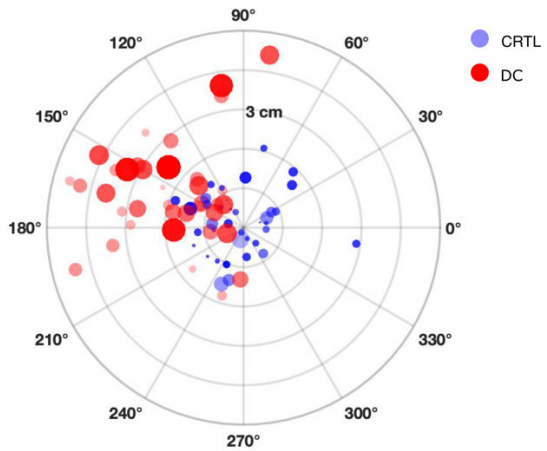

FIGURE 3. Postural shift of COP barycenters from pre- to during exposure. Pre-exposure barycenters are centered at the origin. Each dot location represents the displacement due to exposure. The dot size shows the absolute difference in Δ speed (amplitude only) while transparency shows actual Δ speed (amplitude and sign: most transparent express higher speed in pre-exposure).

III. RESULTS

A. DIFFERENCES IN CTRL CONDITIONS

As seen in Fig. 3 and Fig. 4, all CTRL conditions (blue dots) were equivalent. Indeed, no differences were found for ρ (F (2,74) = 1.58, $p = 0.21$) and for Δ speed (F (2,74) = 0.53, $p = 0.5907$)). Also, no specific sway directions were found in the different CTRL groups ($p > 0.05$).

B. EFFECT OF STIMULATION DEVICE

Postural sway size was not affected by wearing the headset as no significant effect was found on ρ (t (37) = -0.77, p = 0.44) nor on Δ speed (t (37) = 1.19, p = 0.24). Also, the headset did not organize sway spatially since Rao's tests in both conditions showed that $\boldsymbol{\theta}$ values were uniformly distributed ($p > 0.05$).

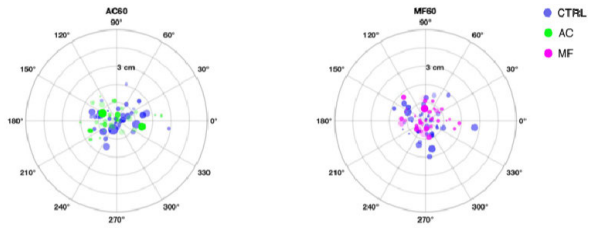

FIGURE 4. Postural modulations for AC (left) and MF (right) at 60 Hz compared to control conditions (blue dots). Results at 60 Hz are representative of all experimental conditions. Pre-exposure barycenters are centered at the origin. Each dot location represents the displacement due to exposure. The dot size shows the absolute difference in Δ speed (amplitude only) while transparency shows actual Δ speed (amplitude and sign: most transparent express higher speed in pre-exposure).

TABLE 1. Descriptive statistics for both stimulation groups (MF and AC) across all frequencies (20, 60, 90,120 and 160 Hz). Mean and standard deviation values for Δ speed and ρ. No information about θ is reported since no mean angle could be computed.

		CTRL	20 Hz	60 Hz	90 Hz	120 Hz	160 Hz
Δ speed (cm/s)	AC	-0.30 ± 2.20	0.35 ± 1.81	-0.31 ± 1.69	0.42 ± 1.67	-0.13 ± 1.71	0.17 ± 1.45
	MF	0.19 ± 1.72	0.45 ± 1.91	0.17 ± 1.85	-0.43 ± 1.59	-0.66 ± 1.32	0.07 ± 1.80
ρ (cm)	AC	0.94 ± 0.59	0.99 ± 0.55	0.95 ± 0.60	1.13 ± 0.47	1.13 ± 0.47	1.06 ± 0.64
	MF	0.96 ± 0.56	1.03 ± 0.70	0.83 ± 0.44	0.88 ± 0.50	0.93 ± 0.58	0.99 ± 0.60

C. EFFECTS OF POSITIVE CONTROL

Δ speed (t (37) = 7.81, p < 0.0001, R^2 = 0.62 - see Fig. 3) and ρ (t (37) = 6.15, p < 0.001, R^2 = 0.5 - see Fig. 3) were significantly greater with DC than CTRL signing more important sway size due to DC. Also DC clearly organised sway spatially. Indeed, as confirmed by the second experimenter, DC induced an important obvious left forward oblique postural sway (Mean $\boldsymbol{\theta} = 157.4° \pm 13°$, $||\vec{\boldsymbol{r}}|| = 0.82$, $p = 0.001$), whereas no significant mean sway was found for CTRL ($p > 0.05$), see Fig. 3).

D. EFFECTS OF AC AND MF STIMULATIONS

Fig. 4 depicts $\boldsymbol{\theta}$, ρ and Δ speed data for both stimulation types (MF and AC) at 60 Hz. Indeed, results at 60 Hz are representative of all frequency conditions (see table 1). Two-way ANOVAs (2 stimulation modalities (AC/MF) × 6 conditions (CTRL plus 5 frequencies)) for repeated measures indicated no significant main effects of stimulation condition for ρ ($F(1,37) = 2.8$, $p = 0.1$) nor on Δ speed ($F(1,37) = 0.80$, $p = 0.37$). Equally, no significant main effects of frequency were found for ρ: ($F(5,185) = 0.70$, $p = 0.62$) and for Δ speed ($F(5,185) = 1.83$, $p = 0.1$). Also, no interaction effects were found for ρ: ($F(5, 185) = 0.88$, $p = 0.49$) and for Δ speed: ($F(5,185) = 1.64$, $p = 0.15$). Rao's test results concerning $\boldsymbol{\theta}$ for all MF and AC experimental conditions as well as for CTRL groups, consistently showed that all angles in each condition were uniformly distributed ($p > 0.05$) underlining that neither MF nor AC stimulations oriented postural control in any given specific direction.

IV. DISCUSSION

This work was a follow up of a previous study from our group [21], from which the methodology and analysis have been improved to account for now known biases and planar modification of the COP.

The aim here was to study the potential acute effect of vestibular exposure to a power-frequency MF on postural outcomes. We hypothesized that MF-induced E-fields would trigger vestibular hair cells and modulate the postural sway in the same way they trigger magnetophosphene perception when applied to retinal cells. AC stimulation is known to impact postural control in humans [29], [42], and was thus used in comparison with the ELF-MF outcomes. However, since induction laws make an E-field's strength proportional to ELF-MF frequency but not with AC, we did not expect similar frequency modulations in the outcomes.

The use of a DC stimulation as a positive control validated the postural variables chosen in this work. The mean lowest DC threshold reported in the literature to induce postural responses in healthy controls is 0.32 mA [43]. Therefore, as expected, our 2 mA DC stimulation resulted in an instantaneous effect on human balance. With higher Δ speed and greater ρ, we observed a greater postural sway in DC than in the CTRL condition. As predicted, participants swayed towards the opposite side of the stimulated ear with a mean direction angle of 157.4° ± 13°, describing a left-oriented oblique forward sway expected for a right monaural cathodal DC stimulation [29]. However, we did not observe differences in sway size and spatial orientation neither with ELF-MF nor AC stimulations.

Since our 2 mA AC stimulation was over 6-fold higher than the reported postural threshold, and since our DC stimulation at the same intensity triggered a postural response, the absence of AC results must, therefore, lay in the time-varying characteristics of that stimulation.

However, for the ELF-MF, we still need to consider the intensity. Indeed, the *in-situ* induced E-Field strength is intimately tethered to the stimulation's frequency. ICNIRP and IEEE-ICES suggest estimating the intensity of ELF-MF with the in-situ E-Field based on an ellipsoid model implementing Maxwell equations [44]. Nonetheless, it is acknowledged that good estimates of in-situ E-Fields can be computed with analytical spherical models [23]. Therefore, as previously done in our own work [21], [22], we estimated the in-situ E-Field using the following equation, which is derived from Maxwell's third law: $E = \frac{r}{2}\frac{\partial B}{\partial t} = \pi r f B$, where E represents the induced E-Field and r the radius of the Faraday's loop within a homogeneous alternating flux density B of frequency f. For a constant value of B at a given value of r, E will depend on the frequency f of stimulation. Following this strategy, with a flux density measured at 141.42 mT peak at the vestibular system, and frequencies ranging from 20 to 160 Hz, we obtain peak dB/dt between 18 T/s and 142 T/s. Considering a radius of 6 mm encompassing the entire vestibular system [45] (Fig. 5), peak E-Fields could be estimated between 0.054 V/m and 0.426 V/m. The entire E-Fields values for the respective frequencies can be found in Table 2.

To date, we did not find specific dosimetry work concerning the vestibular system published in the literature. However, in implanted epileptic participants, Huang *et al.* [46] found

TABLE 2. ELF-MF and AC stimulations estimated peak E-Field values across all frequencies.

		20 Hz	60 Hz	90 Hz	120 Hz	160 Hz
Estimated E-Field (V/m)	MF	0.054	0.159	0.24	0.321	0.426
	AC	0.4	0.4	0.4	0.4	0.4

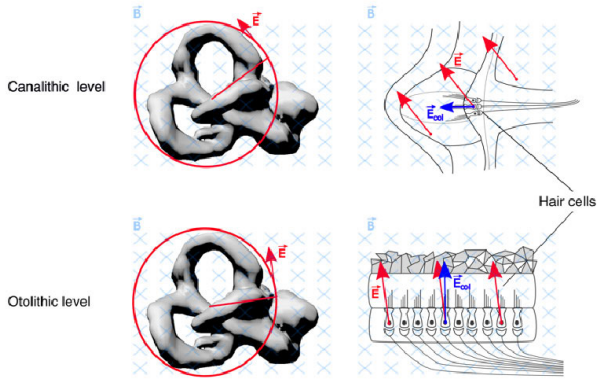

FIGURE 5. Lateral view of E-fields impacting the right vestibular system (grey structures) upon homogeneous MF increasing right-to-left ($\vec{B}$). Considering a Faraday's loop encompassing the entire vestibular system (red circles), E-fields ($\vec{E}$) are generated tangentially to $\vec{B}$ and can be applied either at the canalithic (upper panels) or otolithic level (lower panels). $\vec{E}_{col}$ (dark blue arrows) represents the component of $\vec{E}$ colinear with the hair cells showing a greater impact at the utricular level.

that 2 mA sinusoidal transcranial electrical stimulation generates 0.4 V/m at the cortical level. More interestingly, they also found such E-Fields values at deep brain structures like the anterior cingulate and the periventricular white matter [46], underlining that 2 mA at the skull, could translate to 0.4 V/m globally to the entire CNS. Furthermore, given the high conductivity values of the perilymph [47] the endolymph and the vestibular structures [48], the currents are easily drawn to the vestibular system. Hence, it is reasonable to estimate that our DC and AC 2 mA stimulations can also generate 0.4 V/m at the vestibular level (Table 2).

Considering the linear relationship between the applied current's intensity and the E-Field presented in Huang *et al* [46], 0.32 mA translates to 0.064 V/m at the vestibular system, suggesting that the ELF-MF level at 20 Hz was the only condition below the postural threshold.

Yet, we must also consider the MF orientation and more especially the relevant E-Field fraction relative to the vestibular sensors, as it would lower the impact on the structures. Indeed, phosphene literature provides evidence that fields' orientation is of paramount importance. Hirata *et al.* [23] found close to a 2.5-fold difference in magnetophosphene threshold values depending on whether fields were oriented top-down or front-back relative to the retina. It has been shown that only E-Fields colinear to the body of the neuronal cells have a maximum impact [49]. Therefore, we need to consider field orientation relative to the anatomical structures.

ELF-MF go through the anatomical structures without any hindrance and the induced E-Fields are orthogonal to the MF, constraining the currents in specific directions. Using high-resolution X-ray microtomography imaging techniques, Chacko *et al.* [50] showed important inter-variability in the orientation of the canalithic membranous labyrinths. Thus, it is hard to consider how the canalithic hair cells were oriented relative to the induced E-Fields. However, utricle and saccule are reported to be mostly planar and lying in the horizontal and vertical plane respectively [51]. It seems reasonable to consider that only a small component of the E-Field orientation was colinear with canalithic hair cells (Fig. 5, top panels), whereas most of the E-Field would be aligned with the utricular hair cells (Fig. 5, bottom panels). The saccule being mostly considered orthogonal to the utricle, the fraction of the E-Field colinear to the hair cells would be almost null.

The orientation of the ELF-MF presented in this work was intended to target the right vestibular system, which by design limited the canalithic impact and favored utricular stimulation. It is also important to emphasize that only a fraction of the peak ELF-MF generated E-fields was delivered at the vestibular sensor level, reinforcing the fact that the stimulation's strength at 20 Hz was below the postural threshold. While it is difficult to precisely assess the *in-situ* E-fields levels generated at the other frequencies for both AC and ELF-MF, the important fact is that there is no record of postural modulations for both stimulation modalities. Once again, this steers to the time-varying characteristics of these stimulations.

The vestibular information involved in postural control is integrated into the vestibular nuclei within specific vestibular-only neurons projecting to the spinal cord, the vestibulo-cerebellum, the thalamus, and the cortex [52]. This is through this integrative process that a potential stimulation frequency effect should be considered.

Although a given E-Field strength indifferently impacts canals and otoliths [29], [53], [54] the information coming from both subsystems does not seem to be equally integrated within these specific vestibular-only neurons. Indeed, as stimulation frequency increases, the weight of the otolithic input raises, whereas the weight of canalithic input decreases [55]. Hence, our high stimulation frequencies would increase otolithic weight and decrease canalithic contribution. Since postural behavioral responses due to vestibular electrical stimulations are thought to mainly result from canalithic activations (for review see [56]), this integrative weighting mechanism could be a reason for the absence of postural modulations with both our ELF-MF and AC stimulations.

As mentioned earlier, for our ELF-MF stimulations, the utricle was potentially the most modulated structure due to E-Field orientation. Interestingly, the utricle is divided mostly in half by a striola, on each side of which the vestibular hair cells are symmetrically polarized such that electrical stimulations excite one half while inhibiting the other [29], [57]. Consequently, for a homogenous E-Field stimulation over the entire utricle, little net vestibular signals

would be generated and integrated, possibly leading to lower utricular effects [58].

Finally, it is hypothesized that for the biomechanical system to work efficiently, only frequencies required to control task-specific muscle physiology are used [59]. Data show that leg muscles only respond to frequencies below 20 Hz, indicating that vestibular inputs above this frequency could be biomechanically low passed filtered at the muscle level [59]–[61]. In the context of our study, this would imply that the biomechanical low pass filtering could have lowered the impact of vestibular stimulations on postural sway.

In summary within the postural control context, the use of high frequencies in both ELF-MF and AC stimulations limited sway responses by i) promoting otolithic activation over the canalithic system, dampening if not inhibiting the emergence of a net oriented head acceleration signal due to cross-striolar inhibition mechanisms and ii) being low pass filtered by the neuromuscular system.

The absence of postural response would therefore not reflect an absence of effect on the vestibular system but rather an absence of functional translation to postural control outcomes.

In that regard, we are proposing to discuss the sensitivity of the vestibular system through a look at pathways mediating quick three neuron arc reflexes such as the vestibulo-ocular and vestibulospinal reflexes [20]. In this perspective, mean DC stimulations as low as 0.1 mA have been reported to trigger reflexive eye movement in healthy participants [25]. Once again, considering the linear relationship between the current's intensity and the E-Field [46], 0.1 mA translates to 0.02 V/m at the vestibular system.

In this case, all ELF-MF generated E-Fields (Table 2) are now all above the reflexive vestibular threshold. Furthermore, Forbes *et al.* [62], recording human neck motoneuron activity, showed that 300 Hz AC stimulations modulate canalithic activity. This not only highlights their sensitivity to such high frequencies but also that their activation translates into myogenic reflexive activity through less integrated vestibulospinal pathways [62]. Moreover, otolithic hair cells phase lock with frequencies above 2000 Hz [63]. Therefore, from a vestibulo-reflexive pathway standpoint, our stimulations' intensities were high enough to modulate vestibular activity and our stimulations' frequencies were not a limiting parameter.

Altogether, our stimulations' frequencies stand out as the main limiting postural factor in our study. Taken together with our previous studies [21], [22], this work suggests that powerline-frequency vestibular specific ELF-MF stimulations cannot have functional effects on postural outcomes. Yet, vestibular reflexes are sensitives to both higher stimulation frequencies and lower stimulation intensities. Consequently, further protocols should implement eye-tracking methods [64], [65] to study the ELF-MF impact on the vestibulo-ocular reflex. Furthermore, given the field orientation, and the implication of the otolithic activity at higher frequencies, the focus of further investigations could also point to specific otolithic tests such as ocular and cervical vestibular evoked myogenic potentials, both sensitive to E-fields [66], [67].

V. CONCLUSION

We did not find postural modulations with our lateral vestibular ELF-MF stimulations, which is consistent with the finding from our strong entire head top-down ELF-MF stimulation study [Bouisset *et al.*, 2020]. Based on Lövsund *et al.* [Lövsund *et al.*, 1979], and Saunders and Jefferys [Saunders and Jefferys, 2007], the synaptic threshold is 0.075 V/m peak in the ELF-MF range [3]. This threshold is reported to trigger synaptic modulations, potentially leading to adverse effects in the brain [3]. However, using *in-situ* E-Fields up to 0.426 V/m at the vestibular level, this study also subjected the CNS structures under the coils to fields well above this synaptic threshold and no sensorimotor effects were found (in line with our previous results [22]). Therefore, these results challenge the idea assuming that neurophysiological similarities between sensory systems would trigger equivalent responses and the possibility to generalize local effects to other parts of the CNS. It also raises the questions of the functional scale at which the E-fields are to be estimated as well as fields' orientations relative to the structure of concern. Finally, our study highlights the importance of understanding the mechanisms of neuronal integration. These are critical questions to be addressed to fill the actual knowledge gaps [68] which will surely be useful in the future writing of both ICNIRP guidelines and IEEE-ICES standards.

ETHICAL STANDARD

The Health Sciences Research Ethics Board (#106122) at Western University approved this study, therefore, performed following the ethical standards laid down in the 1964 Declaration of Helsinki.

ACKNOWLEDGMENT

The authors would like to thank L. Keenliside for the development and construction of the exposure head coils as well as L. Hug and Dr. I. Laakso for their contribution in the graphical abstract and Figure 1. The funders had no role in study design, data collection and analysis, decision to publish, study approvals, or preparation of the manuscript.

REFERENCES

[1] *IEEE Standard for Safety Levels with Respect to Human Exposure to Electromagnetic Fields, 0–3 kHz*, Standard C95.6, Mar. 2002.

[2] ICNIRP, "Guidelines for limiting exposure to time-varying electric and magnetic fields (1 Hz to 100 kHz)," *Health Phys.*, vol. 99, no. 6, pp. 818–836, 2010.

[3] *IEEE Standard for Safety Levels With Respect to Human Exposure to Electric, Magnetic, and Electromagnetic Fields, 0 Hz to 300 GHz*, Standard 2005, Apr. 2019.

[4] A. D'Arsonval, "Dispositifs pour la mesure des courants alternatifs de toutes fréquences," *Compt. Rend. Soc. Biol.*, vol. 3, no. 2, pp. 450–451, May 1896.

[5] D. Attwell, "Interaction of low frequency electric fields with the nervous system: The retina as a model system," *Radiat. Protection Dosimetry*, vol. 106, no. 4, pp. 341–348, 2003.

[6] M. Juusola, A. S. French, R. O. Uusitalo, and M. Weckström, "Information processing by graded-potential transmission through tonically active synapses," *Trends Neurosci.*, vol. 19, no. 7, pp. 292–297, Jul. 1996.
[7] L. Lagnado, A. Gomis, and C. Job, "In the Synaptic Terminal of Retinal Bipolar Cells," *Cell*, vol. 17, pp. 957–967, 1996.
[8] R. A. Eatock and J. E. Songer, "Vestibular hair cells and afferents: Two channels for head motion signals," vol. 34, no. 1, 2011.
[9] K. K. Ghosh, S. Haverkamp, and H. Wässle, "Glutamate receptors in the rod pathway of the mammalian retina," *J. Neurosci.*, vol. 21, no. 21, pp. 8636–8647, Nov. 2001.
[10] S. G. Sadeghi, S. J. Pyott, Z. Yu, and E. Glowatzki, "Glutamatergic signaling at the vestibular hair cell calyx synapse," *J. Neurosci.*, vol. 34, no. 44, pp. 14536–14550, Oct. 2014.
[11] J. M. Goldberg, *The Vestibular System: A Sixth Sense*. New York, NY, USA: Oxford Univ. Press, 2012.
[12] R. A. Eatock, R. R. Fay, and A. N. Popper, *Vertebrate Hair Cells*. New York, NY, USA: Springer, 2006.
[13] I. D. Evans, S. Palmisano, S. P. Loughran, A. Legros, and R. J. Croft, "Frequency-dependent and montage-based differences in phosphene perception thresholds via transcranial alternating current stimulation," *Bioelectromagnetics*, vol. 40, no. 6, pp. 365–374, 2019.
[14] J. Dlugaiczyk, K. D. Gensberger, and H. Straka, "Galvanic vestibular stimulation: From basic concepts to clinical applications," *J. Neurophysiol.*, vol. 121, no. 6, pp. 2237–2255, Jun. 2019.
[15] K. D. Gensberger, A.-K. Kaufmann, H. Dietrich, F. Branoner, R. Banchi, B. P. Chagnaud, and H. Straka, "Galvanic vestibular stimulation: Cellular substrates and response patterns of neurons in the vestibulo-ocular network," *J. Neurosci.*, vol. 36, no. 35, pp. 9097–9110, Aug. 2016.
[16] S. T. Aw, M. J. Todd, G. E. Aw, K. P. Weber, and G. M. Halmagyi, "Gentamicin vestibulotoxicity impairs human electrically evoked vestibulo-ocular reflex," *Neurology*, vol. 71, no. 22, pp. 1776–1782, Nov. 2008.
[17] H. P. Zenner, G. Reuter, S. Hong, U. Zimmermann, and A. H. Gitter, "Electrically evoked motile responses of mammalian type I vestibular hair cells," *J. Vestibular Res.*, vol. 2, no. 3, pp. 181–191, 1992.
[18] C. H. Norris, A. J. Miller, P. Perin, J. C. Holt, and P. S. Guth, "Mechanisms and effects of transepithelial polarization in the isolated semicircular canal," *Hearing Res.*, vol. 123, nos. 1–2, pp. 31–40, Sep. 1998.
[19] E. E. Hansson, A. Beckman, and A. Håkansson, "Effect of vision, proprioception, and the position of the vestibular organ on postural sway," *Acta Oto-Laryngologica*, vol. 130, no. 12, pp. 1358–1363, Dec. 2010.
[20] K. E. Cullen, "The vestibular system: Multimodal integration and encoding of self-motion for motor control," *Trends Neurosci.*, vol. 35, no. 3, pp. 185–196, Mar. 2012.
[21] S. Villard, A. Allen, N. Bouisset, M. Corbacio, A. Thomas, M. Guerraz, and A. Legros, "Impact of extremely low-frequency magnetic fields on human postural control," *Exp. Brain Res.*, vol. 237, no. 3, pp. 611–623, Mar. 2019.
[22] N. Bouisset, S. Villard, and A. Legros, "Human postural control under high levels of extremely low frequency magnetic fields," *IEEE-Access*, vol. 8, pp. 101377–101385, 2020.
[23] A. Hirata, Y. Takano, O. Fujiwara, T. Dovan, and R. Kavet, "An electric field induced in the retina and brain at threshold magnetic flux density causing magnetophosphenes," *Phys. Med. Biol.*, vol. 56, no. 13, pp. 4091–4101, Jul. 2011.
[24] S. T. Aw, M. J. Todd, and G. M. Halmagyi, "Latency and initiation of the human vestibuloocular reflex to pulsed galvanic stimulation," *J. Neurophysiol.*, vol. 96, no. 2, pp. 925–930, 2006.
[25] A. Séverac Cauquil, M. Faldon, K. Popov, B. L. Day, and A. M. Bronstein, "Short-latency eye movements evoked by near-threshold galvanic vestibular stimulation," *Exp. Brain Res.*, vol. 148, no. 3, pp. 414–418, Feb. 2003.
[26] I. Laakso, S. Kännälä, and K. Jokela, "Computational dosimetry of induced electric fields during realistic movements in the vicinity of a 3 T MRI scanner," *Phys. Med. Biol.*, vol. 58, no. 8, pp. 2625–2640, 2013.
[27] P. Lövsund, P. Å. Öberg, S. E. G. Nilsson, and T. Reuter, "Magnetophosphenes: A quantitative analysis of thresholds," *Med. Biol. Eng. Comput.*, vol. 18, no. 3, pp. 326–334, May 1980.
[28] J.-F. Yu, K.-C. Lee, R.-H. Wang, Y.-S. Chen, C.-C. Fan, Y.-C. Peng, T.-H. Tu, C.-I. Chen, and K.-Y. Lin, "Anthropometry of external auditory canal by non-contactable measurement," *Appl. Ergonom.*, vol. 50, pp. 50–55, Sep. 2015.
[29] R. C. Fitzpatrick and B. L. Day, "Probing the human vestibular system with galvanic stimulation," *J. Appl. Physiol.*, vol. 96, no. 6, pp. 2301–2316, 2004.
[30] J.-P. Bresciani, "Vestibular signals contribute to the online control of goal-directed arm movements," *Current Psychol. Cogn.*, vol. 21, pp. 263–280, 2002.
[31] T. D. Wilson, J. M. Serrador, and J. K. Shoemaker, "Head position modifies cerebrovascular response to orthostatic stress," *Brain Res.*, vol. 961, no. 2, pp. 261–268, Jan. 2003.
[32] C. K. Rhea, A. W. Kiefer, F. J. Haran, S. M. Glass, and W. H. Warren, "A new measure of the CoP trajectory in postural sway: Dynamics of heading change," *Med. Eng. Phys.*, vol. 36, no. 11, pp. 1473–1479, Nov. 2014.
[33] L. Chiari, L. Rocchi, and A. Cappello, "Stabilometric parameters are affected by anthropometry and foot placement," *Clin. Biomech.*, vol. 17, nos. 9–10, pp. 666–677, Nov. 2002.
[34] L. Rocchi, L. Chiari, and A. Cappello, "Feature selection of stabilometric parameters based on principal component analysis," *Med. Biol. Eng. Comput.*, vol. 42, no. 1, pp. 71–79, 2004.
[35] A. Séverac Cauquil, P. Martinez, M. Ouaknine, and M.-F. Tardy-Gervet, "Orientation of the body response to galvanic stimulation as a function of the inter-vestibular imbalance," *Exp. Brain Res.*, vol. 133, no. 4, pp. 501–505, Aug. 2000.
[36] L. F. Oliveira, D. M. Simpson, and J. Nadal, "Calculation of area of stabilometric signals using principal component analysis," *Physiol. Meas.*, vol. 17, no. 4, pp. 305–312, Nov. 1996.
[37] J. E. Fitzgerald, A. Murray, C. Elliott, and J. P. Birchall, "Comparison of body sway analysis techniques: Assessment with subjects standing on a stable surface," *Acta Oto-Laryngol.*, vol. 114, no. 2, pp. 115–119, Jan. 1994.
[38] G. Nagymáté, Z. Orlovits, and R. M. Kiss, "Reliability analysis of a sensitive and independent stabilometry parameter set," *PLoS ONE*, vol. 13, no. 4, 2018, Art. no. e0195995.
[39] I. Laakso, S. Tanaka, S. Koyama, V. De Santis, and A. Hirata, "Inter-subject variability in electric fields of motor cortical tDCS," *Brain Stimulation*, vol. 8, no. 5, pp. 906–913, Sep. 2015.
[40] R. C. Team, *R: A Language and Environment for Statistical Computing*, R Found. for Stat. Comput. Vienna Austria, Vienna, NSW, Austria, 2016.
[41] P. Berens, "CircStat: A MATLAB toolbox for circular statistics," *J. Stat. Softw.*, vol. 31, no. 10, pp. 1–21, 2009.
[42] H. Petersen, M. Magnusson, P. A. Fransson, and R. Johansson, "Vestibular stimulation perturbs human stance also at higher frequencies," *Acta Oto-Laryngologica*, vol. 520, no. 2, pp. 443–446, 1995.
[43] Y. Yang, F. Pu, X. Lv, S. Li, J. Li, D. Li, M. Li, and Y. Fan, "Comparison of postural responses to galvanic vestibular stimulation between pilots and the general populace," *BioMed Res. Int.*, vol. 2015, pp. 1–6, 2015.
[44] J. P. Reilly, "Magnetic field excitation of peripheral nerves and the heart: A comparison of thresholds," *Med. Biol. Eng. Comput.*, vol. 29, pp. 571–579, Nov. 1991.
[45] R. W. Baloh, V. Honrubia, and A. Kerber, *Clinical Neurophysiology of the Vestibular System*, 4th ed. New York, NY, USA: Oxford Univ. Pres, 2011.
[46] Y. Huang, A. A. Liu, B. Lafon, D. Friedman, M. Dayan, X. Wang, M. Bikson, W. K. Doyle, O. Devinsky, and L. C. Parra, "Measurements and models of electric fields in the *in vivo* human brain during transcranial electric stimulation," *eLife*, vol. 6, pp. 1–27, Feb. 2017.
[47] S. B. Baumann, D. R. Wozny, S. K. Kelly, and F. M. Meno, "The electrical conductivity of human cerebrospinal fluid at body temperature," *IEEE Trans. Biomed. Eng.*, vol. 44, no. 3, pp. 220–223, Mar. 1997.
[48] A. Antunes, P. M. Glover, Y. Li, O. S. Mian, and B. L. Day, "Magnetic field effects on the vestibular system: Calculation of the pressure on the cupula due to ionic current-induced Lorentz force," *Phys. Med. Biol.*, vol. 57, no. 14, pp. 4477–4487, 2012.
[49] T. Radman, R. L. Ramos, J. C. Brumberg, and M. Bikson, "Role of cortical cell type and morphology in subthreshold and suprathreshold uniform electric field stimulation *in vitro*," *Brain Stimulation*, vol. 2, no. 4, pp. 215–228, 2009.
[50] L. Johnson Chacko, D. T. Schmidbauer, S. Handschuh, A. Reka, K. D. Fritscher, P. Raudaschl, R. Saba, M. Handler, P. P. Schier, D. Baumgarten, N. Fischer, E. J. Pechriggl, E. Brenner, R. Hoermann, R. Glueckert, and A. Schrott-Fischer, "Analysis of vestibular labyrinthine geometry and variation in the human temporal bone," *Frontiers Neurosci.*, vol. 12, pp. 1–13, Feb. 2018.
[51] S. Khan and R. Chang, "Anatomy of the vestibular system: A review," *NeuroRehabilitation*, vol. 32, no. 3, pp. 437–443, 2013.
[52] K. E. Cullen, "Vestibular processing during natural self-motion: Implications for perception and action," *Nature Rev. Neurosci.*, vol. 20, no. 6, pp. 346–363, Jun. 2019.

[53] I. S. Curthoys and H. G. MacDougall, "What galvanic vestibular stimulation actually activates," *Frontiers Neurol.*, vol. 3, pp. 1–5, 2012.
[54] B. L. Day, E. Ramsay, M. S. Welgampola, and R. C. Fitzpatrick, "The human semicircular canal model of galvanic vestibular stimulation," *Exp. Brain Res.*, vol. 210, nos. 3–4, pp. 561–568, May 2011.
[55] J. Carriot, M. Jamali, J. X. Brooks, and K. E. Cullen, "Integration of canal and otolith inputs by central vestibular neurons is subadditive for both active and passive self-motion: Implication for perception," *J. Neurosci.*, vol. 35, no. 8, pp. 3555–3565, Feb. 2015.
[56] R. F. Reynolds and C. J. Osler, "Galvanic vestibular stimulation produces sensations of rotation consistent with activation of semicircular canal afferents," *Frontiers Neurol.*, vol. 3, pp. 1–2, 2012.
[57] A. B. Tascioglu, "Brief Review of Vestibular Anatomy and Its Higher Order Projections," *Neuroanatomy*, vol. 4, no. 4, pp. 24–27, 2005.
[58] Y. Uchino, "Cross-striolar and commissural inhibition in the otolith system," *Ann. Otol. Rhinol. Laryngol.*, vol. 871, no. 1, pp. 162–172, 1999.
[59] P. A. Forbes, "Frequency response of vestibular reflexes in neck, back, and lower limb muscles," *J. Neurophysiol.*, vol. 110, pp. 1869–1881, Jul. 2013.
[60] C. J. Dakin, B. L. Luu, K. van den Doel, J. T. Inglis, and J.-S. Blouin, "Frequency-specific modulation of vestibular-evoked sway responses in humans," *J. Neurophysiol.*, vol. 103, no. 2, pp. 1048–1056, 2010.
[61] P. A. Forbes, G. P. Siegmund, A. C. Schouten, and J.-S. Blouin, "Task, muscle and frequency dependent vestibular control of posture," *Frontiers Integrative Neurosci.*, vol. 8, p. 94, Jan. 2014.
[62] P. A. Forbes, A. Kwan, B. G. Rasman, D. E. Mitchell, K. E. Cullen, and J.-S. Blouin, "Neural mechanisms underlying high-frequency vestibulocollic reflexes in humans and monkeys," *J. Neurosci.*, vol. 40, no. 9, pp. 1874–1887, Feb. 2020.
[63] I. S. Curthoys, "The new vestibular stimuli: Sound and vibration—anatomical, physiological and clinical evidence," *Exp. Brain Res.*, vol. 235, no. 4, pp. 957–972, 2017.
[64] S. W. Mackenzie and R. F. Reynolds, "Ocular torsion responses to sinusoidal electrical vestibular stimulation," *J. Neurosci. Methods*, vol. 294, pp. 116–121, Jan. 2018.
[65] J. Otero-Millan, D. C. Roberts, A. Lasker, D. S. Zee, and A. Kheradmand, "Knowing what the brain is seeing in three dimensions: A novel, noninvasive, sensitive, accurate, and low-noise technique for measuring ocular torsion," *J. Vis.*, vol. 15, no. 14, pp. 1–15, 2015.
[66] H. Monobe and T. Murofushi, "Vestibular testing by electrical stimulation in patients with unilateral vestibular deafferentation: Galvanic evoked myogenic responses testing versus galvanic body sway testing," *Clin. Neurophysiol.*, vol. 115, no. 4, pp. 807–811, Apr. 2004.
[67] S. M. Rosengren, P. Jombik, G. M. Halmagyi, and J. G. Colebatch, "Galvanic ocular vestibular evoked myogenic potentials provide new insight into vestibulo-ocular reflexes and unilateral vestibular loss," *Clin. Neurophysiol.*, vol. 120, no. 3, pp. 569–580, Mar. 2009.
[68] "Gaps in knowledge relevant to the 'Guidelines for limiting exposure to time-varying electric and magnetic fields (1 Hz–100 kHz),'" *Health Phys.*, vol. 118, no. 5, pp. 533–542, 2020.

NICOLAS BOUISSET received the B.A.Sc. degree in sport science from the Université de la Réunion, France, in 1999, and the physiotherapy degree from the Université Catholique de Louvain-la-Neuve, Belgium, in 2003. After working clinically for over a decade, he received the M.S. degree in human movement science from the Université de Montpellier, France, in 2015. He is currently pursuing the Ph.D. degree with the Human Threshold Research Group, Lawson Health Research Institute, The University of Western Ontario, Canada. His current research interests include vestibular function modulations using both Galvanic Vestibular stimulations and extremely low frequency magnetic fields.

SÉBASTIEN VILLARD received the Ph.D. degree in human movement science from the University of Montpellier, France, in 2005. He is currently an Assistant Professor with the Department of Health Sciences, The University of Western Ontario. He is also working as a Scientist with the Bioelectromagnetics and Human Threshold Research Group, Imaging Program, Lawson Research Institute. After focusing on neurological and biomechanical interactions between respiration and locomotion in humans, he moved to Minneapolis, MN to study postural instability in visually induced motion sickness. His role was to design new experiments but also new methodologies to assess postural instability generated by the interaction between human and virtual environments such as simulator or videogames. After about three years of collaboration at the Human Factors Research Laboratory, University of Minnesota, he accepted a Research Engineer position for a European Integrated Project in the SKILLS consortium in France. His role was to drive the fundamental research regarding the optimization of energy consumption management in the development of a rowing prototype for high-level athletes. Since 2015, he has been combining his former experiences to study the impact of low-frequency magnetic fields on postural control and more specifically on the vestibular system. His research interests include the crossroad between neuroscience, physiology, biomechanics and psychology, and how humans interact with new technologies.

ALEXANDRE LEGROS was born in Versailles, France, in 1976. He received the Ph.D. degree in human movement sciences, in 2004, and completed a First Postdoctoral Fellowship on electrical deep brain stimulation (DBS) in dystonic syndromes at the Neurosurgery, Guy de Chauliac Hospital, Montpellier, France. He also completed a Second Postdoctoral Fellowship at the Bioelectromagnetics Group, Lawson, from 2005 to 2007, where he was a Scientist, in September 2007. He is currently the Director of the Bioelectromagnetics and Human Threshold Research Group, Lawson Health Research Institute, London, ON, Canada, and an Associate Professor with the Departments of Medical Biophysics, Medical Imaging, and Kinesiology, Western University, Canada. He is also an Associate Scientist with the EuroStim/EuroMov, University of Montpellier, France, where he is duplicating his Canadian Laboratory to develop new collaborative research projects involving human responses to high levels of ELF-MF. He has expertise in the fields of neurosciences, kinesiology, and biophysics applied to the study of neurostimulation and neuromodulation. His research interests include the effects of specific electric and magnetic stimuli (DBS, transcranial magnetic stimulation, and time-varying magnetic fields) on human brain processing, motor control, and cognitive functions. He was a Board Member, from 2013 to 2015. He was the Chair of the Local Organising Committee for BioEM2019. He is also a Secretary of the Board of Directors of the Bioelectromagnetics Society (BEMS). He is a Technical Program Committee Co-Chair for BioEM2020. He is the Canadian Chair for URSI commission K and the Chair of the Non-Ionizing Radiations Task Group, IRPA (International Radio Protection Association). He is also the Co-Chairing a working group within the IEEE-ICES TC95 Subcommittee six and Chairing a task force on low frequencies recommendations.