Measuring hearing instrument sound modification using integrated ear-EEG

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We integrated ear electrodes into a live hearing system and evaluated the feasibility of recording electroencephalography (EEG) features with this setup using an auditory discrimination experiment. The long-term goal is to construct a closed-loop brain-computer-interface that is integrated in a mobile research hearing system. Here, the EEG setup consists of 3 electrodes embedded in the earmoulds of an experimental hearing system and 10 flex-printed electrodes positioned around each ear, all connected to a wireless EEG amplifier. Four consecutive identical broadband stimuli were played in headphones while the spectral profile of sounds arriving at the eardrum was altered by switching the signal processing setting of the hearing system. Such switches were made between presentation of the third and the fourth stimulus, in half of all epochs. Seventeen normal hearing subjects participated and were instructed to indicate whether the last stimulus sounded different. The behavioural data verified clear audibility of the switches. The EEG analysis revealed differences between switch and no-switch trials in the N1 and P3 latency range. Importantly, changes in the spectral content of the noise floor of the hearing device were already sufficient to elicit these responses. These results confirm that stimulus-related brain signals acquired from ear-EEG during real-time audio processing can be successfully derived.

INTRODUCTION

Neuro-control of hearing devices has the potential to improve hearing support by taking the electroencephalography (EEG) derived listening intent of a person into account for better control of algorithms and hearing aid setting. For instance, it has been demonstrated that the direction of attention in a competing talker situation can be decoded from EEG signals within a reasonable duration (O’Sullivan *et al.*, 2015; Mirkovic *et al.*, 2016). Such information about the direction of attention may be utilized to enhance the speech signal of the desired speaker by, for example, steering a beamformer to the position of the attended speaker (Doclo *et al.*, 2015).

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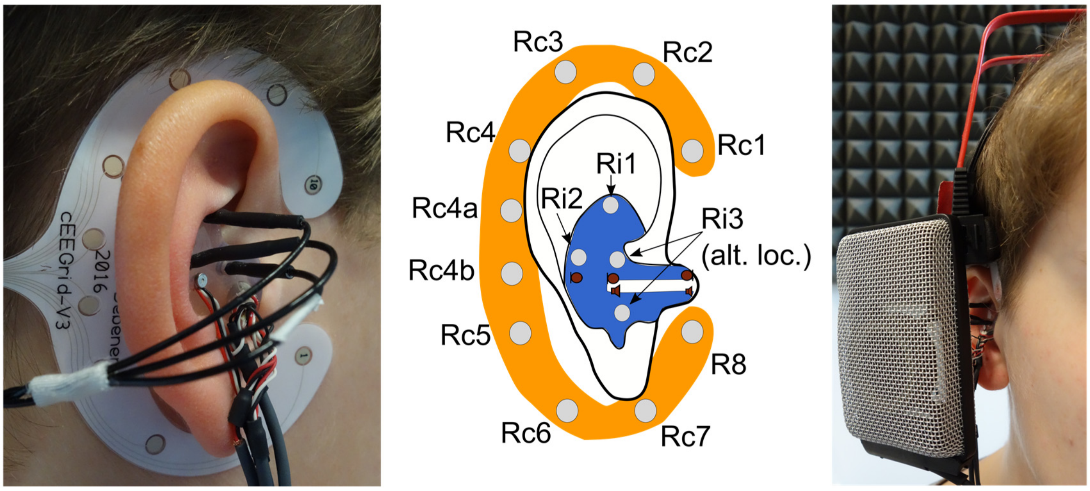

Fig. 1: Left: Photograph of the setup in the ear of a participant. In the concha, the earmould containing the hearing system and 3 electrodes (black sticks) are placed. The cEEGrid (www.ceegrid.com) is glued around the ear. Centre: Schematic view of the layout in the ear. Grey circles indicate electrodes with their according nomenclature; Positions of electro-acoustic transducers are marked by according symbols. The shaded area marks the part of the earmould which is inserted into the ear canal. Right: Participant wearing super-aural headphones (*AKG K-1000*) over the in-ear setup.

For an integration of EEG with hearing devices, however, it is necessary that the EEG signal can be acquired in a socially acceptable manner, with as little inconvenience for the hearing aid user as possible (Bleichner and Debener, 2017). In order to achieve such a transparent EEG acquisition, several ear-EEG approaches have been presented that allow to record EEG reliably in and around the ears (Looney *et al.*, 2011; Bleichner *et al.*, 2015; Bleichner and Debener, 2017). It has been shown that ear-EEG can be used to capture a wide variety of auditory perception-related processes: auditory steady state responses (Kidmose *et al.*, 2012), auditory onset responses, mismatch negativity as well as alpha attenuation (Mikkelsen *et al.*, 2015). Moreover, ear-EEG can also be used to detect the direction of auditory attention (Mirkovic *et al.*, 2016; Bleichner *et al.*, 2016).

The next step towards closing the loop between the hearing device and EEG is the integration of ear-centred EEG hardware into a live hearing system. We present a feasibility study of this combination, where electrodes are placed in and around the ear of a participant, integrated with an experimental hearing system (Denk *et al.*, 2017). To our best knowledge, this setup is the closest to a functional hearing device with integrated EEG electrodes that has been reported. Our goal here was to determine whether an audible switch in the hearing device processing is reflected in auditory evoked potentials (AEP) measured with ear-EEG.

METHODS

Setup

The participants were equipped with a prototype hearing system as presented by (Denk *et al.*, 2017), consisting of an individual silicone earmould that contains a set of electro-acoustic transducers shown in Fig. 1. External sound is captured with a microphone located in the concha, processed, and played back via an included receiver. Real-time processing is performed on a laptop running the Master Hearing Aid (MHA) platform (Grimm *et al.*, 2006), which is connected to the transducers through a *Multiface II* soundcard (RME, Haimhausen, Germany) with an input-output delay of 7.8 ms. By means of an individual in-situ calibration routine, the processing chain (here a finite impulse response filter) is adapted in a way that the superposition of electro-acoustically generated sound and sound leaking through the vented earpiece approximates the pressure at the eardrum that is observed with an open ear. Hence, acoustically transparent reproduction of the acoustic environment is provided while having the possibility to modify the presented sound in a desired manner by changing the output filter F.

EEG was acquired with ear-centred electrodes. In each ear three cylindrical electrodes (2 x 4 mm, Ag/AgCl, EasyCap, Herrsching, Germany; cf. Bleichner *et al.*, 2015) were distributed in the cavum concha by insertion into bores in the earmould. Additionally, ten printed Ag/AgCl electrodes were placed around the ear using the commercially available cEEGrid system (www.ceegrid.com), which is a flex-printed C-shaped electrode array placed around the ear (Debener *et al.*, 2015). After skin preparation with an abrasive gel and alcohol, a small amount of electrolyte gel (Abralyt HiCl, Easycap GmbH, Germany) was applied to the electrodes and the cEEGrids were placed with a double-sided adhesive tape around the ear. The cylindrical electrodes were inserted into the earmould after a drop of electrolyte gel was administered into the bores. The whole setup in the ear of a subject, as well as the schematic layout, is shown in Fig. 1. All electrodes were connected to a portable wireless 24-channel EEG amplifier attached to the subjects' heads (SMARTING, mBrainTrain, Belgrade, Serbia) and recording EEG signals with a sampling rate of 500 Hz and 24-bit resolution. A Bluetooth connection enabled wireless EEG recording on a separate computer. Although the system is a laboratory-state prototype, the suggested electrode layout is readily applicable in a real hearing system, or a fully mobile prototype.

The participants performed all tasks autonomously using graphical interfaces shown on a laptop that also controlled the hearing device while participants were seated in a sound-proof booth. Auditory stimulation and experimental control was implemented in MATLAB on the same laptop, which was also used to send EEG triggers synchronously to audio stimulation via Lab Streaming Layer (LSL; Kothe, 2015). On an additional computer located outside the booth, the Bluetooth EEG signal was recorded together with the trigger stream and a mirror of the acoustic stimuli.

Stimuli were presented via super-aural headphones (K-1000, AKG, Vienna, Austria), which are shown in Fig. 1. The special design assures that neither the electrodes nor the hearing device was touched by the headphone. Whereas EEG was recorded at both ears, the stimuli were presented monaurally to the right ear. Consequently, only the right ear was equipped with a hearing device and the left ear was fully occluded.

Paradigm and stimuli

Two different listening conditions were implemented by variable operation modes of the hearing device, while in all compared trials the identical stimulus waveform was played on the headphones. In one adjustment, the output filter of the hearing device was adjusted by individual calibration prior to the main experiment (filter F1). In the other condition, the output filter resulting from equivalent calibration of the system on a dummy head was used (F2), which results in a notable difference in the spectral profile arriving at the eardrum. Alternative cues that may arise from differences in loudness were compensated through an additional broadband gain applied to the dummy head filter, which was adjusted by means of an adaptive 1-up 1-down procedure prior to the main experiment.

Three types of stimuli were included: white noise ("*Noise*"), a logatome spoken by a female voice referred to as "*Speech*" (*Sass*, from the OLLO corpus; Meyer *et al.*, 2010), and the superposition of both with an SNR of 5 dB ("*Speech-In-Noise*"). To all stimuli, bandpass filtering between 0.1 and 12 kHz was applied.

Four identical stimuli were presented sequentially, in 50% of the trials the last stimulus was presented with a different filter setting (deviant condition, e.g., F1 F1 F1 F2) in 50% with the same filter (identical condition, e.g., F2 F2 F2 F2). The onset of the *n*-th stimulus is referred to as T*n*. Each stimulus was 500-ms long, separated by 300-ms breaks. To assure the participants' attention, they were asked to indicate whether the last stimulus was perceived as identical to the three prior sounds or not by pressing buttons on the laptop (y/n) guided by a graphical user interface. The response time window was limited to 1 second to get a spontaneous response from the participants, followed by a pause lasting randomly between 2.5 and 3.5 seconds.

The waveform of the *Speech-In-Noise* stimulus is shown together with the AEPs in Fig. 3 (Results section). Since a real-time hearing device was used, a noise floor was perceivable in silence, originating mainly from the microphone. Aiming to avoid sudden audible modification in noise timbre when the output filter was switched, the hearing device output was briefly deactivated while switching the output filter (or not), 120 ms after presentation of every trial (20-ms pause, with 10-ms ramps).

For each of the three stimuli, 16 deviant epochs in both possible orders (F1F2, F2F1), and the same number of non-deviant epochs in either filter setting (F1F1, F2F2) were presented. Hence, 192 sequences of stimuli were presented in randomized order, subdivided into four blocks of equal case distribution. The experiment included further conditions with a comparable number of trials, which are not considered here. Seventeen participants without any self-reported history of hearing disorder participated in the study. Including calibration of the hearing device and loudness matching of the presentation conditions prior to the main experiment, the experiment lasted about 90 minutes, separated by four small breaks between the experimental blocks.

EEG analysis

The offline analysis was performed with EEGlab (Delorme and Makeig, 2004) and MATLAB (Mathworks, Natick, MA). The data from each block was filtered between 0.1 and 12 Hz with consecutive high-pass and low-pass filters. Epochs were extracted for the entire trial (−1000 ms to 4000 ms relative to T1) as well as to the onset of the device before the last stimulus (−500 ms to 1000 ms). Epochs dominated by artefacts were identified using the probability criteria implemented in EEGLAB (standard deviation: 2) and rejected from further analysis. The grand average AEP over all trials and all participants was computed.

RESULTS AND DISCUSSION

Behavioural results

The behavioural discrimination results are shown in Fig. 2. Generally, the participants were able to discriminate well between identical and deviant trials. On average, the correct response was given in 90% and 93% of all epochs, respectively. Thus, the listening results verify the desired audibility of the difference between the two filter settings.

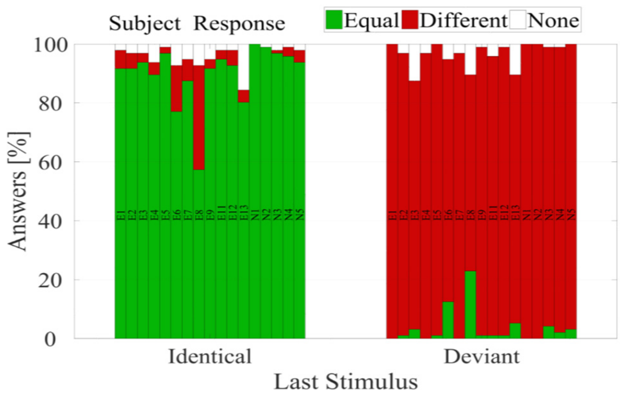

Fig. 2: Behavioural results, pooled over stimuli. Subjects E6, E8 and E13 were excluded from further analysis due to results indicating poor attention.

Some participants performed very clearly below average, which may be attributed to poor vigilance or task compliance. To avoid compromising the physiological results, data from participants were discarded if the following criteria were not fulfilled:

1. Identical stimuli sequences indicated as “identical” in more than 80 % of all epochs;
2. Deviant stimuli sequences marked as “identical” in less than 20 % of all epochs;
3. Answer given in more than 90% of all trials.

Consequently, the data from subjects E6, E8 and E13 were excluded from further analysis.

Auditory evoked potentials

Extensive pilot studies, including stimulation over distant loudspeakers with the hearing device deactivated, verified that the signals obtained in the electrodes originate from neural activity and not due to crosstalk from the audio transducers or connections.

The grand average AEP is shown in Fig. 3 together with the recording of the *Speech-in-Noise* stimulus. For the latter, the sound pressure measured at the eardrum of a dummy head is shown together with the output voltage of the hearing device’s receiver. The shown AEPs were measured for electrode Rc3 referenced to Rc6 (see Fig. 1). Clearly apparent is the negative deflection (N1) around 150 ms after stimulus onset (for T1). Note that the sound onset of the *Speech* is later (~200 ms) than in the noise conditions and that the N1 is shifted accordingly. Also apparent is the amplitude reduction of the N1 for T2 and T3 relative to T1 for all conditions. Likewise, all stimulus types evoked a negative deflection prior to stimulus onset with a latency that matches the onset of the idle noise when the device was first switched on. When comparing the identical and deviant last tones (T4) we observed a condition difference with a larger N1 amplitude followed by a larger P3 amplitude (at around 2700 ms) for the deviant stimuli. This difference was most pronounced for *Speech*, but was also observed for the other stimuli. Importantly, the peak latency of the N1 did not fit to the onsets of the stimuli, but matched the last switch (Off/On mark) of the hearing device filter. The explanation is most probably that the subjects perceived the difference in hearing device filter setting already in the idle noise.

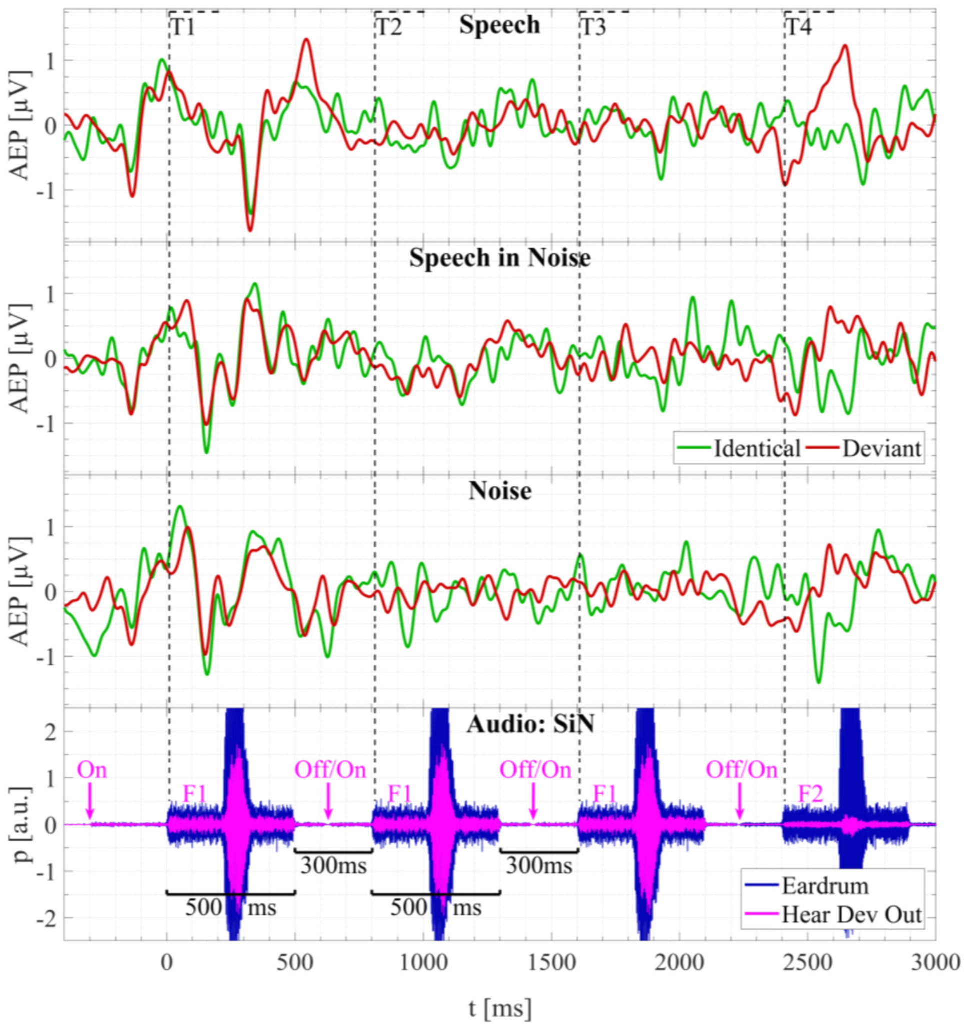

Fig. 1: AEPs averaged over subjects for all stimuli individually, and the recording of the *Speech-In-Noise* stimulus made in a dummy head.

Figure 4 shows the AEP amplitudes relative to the idle noise onset prior to T4 averaged over the identical and deviant stimulus types, respectively. A clear difference was observed in the average AEP, where a N1 and P3 was identified for the deviant, but not in the identical condition. The N1 amplitudes were averaged for the time window between 142 and 182 ms, and the P3 in the time window between 270 and 470 ms. A significant difference between conditions was evident for N1 ($p = 0.0046$) and P3 amplitudes ($p = 0.0078$).

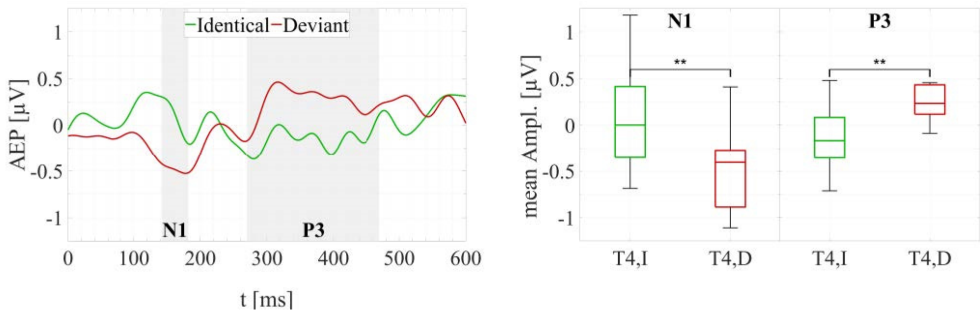

Fig. 4: Left: AEP on device onset prior to T4 (=0ms), averaged over all identical and deviant stimuli, respectively. Shaded areas indicate the time ranges where the average amplitudes for the N1 and P3 were obtained. Right: Boxplot of the N1 and P3 amplitudes for identical (I) and different (D) T4. Whiskers indicate the whole data range, boxes the 25% to 75% quantiles and the median.

SUMMARY AND CONCLUSION

We demonstrated a successful integration of ear-EEG acquisition with live hearing device processing. Using ear-centred electrodes, AEPs could be measured while the hearing device inserted into the same ear was active. This result, along with extensive pilot studies not reported here, demonstrate that potential practical obstacles, such as electro-magnetic crosstalk between audio transducers and EEG electrodes that stand in the way of integrating ear-EEG and hearing devices can be overcome.

It was possible to verify perceived differences in the hearing device processing with AEP differences. The timing of the AEPs with respect to the audio signals revealed that the participants were able to detect the change in filter settings already based on the idle noise of the hearing device. Despite this unforeseen effect we could show that the ear-centred EEG electrode placement in combination with a wireless EEG amplifier and a hearing device, provides conclusive information about auditory perception in this context. Furthermore, the EEG analysis provided additional insight in the perception process that was not apparent from the psychoacoustic results and clearly demonstrates that special considerations are necessary when studying AEPs to stimulation with a live hearing device. Future work will include further evaluation of the current dataset, particularly a quantification of the importance of electrode positioning and the evaluation of single-subject and single-trial data.

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