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Toward EEG-Assisted Hearing Aids: Objective Threshold Estimation Based on Ear-EEG in Subjects With Sensorineural Hearing Loss

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Abstract

Electrophysiological feedback on activity in the auditory pathway may potentially advance the next generation of hearing aids. Conventional electroencephalographic (EEG) systems are, however, impractical during daily life and incompatible with hearing aids. Ear-EEG is a method in which the EEG is recorded from electrodes embedded in a hearing aid like earpiece. The method therefore provides an unobtrusive way of measuring neural activity suitable for use in everyday life. This study aimed to determine whether ear-EEG could be used to estimate hearing thresholds in subjects with sensorineural hearing loss. Specifically, ear-EEG was used to determine physiological thresholds at 0.5, 1, 2, and 4 kHz using auditory steady-state response measurements. To evaluate ear-EEG in relation to current methods, thresholds were estimated from a concurrently recorded conventional scalp EEG. The threshold detection rate for ear-EEG was 20% lower than the detection rate for scalp EEG. Thresholds estimated using in-ear referenced ear-EEG were found to be elevated at an average of 5.9, 2.3, 5.6, and 1.5 dB relative to scalp thresholds at 0.5, 1, 2, and 4 kHz, respectively. No differences were found in the variance of means between in-ear ear-EEG and scalp EEG. In-ear ear-EEG, auditory steady-state response thresholds were found at 12.1 to 14.4 dB sensation level with an intersubject variation comparable to that of behavioral thresholds. Collectively, it is concluded that although further refinement of the method is needed to optimize the threshold detection rate, ear-EEG is a feasible method for hearing threshold level estimation in subjects with sensorineural hearing impairment.

Keywords

audiometry, hearing aids, sensorineural hearing loss, auditory evoked potentials, auditory, electroencephalography

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Introduction

More than 5% of the world's population suffer from hearing impairment to a degree that is considered to be disabling (World Health Organization, 2012). The largest proportion of hearing impairment in the population is caused by irreversible sensorineural hearing loss associated with the aging process (Lin, Niparko, & Ferrucci, 2011; Linssen, van Boxtel, Joore, & Anteunis, 2013). With the anticipated increase in global life expectancy, the prevalence of hearing impairment is also predicted to rise in the future.

A hearing aid is a medical device that aims to make sound audible for hearing-impaired persons through amplification and other sound processing strategies. It is therefore crucial that hearing aids are fitted in close accordance with the hearing abilities of the individual user for hearing aids to provide significant improvements in communication and day-to-day activities.

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Traditionally, hearing aid fitting is carried out in the clinic using behavioral tests such as pure tone audiometry (PTA; Hughson & Westlake, 1944) and speech discrimination testing (Hagerman, 1982), which provide information on the degree and nature of the hearing loss.

Hearing loss is often progressive with increasing age, and, in some instances, it can fluctuate over shorter time periods. It would be advantageous if a hearing aid itself could assess the hearing loss of the user and tailor its audio processing according to this. Ideally, this process could be carried out outside of the clinic. Neural feedback from a user's auditory system to the hearing aid could thus be used to update the hearing aid fitting and improve its performance and usability for that individual. This approach could benefit all hearing aid users, particularly those suffering from fluctuating hearing loss. For example, Ménière's syndrome (Paparella & Sajjadi, 1999) sufferers are known to highly value the possibility of reprogramming their hearing aids on a regular basis (McNeill, McMahon, Newall, & Kalantzis, 2008).

Electrical potentials related to neural activity evoked by auditory stimulation can be recorded noninvasively from electrodes placed on the scalp (reviewed in Burkard, Eggermont, & Don, 2007). This approach can be utilized to obtain information on many aspects of the hearing of a subject. Measurements of electroencephalographic (EEG) signals have been used to establish hearing threshold levels (Dimitrijevic et al., 2002; Lins et al., 1996; Mühler, Mentzel, & Verhey, 2012; Picton, Dimitrijevic, Perez-Abalo, & Van Roon, 2005; Rance, Rickards, Cohen, De Vidi, & Clark, 1995; Seidel, Flemming, Park, & Remmert, 2015), frequency selectivity (Butler, 1968) and loudness perception (Ménard, Gallégo, Berger-Vachon, Collet, & Thai-Van, 2008), which can be used for hearing aid fitting and selecting audio processing strategies. Integration of EEG recording into hearing aids could therefore potentially provide an objective tool for individualized refitting without the need for costly clinical testing.

Several technical issues prevent traditional EEG recording from being used during everyday life. First, conventional EEG recording equipment is obtrusive because it requires the use of wired electrodes attached to the scalp and typically uses a relatively large biosignal amplifier. Second, assistance of trained personnel is required to set up recording sessions, perform electrode placement and impedance checks, and control the procedure. Applications are therefore often restricted to the hearing screening of newborns and diagnostic testing in prelingual infants and adults who are difficult to test. Finally, the recordings are performed in specialized laboratories or clinics, involving high costs and an inconvenience to the user.

Ear-EEG is a relatively new EEG recording approach in which the EEG is recorded from electrodes placed in or around the ear (Debener, Emkes, De Vos, & Bleichner, 2015; Looney et al., 2011, 2012). Ear-EEG thus enables discrete monitoring of brain activity outside of the laboratory in the natural environment and everyday life of the users. The technology is still under development; thus, the different ear-EEG systems are all experimental platforms requiring as much assistance during setup and operation as conventional scalp EEG setups. Most ear-EEG recordings have been performed under well-controlled laboratory conditions using stationary EEG systems, but portable systems have been used to record EEG during daily life situations (e.g., Debener, Minow, Emkes, Gandras, & De Vos, 2012; De Vos, Gandras, & Debener, 2014; Kappel & Kidmose, 2018). With the refinement of ear-EEG technology and the development of miniature mobile EEG amplifiers (Zhou et al., 2016), the integration of ear-EEG into hearing aids is within technological reach. Therefore, investigations of possible applications within audiology and hearing aids are relevant and necessary.

Initial investigations have shown that well-established auditory evoked potentials, such as the P_1–N_1–P_2 complex and the auditory steady-state response (ASSR), can be observed from ear-EEG recordings (Looney et al., 2012; Mikkelsen, Kappel, Mandic, & Kidmose, 2015). Based on this, the feasibility of ear-EEG in hearing threshold estimation was evaluated in a recent study in normal hearing subjects (Christensen, Harte, Lunner, & Kidmose, 2018). It was demonstrated that ASSR thresholds, though elevated compared with ASSR thresholds determined from conventional scalp EEG, can be determined from ear-EEG recordings made with both measurement and reference electrodes placed within the same ear. The aim of the current study was to determine whether ear-EEG could be used to estimate ASSR hearing thresholds of subjects with sensorineural hearing impairment. In conventional EEG, neural recruitment results in more accurate ASSR threshold estimates, which are closer to gold standard PTA thresholds in subjects with sensorineural hearing impairment than those in subjects without hearing loss or purely conductive hearing loss (Mühler et al., 2012; Picton et al., 2005; Seidel et al., 2015). As neural recruitment is expected to also occur in neural sources dominating the ear-EEG, this effect can also be expected to affect ear-EEG ASSR thresholds.

Here, ASSR thresholds to CE-chirp stimuli were estimated using both in-ear ear-EEG and conventional scalp EEG in parallel to evaluate ear-EEG ASSR thresholds in comparison to scalp ASSR thresholds. Furthermore, the ear-EEG thresholds for hearing-impaired subjects found here were compared with ear-EEG thresholds for normal hearing subjects (Christensen, Harte, et al., 2018). As the study was not intended as a test of the usability of the ear-EEG setup during daily life, the quality of the acquired EEG data was optimized at the expense of the discreetness and user friendliness of the setup. The recordings were thus conducted under well-controlled laboratory conditions using a conventional EEG amplifier, and the ear electrodes were thoroughly prepared in a way similar to conventional scalp electrodes.

Methods

Subjects

Nineteen volunteer subjects (16 females and 3 males) aged 67.3 ± 9.6 years (ranging from 52 to 79 years) diagnosed with sensorineural hearing loss of 30 to 65 dB hearing level (dB HL) at 0.5, 1, 2, and 4 kHz were recruited for the study. One subject was excluded from the study, as the test was stopped shortly into the ASSR measurement due to unrest and general discomfort. Another subject was excluded because the EEG recordings were highly contaminated by muscle artifacts due to teeth grinding. Two test subjects were only stimulated in one ear because of discomfort with binaural stimulation. Hence, of the 38 ears originally included in the study, recordings from 32 ears were included in the data analysis. All ears had a middle ear pressure within ±50 daPa. None of the subjects had conductive hearing loss, as measured by bone conduction audiometry. All experimental procedures were approved by the Danish Health and Medicines Agency (ref no. 2015111607) and the Regional Committee on Health Research Ethics (ref no. 1-10-72-333-15), and informed consent was obtained from all subjects before inclusion in the study.

Experimental Setup

Individual, custom-made earpieces (Figure 1(a) and (b)) were produced for all subjects using the same manufacturing processes (impression, three-dimensional scanning, CAD modeling, and three-dimensional printing) that are standard in the production of customized hearing aid earpieces. Six silver electrodes (two in the concha and four in the ear canal) and a small hearing aid-type receiver (SPEAKER ASSY, 85 MINIFIT, Oticon, Smørum, Denmark) were mounted onto the earpieces. The electrodes were connected to a g.USBamp amplifier (g.tec medical engineering GmbH, Schiedlberg, Austria) from which data were collected in MATLAB (2012a, MathWorks, Natick, MA, USA) with 1,200 samples per second. Sound stimuli were presented to the earpiece receiver via an ESI U46 XL soundcard (ESI Audiotechnik GmbH, LeonBerg, Germany) using a sampling frequency of 48 kHz. A g.tec trigger box (g.TRIGbox) that received input from the soundcard was used to trigger ASSR data collection with a trigger frequency of 0.25 Hz. The receiver was calibrated using an ear and cheek simulator (43AG, GRAS Sound and Vibration, Holte, Denmark) powered by a 12AA power module (GRAS) and calibrated using a 42AA pistonphone (GRAS). All recordings were performed in a double-walled, sound-attenuated booth at Eriksholm Research Centre, Snekkersten, Denmark. The background noise levels measured inside the booth were within the requirements of International Organization for Standardization (ISO) 8253-1 (ISO, 2010) and ISO 389-9 (ISO, 2009).

Experimental Procedures

The experimental procedures were completed in two test sessions performed on 2 separate days. Otoscopy and tympanometry (Titan, Interacoustics, Middelfart, Denmark) were performed to ensure that only test subjects with healthy middle ears participated in the study. During the first session, monaural pure-tone audiometry (ISO, 2010) was performed using an Interacoustics AC40 audiometer to determine the hearing threshold of all test subjects at 0.5, 1, 2, and 4 kHz. Moreover, ear impressions were taken for the production of the individualized earpieces to be used in the second session.

During the second session, hearing thresholds to one octave wide narrowband (NB) CE-chirp stimuli centered at 0.5, 1, 2, and 4 kHz (Elberling & Don, 2010) were determined from behavioral audiometry using the earpiece speakers. Thereafter, ASSR thresholds were determined using the individualized ear-EEG earpieces and conventional EEG electrodes.

Stimulus levels. Individual sensation levels and maximum endurable intensity ranges of the test subjects to chirp stimuli were determined to guide the intensity levels used in the ASSR measurements. Sensation levels were determined for each of the four different chirps individually by the ascending method described in ISO 8253-1 (ISO, 2010) using a custom routine written in MATLAB. The chirp stimuli were presented with the repetition rates used in the ASSR measurements in blocks of 1-second duration. The maximum endurable intensity range of the subjects was estimated by presentation of the multichirp stimuli binaurally with increasing sound pressure levels in 5 dB steps for 10 seconds at each level starting at 5 dB sensation level (dB SL) and up to the level that each test subject found the sound to be unpleasantly high or to a maximum of 35 dB SL. Furthermore, the sound levels used for the ASSR measurements did not exceed the maximum noise exposure regulations of the Danish Work Environment Authority (www.at.dk).

ASSR measurements. The skin in the ear canal and concha was cleaned with ethanol and prepared with an abrasive gel to optimize skin conductivity before insertion of the

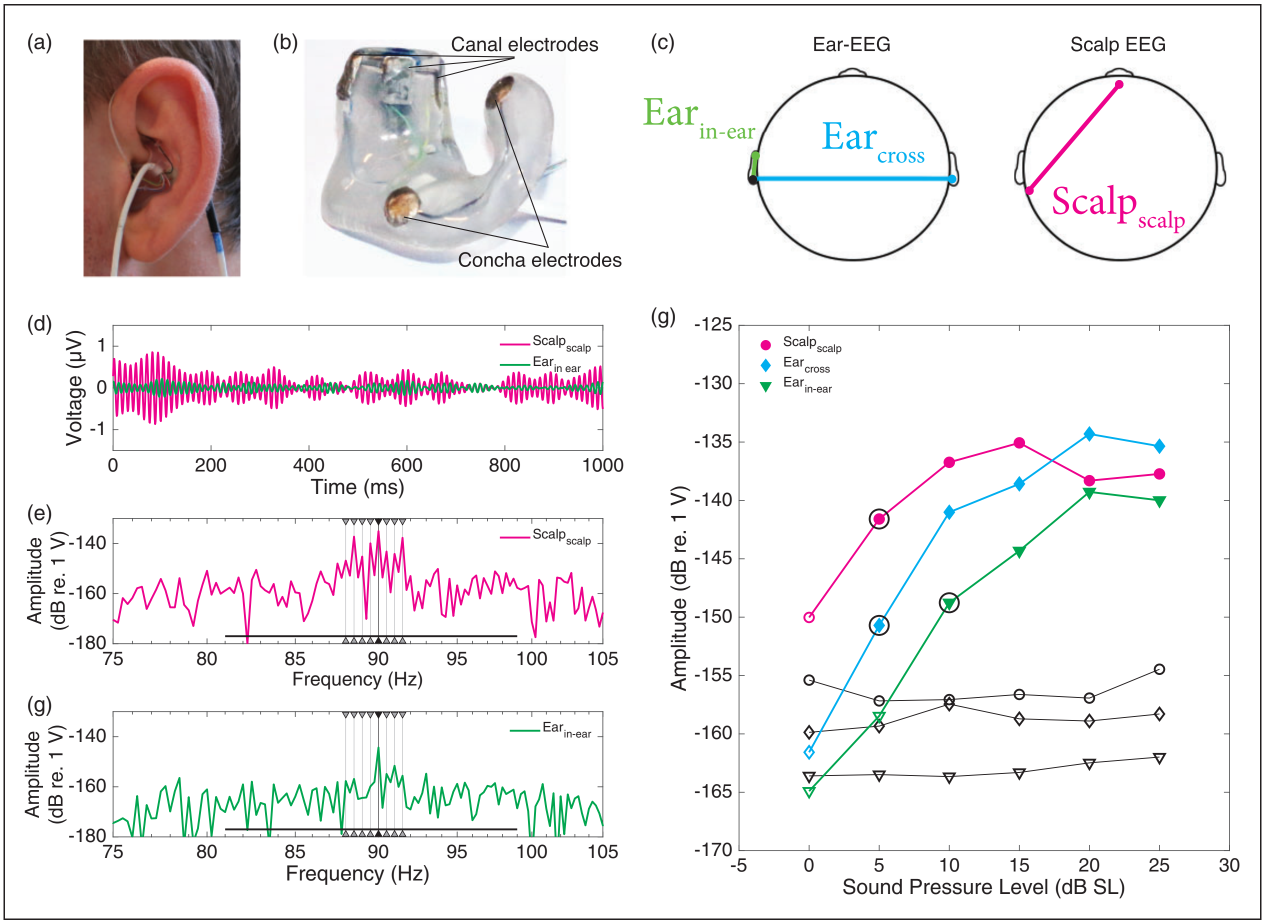

Figure 1. Individualized ear-EEG earpieces were made for all subjects. (a) An earpiece placed in the ear. (b) An example of the earpieces used in the study showing the silver electrodes embedded on the surface. Two electrodes were embedded in the concha and four in the ear canal. (c) Ear-EEG was acquired from an earpiece electrode as the measurement electrode and with the reference electrode either in the same ($Ear_{in\text{-}ear}$) or opposite ear (Ear_{cross}). In addition, EEG was recorded using a conventional scalp configuration with a measurement electrode placed on the mastoid and a reference electrode placed on the forehead ($Scalp_{scalp}$). (d) Averaged EEG data recorded using $Scalp_{scalp}$ and $Ear_{in\text{-}ear}$ electrode configurations at 15 dB SL. (e) Fast Fourier Transform (FFT) spectrum of $Scalp_{scalp}$ shown in panel (d). Repetition rates are indicated by vertical lines. Frequency bins included in the noise measurement are indicated by the horizontal black line. (f) FFT spectrum of $Ear_{in\text{-}ear}$ shown in panel (d). (g) Example of ASSR (coloured symbols) and related noise levels (black symbols) as a function of sound pressure level. Filled symbols indicate the statistical significance of responses relative to noise. Thresholds are indicated by black circles. dB SL = dB sensation level; EEG = electroencephalographic.

earpieces. The electrodes were coated with conductive gel (EEG Paste and Ten20), and additional gel (GAMMAgel and g.tec) was applied to the electrodes in the concha after insertion of the earpiece if needed. The electrode impedances were measured using g.Recorder software (g.tec) both before and after the ASSR measurement. Earpiece electrodes with an impedance above 20 kΩ and scalp electrodes with an impedance above 5 kΩ in either measurement were omitted from the subsequent analysis.

During the ASSR measurements, the subjects were placed in a relaxed supine position and encouraged to relax or, if possible, to sleep to minimize the EEG noise levels. A multiple stimulus paradigm (John, Lins, Boucher, & Picton, 1998) was used to record ASSR responses to the four NB CE-chirp stimuli in both ears simultaneously using different repetition rates for the individual chirp stimulus in the two ears. In doing so, the individual chirps were presented in 5 dB steps from −5 dB SL to the maximum endurable sound pressure level of the individual subject or a maximum of 35 dB SL. To prevent masking effects between neighboring frequencies, level differences between the sound stimuli of two adjacent frequencies were limited to 20 dB. Repetition rates of approximately 90 Hz were chosen to avoid the effects of sleep on the ASSR (Cohen, Rickards, & Clark, 1991). In left ears, repetition rates of 88.5, 89.5, 90.5, and 91.5 Hz were used for the 0.5, 1, 2, and 4 kHz chirps, respectively. In right ears, 88, 89, 90, and 91 Hz were used as repetition rates for the 0.5, 1, 2, and 4 kHz chirps, respectively. The stimuli were composed of stimulation trials of 32-second long chirp trains windowed with a 0.15-second Blackman window to provide a ramped rise and fall of the chirp stimuli. Chirp stimuli were used, as they compensate for the different traveling wave delay of the frequency components in the cochlea to give higher temporal synchronization of the neural excitation and thereby a larger ASSR response (Elberling, Don, Cebulla, & Stürzebecher, 2007). Every second stimulation trial was inverted to minimize possible electromagnetic artifacts in the estimated evoked potentials.

The EEG data were recorded from 12 earpiece electrodes and 2 conventional scalp electrodes placed on the mastoids. The reference electrode was placed on the forehead, and the ground electrode was placed on the cheek. During recording, the quality of the recorded EEG data was ensured by monitoring the number of accepted epochs. To do so, the EEG data were high-pass filtered using a fourth-order Butterworth filter with a cutoff frequency of 70 Hz, partitioned into 4-second long epochs, and the number of epochs with a maximum absolute value below 40 μV was counted. Up to 17 stimulation trials or a total stimulation time of 544 seconds was allowed to obtain 120 epochs of 4 seconds duration from each of the mastoid electrodes and at least one electrode from each earpiece for each sound intensity level.

Analysis

ASSR thresholds were determined from three different electrode configurations (Figure 1(c)) to evaluate ear-EEG in comparison to conventional EEG. An electrode configuration with a measurement electrode placed on the mastoid and a reference electrode placed on the forehead ($Scalp_{scalp}$) was used to record conventional scalp EEG. In addition, ear-EEG was measured with two different electrode configurations with the reference electrode placed in either the opposite ear (Ear_{cross}) or the same ear ($Ear_{in\text{-}ear}$) as the measurement electrode. The analysis of the EEG data was performed off-line after the recordings. First, all channels were filtered using a fourth-order Butterworth bandpass filter with cutoff frequencies of 75 and 105 Hz. Subsequently, separate $Scalp_{scalp}$ data sets were created from the two mastoid channels. Furthermore, *cross referenced* (Ear_{cross}) data sets were created by rereferencing ear channels from one earpiece to ear channels on the other earpiece, and *in-ear referenced* ($Ear_{in\text{-}ear}$) data sets were created by referencing ear channels to the other ear channels of the same earpiece. All data sets were then split into epochs of 4-second duration, and epochs with absolute values exceeding 40 μV were rejected. Data sets with less than 25 remaining epochs were excluded from the analysis. Finally, the signal-to-noise ratios (SNRs) for all data sets were individually determined for all four repetition rates as the ratio of the amplitude at the repetition rate and the average noise amplitude ±9 Hz (excluding the other repetition rates) relative to the individual repetition rate. The frequency range from 99.75 to 100.25 Hz was excluded to avoid the effects of the second harmonic of line noise. For an individual subject, the electrode configuration giving the highest SNR at the highest stimulation level was determined for the $Scalp_{scalp}$, Ear_{cross}, and $Ear_{in\text{-}ear}$ data sets for each of the four NB CE-chirps. These configurations were then used across sound levels to estimate hearing thresholds. To automate this, ASSR responses were detected using a statistical F test ($\alpha = .05$) as described by Zurek (1992). Hence, the ratio between the power in the response bin of the Fast Fourier Transform and the average power in the 64 neighboring noise bins (±9 Hz excluding all 8 response bins) was evaluated against an F-distribution with 2 and 128 degrees of freedom. A hearing threshold was defined as the lower of the two lowest successive significant responses (Figure 1(g)). If two successive significant responses were not found, the threshold was considered undetected. Holm's (1979) correction was applied to avoid multiple comparison problems.

Results

Signal and noise amplitudes of the recorded EEG and the resulting SNRs at 25 dB SL are shown in Figure 2 for the scalp EEG ($Scalp_{scalp}$) and in-ear ear-EEG ($Ear_{in\text{-}ear}$) configurations. The signal levels were found to be significantly lower for in-ear ear-EEG than for scalp EEG at 0.5 and 1 kHz, whereas the noise levels were significantly lower for in-ear ear-EEG than for scalp EEG at all frequencies (paired one-tailed t test, $\alpha = .05$). No significant difference was found in SNR at any of the frequencies (Figure 2(b)).

Within the individual ear, the SNR was found to vary more than 20 dB (Figure 3) between the best and worst in-ear electrode configurations. In most subjects, the best electrode configuration was found to be an electrode pair maximizing the interelectrode distance combining a measurement electrode in the ear canal with a reference electrode in the concha (Table 1). However, electrode configurations with both measurement and reference electrode placed within the concha or ear canal alone were found to give the best SNR in some subjects.

Behavioral thresholds to pure tone and chirp stimuli along with objective ASSR thresholds are shown in Figure 4. PTA and behavioral chirp thresholds were obtained for all ears, whereas ASSR thresholds in some cases could not be obtained (Figure 5). Correlation

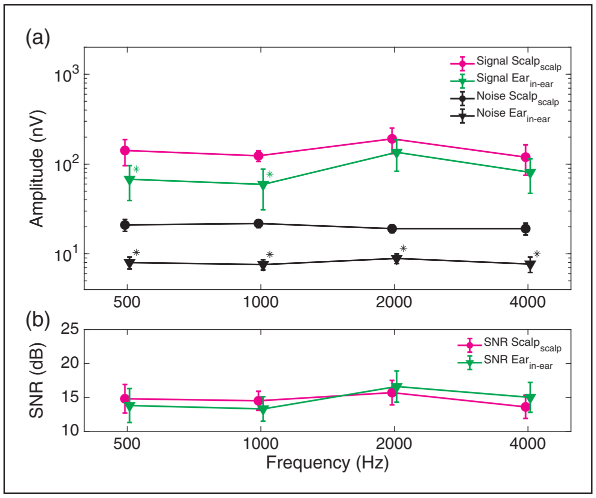

Figure 2. (a) Mean and standard deviations of signal (nV), noise (nV), and (b) average SNR (dB) for both scalp EEG (Scalp$_{scalp}$) and ear-EEG (Ear$_{in-ear}$). Amplitudes for both scalp and ear-EEG were based on paired measurements of 120 epochs of data from the individual test subjects measured at 25 dB SL. Bars indicate the mean ± standard error of the mean. SNR = signal-to-noise ratio.

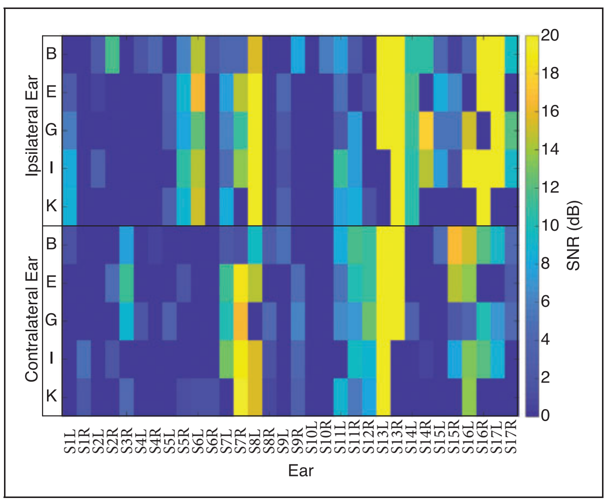

Figure 3. Distribution of SNR across electrodes and ears (S1L: Subject 1 stimulus in left ear, etc.) at the highest sound pressure level for 4 kHz stimuli with reference to an electrode in the upper part of concha (A electrode, see electrode nomenclature in Looney et al., 2012). The five upper electrodes, B (concha electrode) and E, G, I, and K (all channel electrodes), were situated in the ear ipsilateral to the stimulation side, whereas the five lower electrodes, B (concha electrode) and E, G, I, and K (all channel electrodes), were situated in the ear contralateral to the stimulation. All electrodes were referenced to the A electrode of the same ear. SNR = signal-to-noise ratio.

coefficients between ASSR and behavioral chirp thresholds were found to be 0.64–0.93, 0.61–0.91, and 0.52-0.90 across frequencies for the Scalp$_{scalp}$, Ear$_{cross}$, and Ear$_{in-ear}$ configurations, respectively (Figure 6).

Table 1. Optimal Reference Pairs for In-Ear Measurements.

	0.5 kHz	1 kHz	2 kHz	4 kHz
Canal ref. concha	62.5	68.8	59.4	62.5
Canal ref. canal	18.8	18.8	18.8	12.5
Concha ref. concha	18.8	12.5	21.9	25.0

Note. Distribution (percentage) of electrode reference pairs used for in-ear auditory steady-state response measurements. The reference pair giving the highest signal-to-noise ratio at the highest input sound pressure level was used across all sound levels to estimate hearing thresholds.

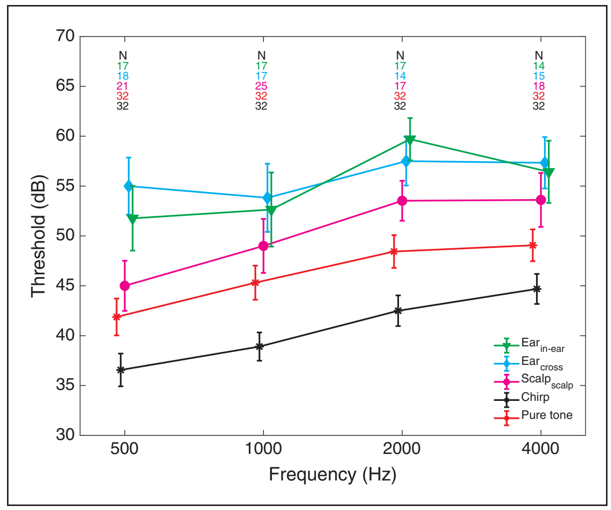

Figure 4. Behavioral hearing thresholds to pure tone and chirp stimuli along with ASSR thresholds to chirp stimuli. Pure tone thresholds are given in dB HL, whereas chirp thresholds are given in dB normal hearing level (dB nHL; 25.5, 24.0, 30.5 and 34.5 dB peak-to-peak equivalent threshold sound pressure levels were used as 0 dB nHL for 0.5, 1, 2, and 4 kHz, respectively). Bars indicate the mean ± standard error of the mean. The data are tabulated in Table S1 in the Supplementary Material.

To evaluate ear-EEG in comparison to conventional scalp EEG, Ear$_{cross}$ and Ear$_{in-ear}$ ASSR thresholds were compared with Scalp$_{scalp}$ thresholds (Figure 7). To do so, paired statistical tests were used to account for intersubject variation and differences in threshold numbers. Ear$_{cross}$ thresholds were found to be elevated 6.9 ± 2.4, 5.0 ± 3.7, 2.7 ± 2.4, and 2.3 ± 1.7 dB (mean ± standard error of the mean) relative to thresholds estimated using the Scalp$_{scalp}$ configuration. The difference was significant at 0.5 kHz (paired t test $t = 2.920$, $p = .012$). Thresholds estimated using the Ear$_{in-ear}$ configuration were found to be elevated 7.5 ± 3.7, 4.3 ± 3.2, 7.5 ± 3.0, and 0.8 ± 1.7 dB relative to thresholds estimated using the Scalp$_{scalp}$ configuration. The difference was significant at 2 kHz (paired t test: $t = 2.540$, $p = .025$). Moreover, Ear$_{in-ear}$ thresholds were compared with Ear$_{cross}$ thresholds to investigate the effect of having the reference electrode in the same or opposite ear as the measurement electrode. No significant difference was found for any of the frequencies (-0.4 ± 2.9, 1.2 ± 2.5, 3.8 ± 2.4, and -0.9 ± 0.6 dB).

In addition to differences in threshold, it was tested whether the variance of the ear-EEG thresholds was elevated compared with the variance of scalp EEG thresholds and behavioral thresholds. Hence, the threshold variances at the individual frequencies of the $\mathrm{Ear}_{\mathrm{cross}}$ and $\mathrm{Ear}_{\text{in-ear}}$ data sets were compared pairwise with the variance of the $\mathrm{Scalp}_{\mathrm{scalp}}$ data set and behavioral thresholds using a one-sided F test. No significant differences in variance were found between the ear-EEG and scalp EEG or behavioral thresholds at any of the frequencies.

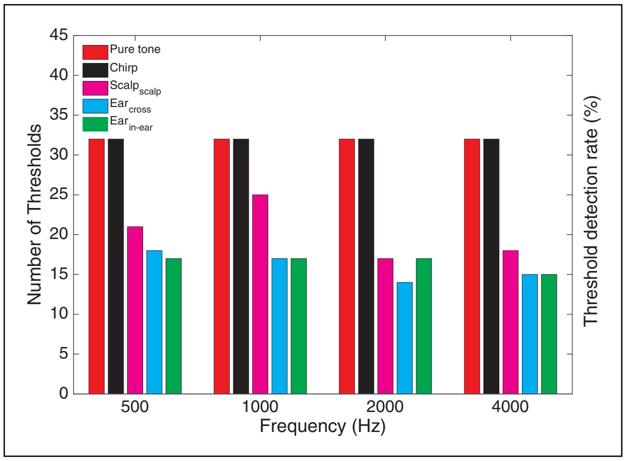

Figure 5. Distribution of thresholds obtained across frequencies for both behavioral and electrophysiological methods.

Discussion

The aim of this study was to determine whether ear-EEG can be used to estimate hearing thresholds in sensorineural hearing-impaired subjects. Physiological thresholds were obtained in parallel using both conventional scalp EEG and ear-EEG and were compared with behavioral thresholds. The approach used in this study was previously validated on normal hearing subjects (Christensen, Harte, et al., 2018).

Scalp ASSR vs PTA

In the current study, ASSR thresholds to 90 Hz modulated NB CE-chirp stimuli were found to be elevated at an average of 2.8 to 5.3 dB relative to PTA thresholds when using an electrode configuration with a forehead

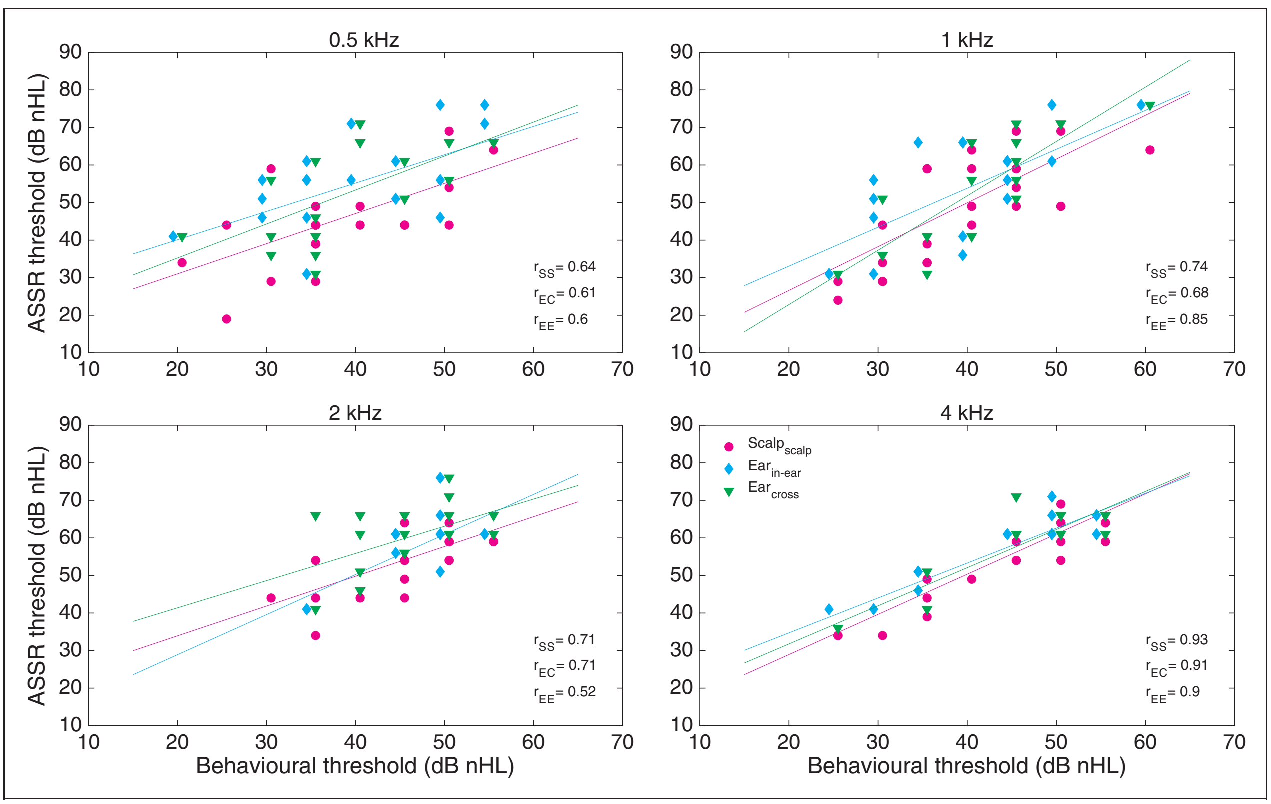

Figure 6. Scatter plot and corresponding linear regression line between ASSR and behavioral chirp thresholds for the three different electrode configurations: $\mathrm{Scalp}_{\mathrm{scalp}}$, $\mathrm{Ear}_{\mathrm{cross}}$, and $\mathrm{Ear}_{\text{in-ear}}$ at 0.5, 1, 2, and 4 kHz. Correlation coefficients are given in the lower right corner for $\mathrm{Scalp}_{\mathrm{scalp}}$: r_{SS}, $\mathrm{Ear}_{\mathrm{cross}}$: r_{EC}, and $\mathrm{Ear}_{\text{in-ear}}$: r_{EE}. ASSR = auditory steady-state response; dB nHL = dB normal hearing level.

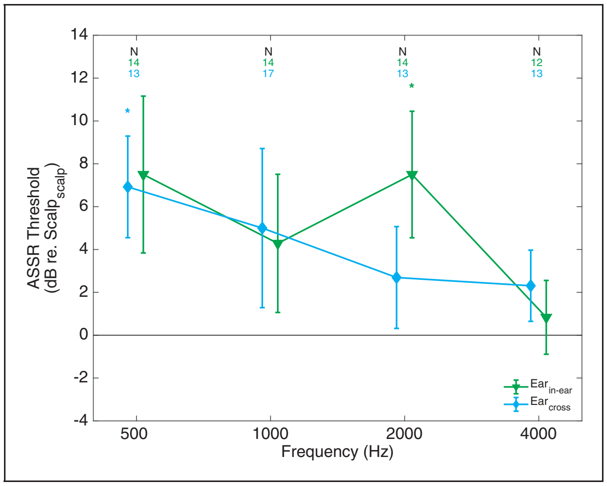

Figure 7. Paired differences in ASSR thresholds for Ear_{cross} and $Ear_{in\text{-}ear}$ relative to $Scalp_{scalp}$. Bars indicate the mean ± standard error of the mean. Asterisks indicate statistical significance (paired *t* test, $p < .05$). Data are tabulated in Table S2 in the Supplementary Material. ASSR = auditory steady-state response.

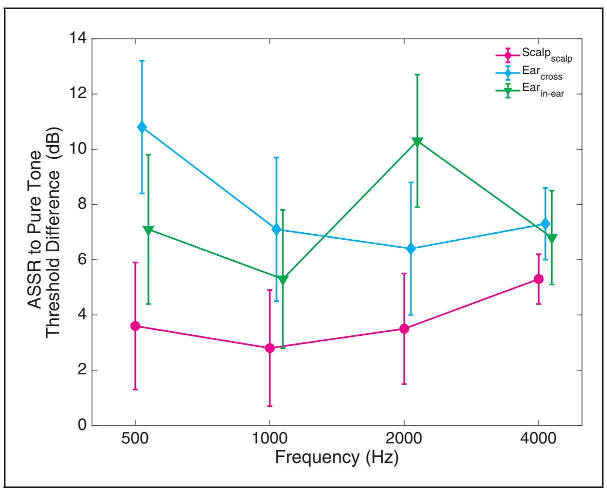

Figure 8. Mean and standard deviation of paired differences between ASSR thresholds (dB normal hearing level) and behavioral pure-tone thresholds (dB HL). Bars indicate the mean ± standard error of the mean. Data are tabulated in Table S3 in the Supplementary Material. ASSR = auditory steady-state response.

reference ($Scalp_{scalp}$, Figure 8). Mühler et al. (2012) and Seidel et al. (2015) estimated ASSR thresholds to 40 Hz modulated chirp stimuli using a conventional scalp EEG setup. These authors found ASSR to PTA threshold differences for adult hearing-impaired subjects of 8 to 12 and 5 to 11 dB, respectively. The discrepancy in ASSR to PTA threshold differences can be attributed to differences in measurement procedures. Both PTA and ASSR thresholds are depended on the particular procedures used to obtain them. For clinical PTA thresholds, the Hughson and Westlake's (1944) procedure is most often used, where the threshold is defined as the lowest sound intensity, to which the listener responds in 2 out of 3 presentation ascents. Using a different method, such as an adaptive forced choice procedure (Levitt, 1971), may yield a different threshold. To enable the comparison of PTA thresholds between clinics and establish normative data, it is therefore important to use the same standardized procedure. For physiological thresholds, the automatic signal detection algorithm that is used to determine whether a response is present will have a certain sensitivity. Using a different algorithm with a higher sensitivity will yield a test that needs a lower effective SNR to determine whether a response is present. This will, in effect, lower the final threshold. The choice of filter settings, how artifacts are handled, the type of averaging procedure, and other recording parameters will lead to different SNRs of the ASSR response at a given stimulus presentation level, all effecting the final threshold obtained. The choice of the overall threshold determination procedure will also alter the final ASSR threshold. Therefore, in light of the procedural impact on thresholds, differences between ASSR and PTA thresholds are expected. As the PTA procedure can be assumed to be standardized between the present and the earlier studies, some differences in the procedure for obtaining physiological thresholds may have given rise to the disparity in ASSR to PTA threshold differences found between the studies. Furthermore, both PTA and ASSR thresholds are dependent on the repetition rate of the used sound stimuli. Behavioral thresholds are lower with an increase in repetition rate (Gøtsche-Rasmussen, Poulsen, & Elberling, 2012). On the other hand, ASSR thresholds in adults have empirically been found to be higher for stimuli with an 80 Hz repetition rate than for an 40 Hz repetition rate (Van Maanen & Stapells, 2005) as the magnitude of the ASSR response is smaller at 80 Hz than at 40 Hz (Picton, John, Dimitrijevic, & Purcell, 2003). The background noise levels also differ with repetition rate. Normally, background EEG has a $1/f$ characteristic, implying that noise will be higher on average for the lower repetition rates. In general, lower EEG noise levels would result in lower ASSR thresholds and will tend to lower the observed difference between ASSR and PTA thresholds. Seidel et al. (2015) did not report EEG noise levels, but the noise levels reported by Mühler et al. (2012) are comparable to those recorded in this study (Figure 2); therefore, noise level differences cannot explain the reduction in ASSR to PTA thresholds differences found between these previous studies and this study.

Finally, it should be noted that the difference between ASSR and PTA thresholds is influenced by the degree of sensorineural hearing impairment. This difference is typically smaller for hearing-impaired than normal hearing subjects (Picton et al., 2005; Seidel et al., 2015). In the study by Mühler et al. (2012), test subjects were defined as hearing impaired at all frequencies, even though they had a PTA threshold above 30 dB HL at only one of the frequencies tested. Subjects with high-frequency hearing loss but with normal low-frequency hearing or vice versa were, despite only having partial hearing impairment, still defined as hearing impaired at all frequencies. This approach may, as pointed out by Seidel et al. (2015), have led to biased ASSR to PTA threshold differences for hearing-impaired subjects at the frequencies where they actually had normal hearing. To prevent this effect, Seidel et al. categorized the individual thresholds by the pure tone value rather than dividing subjects into normal hearing and hearing-impaired groups. The average hearing impairment was not stated, but their definition of hearing impairment (PTA > 20 dB HL) was lower than the lower limit of 30 dB HL used in this study; thus, the average hearing impairment may have been less in their study. Collectively, some of the reduction found in ASSR to PTA threshold differences found between the earlier and the current studies may be explained by differences in the hearing impairment of the subjects tested. The variance of the ASSR to PTA threshold differences found here was comparable to those found for hearing-impaired subjects by Mühler et al. (2012) and Seidel et al. (2015).

Comparison of ear-EEG and scalp EEG

Ear-EEG is inherently characterized by small interelectrode distances and a restricted spatial distribution. Thus, the absolute amplitudes of in-ear ear-EEG signals are generally smaller than those recorded using conventional scalp EEG (Figure 2; Christensen, Harte, et al., 2018; Kidmose, Looney, Ungstrup, Rank, & Mandic, 2013; Looney et al., 2012). However, the ASSR SNRs of in-ear ear-EEG are comparable to those measured on the scalp, as the reduced signal amplitude is accompanied by a comparable reduction in the EEG noise levels (Figure 2; Bleichner et al., 2015; Kidmose et al., 2013; Looney et al., 2012; Mikkelsen et al., 2015).

Although the maximum SNR of the ear-EEG ASSR was comparable to that of scalp EEG, the SNR was found to be highly variable across the different in-ear electrode configurations of the individual subjects (Figure 3). This finding is in line with the findings of Bleichner et al. (2016) and Denk et al. (2018), that both electrode orientation and distance have an effect on the recorded ear-EEG. Furthermore, the electrode configuration yielding the highest SNR was found to vary between subjects (Table 1). Some of this variation may have been caused by bridging of neighboring electrodes in the concha or canal in some subjects. The intersubject variation in optimal electrode configuration is, however, in accordance with the findings of Looney et al. (2012), suggesting that the ASSR potential field is distributed differently from subject to subject. Collectively, optimization of the electrode reference configuration for the individual subject seems crucial to obtaining good SNRs in in-ear ear-EEG. In this study, the electrode pair giving the maximum SNR at the highest tested sound intensity was used across the different ASSR recordings of the hearing test. This practice gives the highest SNR values but is based on the assumption that the electrode–skin connection is stable across the ASSR recordings. Under changing measurement conditions, the use of a weighted average of multiple electrodes may give a smaller but more robust SNR (Kappel, Christensen, Mikkelsen, & Kidmose, 2016). Moreover, high-density ear-EEG measurements provide a more detailed description of the spatial distribution of the ASSR across the ear (Christensen, Kappel, & Kidmose, 2018; Kappel & Kidmose, 2017); therefore, the inclusion of more electrodes or the exploitation of other electrode locations within or around the ear could lead to improved SNRs in the in-ear recorded EEG. Generally, the spatial distribution of the ASSR response across the ear needs to be investigated further.

In normal hearing subjects, nearly all thresholds could be detected using scalp EEG, whereas only 50% of the thresholds were detected with in-ear ear-EEG (Christensen, Harte, et al., 2018). In this study, the overall detection rate of ear-EEG was also found to be 50%, supporting the earlier finding that threshold detection based on ear-EEG is a challenging task. Nevertheless, the number of detected thresholds was only 20% lower for ear-EEG than for scalp EEG (0–32% lower for the individual frequencies; Figure 5). Although it was generally more difficult to obtain an ASSR threshold from hearing-impaired subjects, the performance of ear-EEG, with regard to threshold detection, was nevertheless more comparable to scalp EEG in this group. In normal hearing subjects, the use of a reference electrode in the opposite ear resulted in higher threshold detection rates compared with the use of the in-ear reference. This notion was not supported by the results of this study, where no difference was found in the number of detected thresholds between the $\text{Ear}_{\text{cross}}$ and $\text{Ear}_{\text{in-ear}}$ configurations. Part of the relatively low ASSR detection rates can be explained by the relative conservative threshold criterion of two successive significant measurements. Nevertheless, this criterion was used to minimize the probability of false positives and biased ASSR to PTA threshold differences. Moreover, the maximum sound pressure level used during the ASSR measurements in the present study was limited to 35 dB SL or the level maximally endured by the individual test subject in the pretest. On average, the maximum stimulation level was

26.9 dB SL with a standard deviation of 5.2 dB. This finding is approximately 20 to 30 dB lower than the 45 to 55 dB SL used in normal hearing subjects (Christensen, Harte, et al., 2018).

The ear-EEG and scalp EEG thresholds were found to be moderately to highly correlated with the behavioral thresholds, and the correlation coefficients were comparable between configurations (Figure 6). Moreover, the correlation coefficients were comparable to those reported in earlier chirp ASSR studies (Mühler et al., 2012; Seidel et al., 2015).

Thresholds determined using ear-EEG were generally elevated compared with scalp EEG thresholds (Figure 4). The difference was only significant at 0.5 kHz for the Ear_{cross} and 2 kHz for the $Ear_{in\text{-}ear}$ configurations (Figure 7), but this could be a consequence of the relatively limited number of thresholds detected. Overall, the results suggest that larger correction values are needed to estimate behavioral thresholds from ear-EEG ASSR thresholds than from scalp ASSR thresholds. More importantly, no difference was found in the intersubject variance of the threshold means between the ear-EEG and scalp EEG at any of the frequencies. This suggests that behavioral hearing thresholds can be estimated with the same precision from in-ear ear-EEG as from scalp EEG.

The experimental procedure of this study did not include an evaluation of test–retest reliability. For scalp EEG, ASSR thresholds were found to have moderately strong test–retest reliability with correlation coefficients of .75 to .93 (Kaf, Sabo, Durrant, & Rubinstein, 2006). This should be investigated for ear-EEG in future studies.

Effect of sensorineural hearing impairment

The average ASSR amplitudes were higher relative to the sensation level in hearing-impaired subjects than in normal hearing subjects (Figure 9). This finding is in accordance with earlier reports and has been attributed to increased neural recruitment in sensorineural hearing-impaired subjects (Dimitrijevic et al., 2002; Picton et al., 2005). Interestingly, the relative increase in ASSR amplitude from normal hearing to hearing-impaired subjects was pronounced for in-ear ear-EEG compared with both cross-ear ear-EEG and scalp EEG (Figure 9). The ASSR originates from multiple neural sources distributed along the auditory pathway (Herdman et al., 2002) and can be regarded as the sum of these sources weighted by their individual coupling to the EEG electrodes. As discussed by Bleichner and Debener (2017), neural sources may be projected differently to in-ear ear-EEG and scalp EEG, so the weights of the different neural sources are likely different for in-ear ear-EEG and scalp EEG electrode configurations. However, no systematic macroscopic

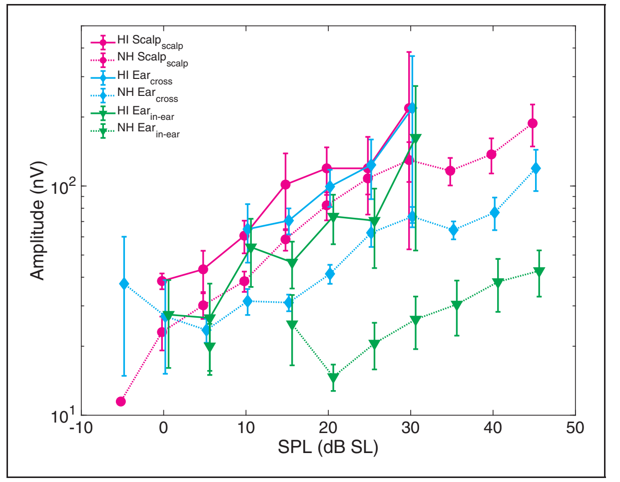

Figure 9. Mean and confidence interval of ASSR hearing thresholds in dB relative to sensation level. Data are tabulated in Table S4 in the Supplementary Material. ASSR = auditory steady-state response; dB SL = dB sensation level.

anatomical differences are expected between the heads of normal hearing and hearing-impaired subjects; thus, the neural activity is expected to be coupled equally well to the in-ear electrodes in these two groups. The fact that the relative increase in ASSR amplitude from normal hearing to hearing-impaired subjects was larger for the in-ear configuration relative to the Ear_{cross} and scalp configurations suggests that increased neural recruitment is especially pronounced in the sources dominating the ear-EEG compared with the neural sources of the scalp EEG. Further investigation is needed to clarify this finding.

The relative increase in ASSR amplitude from normal to hearing-impaired subjects was reflected in lower ASSR threshold levels relative to both chirp sensation level (Figure 10) and PTA thresholds (Figure 8) for the hearing-impaired subjects. This finding is in agreement with that of earlier ASSR studies (Dimitrijevic et al., 2002; Mühler et al., 2012; Picton et al., 2005; Seidel et al., 2015).

Ear-EEG-assisted fitting of hearing aids

From an audiological perspective, the feasibility of ear-EEG based automatic fitting of hearing aids depends on the ability of the ear-EEG method to detect ASSR thresholds and the variation and offset of ear-EEG thresholds compared with behavioral thresholds. An offset between ASSR and behavioral thresholds is not in itself critical for the application of ear-EEG. However, the size of the offset should be within realistic limits to ensure that the sound stimulation levels needed to perform a hearing test are not unreasonably high. The intrasubject threshold variance should be low to

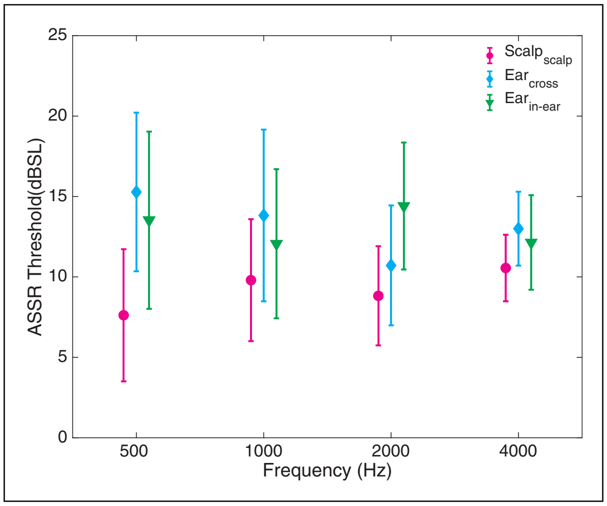

Figure 10. Average ASSRs to 4 kHz chirp stimuli for in-ear and cross-referenced ear-EEG, and scalp EEG for HI (full line) and NH (dotted line) subjects. NH data are adopted from (Christensen, Harte, et al., 2018). Only significant ASSRs are included in the figure. dB SL = dB sensation level; HI = hearing impaired; NH: normal hearing; SPL = sound pressure level.

achieve consistent threshold estimates and thereby precise hearing aid fitting. Finally, the variance of the threshold mean across subjects should be low to have subject-independent fitting paradigms. In this study on subjects with sensorineural hearing loss, the in-ear ear-EEG ASSR thresholds were found to be elevated compared with scalp thresholds, and, consequently, larger correction values would be needed to estimate behavioral thresholds. Nevertheless, in-ear ear-EEG thresholds were observed at an average of 12.1 to 14.4 dB SL, with the upper limit of the confidence intervals being below 21 dB SL at all frequencies (Figure 10). Therefore, the sound intensities needed to perform hearing tests using ear-EEG seem to be realistic. Equally important, the intersubject variation of ear-EEG thresholds was not significantly larger than the variation in behavioral thresholds. This suggests that behavioral thresholds can be estimated from the ear-EEG ASSR thresholds with the same precision as in behavioral tests.

Collectively, the results of this study suggest that ear-EEG-based hearing aid fitting is feasible with regard to the required stimulation level and precision. The detection rate should, however, be improved in a hearing aid application. This can likely be achieved through the development and refinement of a specialized ear-EEG ASSR paradigm. Although the ASSR recording parameters used in the present study were largely adapted from what have been optimized for scalp EEG, the optimal parameters for an ear-EEG-based threshold estimation may be different.

Conclusion

In this study, we investigated the performance of the ear-EEG method to estimate hearing threshold levels of sensorineural hearing-impaired subjects using a comparative setup including ear-EEG, conventional scalp EEG, and behavioral audiometry.

In comparison to scalp EEG, 20% fewer ASSR thresholds were detected with ear-EEG, and ear-EEG thresholds were found to be elevated by 0.8 to 7.5 dB compared with scalp thresholds. However, the variance of the threshold means was not significantly different between ear-EEG and scalp EEG.

Ear-EEG ASSR thresholds could only be obtained in 50% of the cases. The obtained thresholds, however, suggest that ASSR thresholds can be recorded *in-ear* with a reasonable offset relative to behavioral thresholds and with an intersubject variance of ear-EEG thresholds that is not significantly larger than that of the behavioral thresholds.

Collectively, it is concluded that, although further refinement of the method is needed to optimize the threshold detection rate, ear-EEG is a feasible method for hearing threshold level estimation in hearing-impaired subjects.

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Supplemental Material

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