Uncovering auditory evoked potentials from cochlear implant users with independent component analysis

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Abstract

Auditory evoked potentials (AEPs) provide an objective measure of auditory cortical function, but AEPs from cochlear implant (CI) users are contaminated by an electrical artifact. Here, we investigated the effects of electrical artifact attenuation on AEP quality. The ability of independent component analysis (ICA) in attenuating the CI artifact while preserving the AEPs was evaluated. AEPs recovered from CI users were systematically correlated with age, demonstrating that individual differences were well preserved. CI users with high-quality AEPs were characterized by a significantly shorter duration of deafness. Finally, a simulation study revealed very high spatial correlations between original and recovered normal hearing AEPs ($r \geq 0.95$) that were previously contaminated with CI artifacts. The results confirm that after ICA, good quality AEPs can be recovered, facilitating the objective, noninvasive study of auditory cortex function in CI users.

Descriptors: AEP, ICA, Individual differences, Cochlear implant, Brain plasticity

In the last decades, technological developments have made it possible for sensorineural deafness to be reversed by bionic auditory stimulation with a cochlear implant (CI). The CI device mimics the function of the healthy cochlea and delivers electrical stimulation to the auditory nerve, bypassing the damaged inner hair cell, and subsequently to the auditory cortex, allowing the processing of auditory stimuli. Although CIs can bring large benefits to the quality of life of many deafened individuals, the outcome after implantation is variable. Clinical outcome depends on a number of parameters such as duration of deafness, age at implantation, and experience with the implant (Fallon, Irvine, & Shepherd, 2008). It is likely that the degree of auditory cortical rehabilitation also plays an important role (Moore & Shannon, 2009), although this is somehow neglected probably because CIs are unsuitable for functional magnetic resonance imaging. In a few cases, auditory cortical function in CI users has been investigated with magnetoencephalography (e.g., Puntev, Dinnensen, Ross, Wollbrink, & Knief, 2006), but, more frequently, multichannel electroencephalography (EEG) has been used to assess auditory evoked potentials (AEPs) (Debener, Hine, Bleek, & Eyles, 2008; Gilley, Sharma, & Dorman, 2008; Henkin, Tetin-Schneider, Hildesheimer, & Kishon-Rabin, 2009; Sandmann et al., 2009, 2010; Sharma, Dorman, & Spahr, 2002; Zhang, Anderson, Samy, & Houston, 2010). However, the recording of AEPs from CI users presents some challenges. During EEG recordings a large electrical artifact is evoked each time an auditory stimulus is presented, and this artifact obscures the AEPs, as shown in previous studies (Debener et al., 2008; Gilley et al., 2006; Sandmann et al., 2009; Zhang et al., 2010). Accordingly, the time-locked artifact cannot be attenuated by averaging across trials, the conventional procedure for improving the signal-to-noise ratio (SNR) of event-related potentials.

Independent component analysis (ICA) is a linear decomposition method that is typically applied to continuous or concatenated single-trial, multi-channel EEG data (Makeig, Debener, Onton, & Delorme, 2004; Onton, Westerfield, Townsend, & Makeig, 2006). ICA can be used for the reduction of conventional EEG artifacts (Mennes, Wouters, Vanrumste, Lagae, & Stiers, 2010; Viola et al., 2009), and has recently been shown to outperform other correction approaches with regard to eye blink correction (Hoffmann & Falkenstein, 2008), although for other conventional EEG artifacts, such as the myogenic artifact, a much less complete separation between artifact and brain-related activity has been found (McMenamin, Shackman, Greischar, & Davidson, 2010). In principle it should be possible to recover AEPs free from the electrical CI artifact by correction of the artifact independent components (ICs). Indeed, several studies
support this conclusion by reporting that, after ICA-based CI artifact attenuation, typical AEPs could be evaluated in detail (Debener et al., 2008; Gilley et al., 2008; Sandmann et al., 2009, 2010; Zhang et al., 2010, although others have been less successful). Nevertheless, it is still not clear how the ICA approach affects the quality of the resulting AEPs, that is, how sensitive and specific ICA correction for this type of artifact is (Sandmann et al., 2009). Moreover, previous results have been obtained using small samples with similar CI devices, or even single cases (Debener et al., 2008), raising concerns about the generality of the results. Indeed, it is known that CI artifact properties vary across devices and individuals, and also depend on the type of stimulation used (Gilley et al., 2006). This problem is illustrated in Figure 1, which shows electrical artifacts from four different CI users presented with two different environmental sounds. As can be seen, different sounds produce somewhat different artifacts for the four CI users. It is also evident that the same sound can produce different artifacts across individuals even when the same CI devices are used.

The present study sought to investigate the quality of late AEPs recovered from a relatively large sample of CI users (N = 18) using a variety of CI devices stimulated with a variety of environmental sounds and pictures of natural scenes. To explore the ability of ICA to reduce the CI artifact, the degree of CI artifact attenuation was computed. Henceforth, we refer to this procedure as evaluating ICA “sensitivity.” AEPs were also

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**Figure 1.** Cochlear implant (CI) artifact evoked by different environmental sounds. Panels in the top row show the sound profile of two stimuli used in the experiment, on the left side a “fire alarm” sound, on the right side a “hitting baseball bat” sound. The panels in the middle row show the root mean square (RMS) of the EEG activity of two CI users when stimulated with the two sounds during the first 400 ms of presentation, respectively. Both implantees used a Nucleus CI system on the right side with the ESPrit G processor and the ACE speech processing strategy. The topography at the latency of maximum amplitude is illustrated for the same subject in both panels. The panels in the bottom row show RMS activity for another two CI users and the respective topography at the latency of maximum amplitude for one of them. Both implantees used the Advanced Bionics CI system on the left side with the AB Harmony processor and the HiRes-S with Fidelity 120 speech processing strategy. Gray shadow represents the average time window of CI artifact onset.
evaluated with an SNR measure. SNR was compared between CI users and age-matched normal hearing (NH) controls and, for the CI users, SNR was then related to clinical parameters. For both CI users and NH controls, we also evaluated the correlations of AEP amplitude with age. Although exact predictions could not be made, we speculated that, if data quality were sufficient, age-related effects evident in AEPs of NH listeners (Kerr, Rennie, & Robinson, 2011; Schiff et al., 2008) should also be preserved in CI users. In order to evaluate the ability of ICA to preserve the cortical evoked responses in the data, two procedures were applied. First, a hybrid simulation was performed. Here, the CI artifact extracted from CI user datasets using ICA was added to the single-trial EEG data from NH participants. The resulting simulated datasets were then submitted to ICA. The AEPs obtained after ICA-based CI artifact attenuation were then compared with the original NH AEPs. Second, visual evoked potentials (VEPs) from CI users collected in the same task were compared before and after ICA-based CI artifact attenuation. These procedures are henceforth referred to as evaluating ICA “specificity.” Taken together, we expected that this set of analyses would allow us to systematically evaluate the quality of ICA-recovered cortical AEPs from deaf individuals using a CI.

Methods

Participants

Eighteen postlingually deafened cochlear implant (CI) users (10 females, $M = 59.89$, $SD = 13.06$ years) and 18 age and gender matched NH participants (10 females, $M = 55.17$, $SD = 12.31$ years) took part in the study after giving written informed consent. NH participants in the control group were screened for normal hearing, defined as pure-tone thresholds better than 20 dB hearing level (HL) at 0.5, 1, 2, and 4 kHz. Eight NH participants were aged between 30 and 50 years and had pure-tone thresholds $\leq 20$ dB HL for all frequencies tested. For the ten older NH participants, higher thresholds for the higher frequencies (2 kHz, left ear [LE]; $M = 19.50$, $SD = 10.66$ dB, right ear [RE]; $M = 18.50$, $SD = 16.33$ dB; 4 kHz, LE: $M = 41.00$, $SD = 20.39$ dB, RE: $M = 35.50$, $SD = 30.86$ dB) were accepted, reflecting natural age-related hearing loss (Humes, Kewley-Port, Fogerty, & Kinney, 2010). All CI users were implanted unilaterally except one. During the task, only the first implanted CI was switched on for this user (Table 1). The CI users were recruited from the South of England Cochlear Implant Centre in Southampton, UK. The clinical profile of each CI user is presented in Table 1. All 36 participants were right-handed and had no history of neurological or psychiatric disorders and had normal or corrected-to-normal vision. Procedures were approved by the local ethics committee and conformed to the Declaration of Helsinki.

Table 1. Cochlear Implant Users’ Clinical Profile

<table>
<thead>
<tr>
<th>CI user</th>
<th>Age (years)</th>
<th>Gender</th>
<th>CI side</th>
<th>Duration deafness (months)</th>
<th>Age implantation (months)</th>
<th>Device</th>
<th>Processor</th>
<th>Sound coding strategy</th>
<th>CI use (months)</th>
<th>Score (% correct)</th>
<th>Sound level (dB SPL)</th>
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<tr>
<td>01</td>
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<td>Right</td>
<td>150</td>
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<td>627</td>
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<td>22</td>
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Notes. Device names and processor names according to manufacturers’ labeling. Sound coding strategies: ACE = advanced combination encoders; FSP = fine structure processing; HiRes-S = high resolution with fidelity 120; SPEAK = spectral peak coding. Score corresponds to the percentage correct on the Bemford-Kowal-Bench (BKB) speech recognition test in quiet. Sound level in dB SPL chosen by each participant for the presentation of auditory stimuli.†Subject was implanted bilaterally, but during the recording session only the implant on the left side was switched on.
ural objects from the same normative study. The stimuli were presented centrally for 800 ms, with the visual stimuli covering an angle of 8.6° vertically and horizontally. In visual blocks (see Experimental Design and Task), trials included a gray square, which was presented centrally for 800 ms subtending a visual angle of 1.9°. Stimuli were presented using a 23-in. monitor (1280 × 1024 × 32-bit color, 60 Hz refresh rate). Screen background was black at all times. All stimuli were presented using Presentation 10.0 software (Neurobehavioral Systems).

Experimental Design and Task

An adapted audiovisual semantic priming paradigm (Schneider, Debener, Oostenveld, & Engel, 2008; Schneider, Engel, & Debener, 2008) was used. Eighty environmental (auditory primes) and 160 degraded pictures of natural objects (visual targets) were included (taken from www.multimost.com). Each visual target was paired with either a congruent or an incongruent sound, constituting the audiovisual (AV) block, or with a gray square (temporal cue) in a visual only (V) block. The order of the pairs was pseudorandomized to ensure that response type and object category were distributed equally over the course of stimulus presentation. Each trial started with a fixation cross (500 ms), followed by the audio prime (800 ms) or by the square (800 ms). The fixation cross remained on screen until the visual target was presented (800 ms). After the target offset, the participants had a maximum of 2900 ms to indicate a response using a keypad. The next trial started 1000 ms after the response. Each participant was presented with a total of four blocks (two V and two AV) and a break of 1 min was included between blocks. The order of presentation was counterbalanced across participants. The task consisted of judging whether the objects shown in the degraded pictures (visual targets) would fit ("Yes") or not ("No") into a box (dimensions 33.5 × 24.0 × 12.5 cm) shown in the lab. Participants were comfortably seated at a distance of 150 cm from the screen, and all responses were made using a keypad with two keys, marked with "Yes" and "No," respectively (Cedrus RB series, San Pedro, CA, USA). The assignment of the keys was counterbalanced across participants. Note that the behavioral results and the relationship between hearing loss, semantic priming, and event-related EEG are not reported here and will be presented elsewhere.

EEG Recording

Participants were seated in an electrically shielded, sound attenuated, and dimly lit booth (Industrial Acoustics, Winchester, UK). EEG data were recorded from 68 channels using a high-input impedance amplifier system (Compumedics Neuroscan, Charlotte, NC, USA) and a customized electrode cap (Easycap, Herrsching, Germany) specifically designed to improve spatial sampling and facilitate AEP source localization (Hine & Debener, 2007; Hine, Thornton, Davis, & Debener, 2008). The cap was fitted with 66 Ag/AgCl electrodes in an equidistant layout that spans a larger part of the head sphere than standard 10–20 montages. Two additional electrodes were placed below the eyes. For the CI users, EEGs from some electrodes (M = 3.94, SD = 0.94 electrodes, range 2–6 electrodes) could not be recorded due to the location of the CI device (i.e., transmitter-receiver coil, cable to processor, processor). Data were recorded with a sampling rate of 1000 Hz using the nose-tip as reference, and were analogue filtered between 0.1 and 200 Hz. Electrode impedances were maintained below 20 kΩ prior to data acquisition.

Data Processing

EEG data were processed using custom scripts and EEGLAB (Delorme & Makeig, 2004) running in the MATLAB (Mathworks, Natick, MA) environment. Data were offline filtered from 1 to 40 Hz using windowed sinc FIR filters with a Hann window (taken from the FIRfilter plugin for EEGLAB developed by A. Widmann: www.uni-leipzig.de/~biocog/content/widmann/eeeglab-plugins/). Data were then down-sampled to 500 Hz and pruned of unique, non-stereotyped artifacts using the EEGLAB function jointprob.m (Delorme, Sejnowski, & Makeig, 2007). Extended infomax ICA as implemented in EEGLAB was then applied to the remaining data in order to achieve a reliable decomposition (Debener, Thorne, Schneider, & Viola, 2010). Independent components (ICs) representing eye-blinks and electrocardiograph (ECG) artifacts were semiautomatically identified using CORRMAP (Viola et al., 2009) and then corrected from all datasets. These ICs are labeled as conventional artifacts. For the CI users a second, additional step was included. Here, the properties of the remaining ICs were visually inspected to identify those representing the CI artifact. The same features (IC maps showing a centroid on the side of the implanted device; IC activations matching the onset/offset of acoustic stimulation) were used as reported previously (Debener et al., 2008; Gilley et al., 2006; Sandmann et al., 2009). After component identification, CI artifact ICs were corrected from the CI users' datasets. Afterwards all datasets were segmented into epochs from −200 to 600 ms relative to sound onset. AEPs were then obtained by time-domain averaging. A baseline correction was performed using the prestimulus interval (i.e., −200 to 0 ms) and AEPs at missing electrodes were interpolated with the EEGLAB function eeg_interp.m, which implements a smoothed inverse distance approach. AEP amplitude and latency analyses were performed for the fronto-central electrode with the largest grand average amplitude for both groups (approximately FCz). AEP peak amplitudes and latencies were determined using a semiautomatic procedure as implemented in peakdet.m (www.bilauer.co.il/peakdet.html).

ICA “sensitivity”. Previous reports differ regarding the incidence and magnitude of the CI artifact (Zhang et al., 2010). The sensitivity of the ICA approach was evaluated by calculating the individual CI artifact attenuation rate. This rate was calculated as the difference between the mean root mean square (RMS) for the original data and the mean RMS for the corrected data for bins of 50 ms (−200 to 400 ms). This procedure allowed evaluation of the attenuation rate at critical latencies, such as the onset (0–50 ms) of the artifact and at latencies of the N1 and P2 components (100–250 ms).

AEP quality. An SNR measure was calculated to assess the quality of the AEPs. Traditionally the SNR is defined as the power of the signal at a peak latency of interest divided by the mean power of the activity in the baseline period. Here, a more conservative approach was chosen that allowed signal and noise to be estimated at the same, post-stimulus onset latency range (Schimmel, 1967). The plus-minus procedure consists of averaging all trials after polarity reversal of every other trial. The remaining average is an estimation of noise, since all time-locked features, including both AEPs and the CI artifact, can be assumed to sum to zero due to the artificial polarity reversal. In this study, the focus was on the N1 component.
of the AEP, since it is typically the largest response and is easily identified in single subjects. The N1 SNR was calculated by dividing the RMS of the N1 component at peak latency ± 10 ms by the RMS of the estimated noise in the same time window, scaled to dB (20log10(signal/noise)).

In addition, AEPs were further evaluated by comparing peak amplitudes with demographic and clinical scores. It is well known that AEPs change due to aging (Kerr et al., 2011; Schiff et al., 2008). There is a considerable range in age at implantation, and thus a wide age range in the sample of CI users studied. We therefore reasoned that, if individual differences in AEPs are retained, N1-P2 amplitudes should show a systematic correlation with age. A positive outcome would clearly suggest that individual differences in AEPs are not erased by ICA-based CI-artifact attenuation.

ICA “specificity”. In order to evaluate the effect of ICA-based artifact attenuation on the reconstruction of the AEPs, a hybrid simulation was performed. Here, a CI artifact template was added to an original NH subject dataset with a known AEP response, and the modified dataset was then processed using the same procedure as for the CI dataset. This procedure provided a comparison between the “true” AEP and the post-ICA recovered AEP, and was therefore informative about the specificity of the ICA approach. In the semantic priming paradigm employed here, 80 different environmental sounds were presented using two different sequences. The examples in Figure 1 show that different sounds produce different artifacts. Taking this into account, two CI artifact templates matching each presentation sequence were created. In order to produce the two templates, data from two CI users (one for each sequence) were randomly chosen. For each CI user, all ICs except those labeled as CI artifact related were removed from the original data. In one case, six CI artifact ICs were kept in the original data and in the other, three. This procedure resulted in two CI template datasets where only the CI artifact related activity was kept and all other sources of activity were removed. We preserved the single-trial artifact responses in the simulation by adding the respective single-trial template dataset to the single-trial NH datasets. In order to evaluate ICA specificity, spatial correlations between the “true” and the recovered AEPs were computed (Sandmann et al., 2009), thus providing information about the amount of variance in the data that was unaffected.

A further, less direct approach for the assessment of ICA specificity was performed by the evaluation of portions of EEG data that were not primarily related to auditory processing. Here, the reasoning is that these portions of the data should not be substantially altered by the removal of ICs supposedly reflecting the CI artifact. Following this line of reasoning, we analyzed the VEPs from the CI users, of considerable interest in the context of cross-modal compensatory plasticity (Doucet, Bergeron, Lassonde, Ferron, & Lepore, 2006). VEPs were computed for all blocks after the removal of both conventional and CI artifacts with ICA. The similarities between the original VEPs and the CI-corrected VEPs were assessed as described for the simulation study.

![Figure 2. Box plots showing median cochlear artifact (CI) attenuation rate (horizontal line in the middle) across 18 CI users for bins of 50 ms (range: -200–400 ms). Bottom and top of each “box” are the 25th and 75th percentiles of the samples, respectively. Whiskers are drawn from the end of the interquartile range to the furthest observation (cf. Matlab function boxplot.m). Attenuation rate was calculated as the difference between mean RMS for original data and mean RMS for corrected data for each bin.](image-url)

**Statistical Analysis**

All variables were tested for normality using Shapiro-Wilk tests. Comparisons between CI users and NH controls in terms of number of ICs corrected, number of trials included in AEP analysis, AEP N1-P2 peak to peak amplitude, and N1 and P2 peak latencies were evaluated using two-tailed independent t tests or, when normality criteria were not met, Mann-Whitney U tests. For four out of the eighteen CI users, it was not possible to identify a P2 component in the AEPs. For these participants, P2 amplitude was taken by 0 μV and P2 latency as the mean value of the other CI users (245 ms). For the hybrid simulation, comparisons between NH original and post-ICA simulated datasets in terms of number of ICs corrected, number of trials included in AEP analysis, N1-P2 peak to peak amplitude, and N1 and P2 latencies were evaluated using two-tailed paired t tests or, for the variables where the normality test failed, Wilcoxon signed-rank tests. The spatial correlation between original and recovered AEPs was computed for all time frames, in order to investigate topographical similarities. The mean correlation across NH participants was calculated, after first applying Fisher’s Z transformation. VEP differences for P1, N1, and P2 amplitudes were also
assessed using either two-tailed paired $t$ tests or Wilcoxon signed-rank tests, and spatial correlations between original and CI-corrected datasets were calculated as described above. For all tests, differences were considered significant when $p < .05$. For all parametric tests, the effect size was calculated using Cohen’s $d$ estimate (Cohen, 1988). For all non-parametric tests, the effect size estimate $r$ was computed (Rosenthal, 1991). Parametric Pearson correlation coefficients are indicated with $r_P$ and Spearman rank correlations with $r_S$, to avoid confusion.

**Results**

**ICA Sensitivity**

AEPs from all CI users were buried in large electrical CI artifacts. This was evident from evaluation of CI-uncorrected AEPs, which showed large, not biologically plausible deflections in the AEP response interval. Figure 2 shows box plots for the median attenuation rate across the 18 CI users for 50 ms bins from $-200$ to $400$ ms. As can be seen, substantial individual differences were observed in the amount of attenuation, largely reflecting large individual differences in the magnitude of the artifact (range 12 to 760 $\mu$V). The attenuation was largest for the 50–100 ms bin where the median RMS difference was $>15$ $\mu$V. This is the typical morphology, characterized by a sharp onset followed by a “pedestal” (Debener et al., 2008; Gilley et al., 2006; Sandmann et al., 2009). Only after CI artifact attenuation was it possible to recover AEPs for CI users that resembled in morphology and topography those that can be observed in NH subjects.

**AEP Quality**

No differences were found between CI users and NH participants in terms of the number of trials included in the AEP calculation (Mann-Whitney, $U = 204.00$, n.s.). The number of rejected ICs representing conventional artifacts was not significantly different between NH participants and CI users (Mann-Whitney, $U = 154.5$, n.s.). When both conventional and CI artifacts were accounted for, the total number of rejected ICs was significantly larger for CI users compared with NH participants.

![Figure 2.](image2.png)

**Figure 2.** Auditory evoked potentials (AEPs) at a frontal-central electrode for all 36 participants. Each column shows the AEPs sorted by N1 signal to noise (SNR), for both normal hearing (NH) participants (left) and cochlear implant (CI) users (right). On the bottom of each column the grand average AEP, the median SNR, and the range in dB are shown. For illustrative purposes, all AEPs were normalized by the single subject RMS of the noise in the latency range of the N1 component. Hence values are expressed in arbitrary units (a.u.).
Uncovering AEPs from CI users

Whitney, U = 293.00, p < .001, r = .12). Table 2 presents a summary of these comparisons.

No significant differences were found when comparing SNR values in the N1 interval in NH participants (\(Mdn = 25.19 \text{ dB}\)) and in CI users (\(Mdn = 17.93 \text{ dB}\)) (Mann-Whitney, U = 120.00, n.s.). Figure 3 shows the single-subject AEPs for all NH participants and CI users, sorted by SNR in descending order and divided into high and low groups using a median split. AEPs were first divided by the mean noise RMS for better visualization and comparison. Although the SNR did not significantly differ between groups, the range was larger for the CI users (SNR range, NH: 15.51 to 32.71 dB; CI: 9.18 to 35.23 dB). It is worth noting that, for the large SNR groups, the largest CI SNR was 35.23 dB, while the largest NH SNR was slightly smaller (32.71 dB), which indicates that for some CI users the recovered AEPs had very robust N1 responses. For all NH datasets, the N1 and P2 components could be easily identified, even for the participant with the smallest SNR (15.51 dB). For all CI users the N1 AEP response was identified, but for four CI users the P2 response was difficult to discern, if not completely absent. CI users showed a significantly reduced N1-P2 peak-to-peak amplitude (\(r = -0.53, p < .001, d = 1.76\)). The N1 peak latency was significantly delayed for CI users (\(r = 4.18, p < .001, d = 1.43\)), as was P2 peak latency (\(r = 2.16, p = .038, d = .74\)). Table 2 presents a summary of these comparisons.

The correlation between age and N1-P2 peak-to-peak amplitude is shown in Figure 4. As expected, a significant negative correlation between age and peak-to-peak amplitude was found for the NH participants (\(r = -0.56, p = .015\)). For the CI users a similar pattern was found (\(r = -0.70, p = .001\)), indicating that individual differences were preserved after CI artifact attenuation. In order to further evaluate the quality of the recovered AEPs, the CI group was split into high and low SNR subgroups based on the SNR median (17.93 dB). Note that the age for CI users in the two subgroups was not significantly different (HIGH: \(Mdn = 50.89, SD = 14.79\) years, LOW: \(Mdn = 57.89, SD = 14.79\) years; \(t[16] = 30, n.s.\)). CI users with low SNR values had been deaf for a significantly longer period (HIGH: \(M = 287.56, SD = 170.46\) months, LOW: \(M = 519.67, SD = 109.59\) months; \(t[16] = 2.88, p < .001, d = 1.94\)), but no significant differences were found when comparing age at implantation (HIGH: \(M = 600.89, SD = 157.79\) months, LOW: \(M = 650.78, SD = 163.33\) months; \(t[16] = .53, n.s.\)), as shown in Figure 5. The period of CI experience was not significantly different for the two subgroups (HIGH: \(Mdn = 28.00\) months, LOW: \(Mdn = 28.00\) months; \(U = 43.50, n.s.\)). The score in the speech recognition test was not significantly different among CI users with low and high SNR (HIGH: \(Mdn = 93.00\%\) correct, LOW: \(Mdn = 83.00\%\) correct, \(U = 24.00, n.s.\)).

ICA Specificity

After ICA-based artifact attenuation, no differences were found between post-ICA simulated and original NH datasets in terms of the number of trials included in the AEP calculation (Wilcoxon, \(T = 34.50, n.s.\)). Fewer conventional artifact ICs were rejected for the post-ICA simulated data (Wilcoxon \(T = 0.00, p = .02, r = -.55\)). No significant differences were found for the N1 peak latency (\(t[17] = -1.70, n.s.\)) and P2 peak latency (\(t[17] = -1.62, n.s.\)). The N1-P2 peak-to-peak amplitude was smaller for simulated data (\(t[17] = 3.23, p = .005, d = 1.57\)). Although the amplitude was slightly altered, the correlations between original and post-ICA simulated amplitudes were very high (\(r_p = .99, p < .001\)). Table 2 presents a summary of these comparisons. Moreover, the spatial correlation was generally high for all time frames (range .82 to .99), indicating that topographies were not substantially altered by CI attenuation. Figure 6 shows the grand average results across the 18 pre-ICA corrected simulated NH datasets. As shown, the artifact completely obscured the AEP response. The comparison between original and post-ICA simulated grand average AEPs is also shown, as well as the grand average spatial correlation for all time frames. Only for latencies between 0 and 50 ms were the correlations below \(r_p < .90\). This interval corresponds to the onset of the artifact and visual inspection of single-subject data revealed that for some datasets this interval was still contaminated with residual CI artifact. In contrast, between the latencies of the N1 and the P2 components (100 to 300 ms), the correlation was above \(r_p = .95\), indicating that over 90% of the variance was left unaltered by the CI artifact attenuation.

When a similar comparison was performed for original (ORIG) and CI-corrected (CORR) CI users’ VEPs, no differences were found between peak latencies for the P1 (\(t[17] = 1.08, n.s.\)), the N1 (\(t[17] = .49, n.s.\)) or the P2 responses (\(t[17] = .46, n.s.\)). Similarly, no differences were found between the P1 (\(t[17] = 1.52, n.s.\)) and CI-corrected (CORR) CI users’ VEPs.
n.s.), the N1 ($t_{17} = 1.04, \text{n.s.}$) or the P2 peaks ($t_{17} = 1.44, \text{n.s.}$).

Table 3 presents a summary of these comparisons. When assessing the topographical similarity, a high spatial correlation (range .91 to .99) was also found for all time frames. Figure 7 shows a summary of the main VEP results. The RMS amplitude across channels of 18 CI users for original (blue) and CI-corrected (red) datasets is shown, as well as the topographies at P1, N1, and P2 peak latencies. It is clear that the differences between original and CI-corrected VEPs were minimal.

**Discussion**

This study evaluated the quality of AEPs from a large and heterogeneous population of post-lingually deafened CI users. The ICA-based CI artifact attenuation was evaluated in terms of both its sensitivity and specificity. Several previous studies have used ICA to attenuate the CI artifact (Debener et al., 2008; Gilley et al., 2006, 2008; Sandmann et al., 2009, 2010; Zhang et al., 2010), but negative findings exist (Martin, 2007). Overall, our results confirm the value of ICA in the context of AEPs from CI users, which could become an important tool for guiding further implant technology (Moore & Shannon, 2009).

After a careful manual screening of all independent components (ICs), a group of ICs with the characteristics associated with the CI artifact could be found, similar to previous reports from our group and others (Debener et al., 2008; Gilley et al., 2006; Sandmann et al., 2009). However, the identification of ICs reflecting the CI artifact is difficult for several reasons. First, in contrast to the ICA representation of other more conventional artifacts, which are normally represented by one or very few ICs, electrical artifacts from the implant device were represented by up to 11 ICs in this study. In only two out of 18 datasets was the CI artifact represented by a single IC. This raises issues about the subjectivity of the component selection procedure. Second, ICs representing conventional artifacts can be efficiently and semiautomatically identified using topographical information (Viola et al., 2009). Unfortunately, the automatic identification of ICs reflecting the CI artifact does not seem possible based on topographical information alone and currently requires laborious visual inspection of additional features in the component activation patterns. This manual visual screening process can be seen as a major drawback of the ICA procedure, since it is subjective, time consuming, and error prone and also requires expert knowledge. To overcome this limitation, an important goal for the future is to develop software solutions that automatically identify and select components representing the CI artifact. This is a worthwhile goal, since AEPs can objectively inform about auditory cortex function and thus help to guide further improvements in implant technology (Moore & Shannon, 2009).

The sensitivity of ICA in attenuating the CI artifact was high in our study, since the majority of recovered AEPs showed little CI artifact, and were comparable with NH AEPs in terms of SNR and individual differences, such as aging effects. Another advantage of the ICA procedure is the potential for using almost the entire recorded data for the AEP calculations. For instance, in this experiment on average 95% of the original epochs were retained. Previous studies rejected trials contaminated both with ocular and with CI artifacts (Henkin et al., 2009; Sharma et al., 2002), thereby reducing the potential for obtaining good AEP SNR. More importantly, ICA solves the problem discussed in other studies of having to use very short auditory stimulation to avoid overlap of the CI artifact with longer latency AEP components (Gilley et al., 2006; Henkin et al., 2009). Our study shows that good quality AEPs can be acquired using complex sounds with long durations (800 ms). This opens up the possibility of using stimuli with high frequency resolution, complexity, and ecological validity, such as speech or musical sounds.
Previous studies have already reported the high ICA sensitivity shown here, but an evaluation of ICA specificity has not been performed. One possible reason is that specificity issues can be examined only with a rather indirect approach. Since AEPs cannot be obtained from CI users without the CI artifact, it is necessary to use a simulation approach in which the true degree of mixing of signal and artifact is known. One simple yet realistic hybrid simulation approach consists of adding an averaged CI artifact template to the AEPs from NH participants and then correcting the data with ICA. The comparison of original and recovered AEPs then provides information about ICA specificity. In this study, an improvement was made to this simulation approach by matching artifact templates to the individual auditory stimuli. A high spatial correlation between original and post-ICA simulated AEPs was found, indicating a high similarity in topography between the two, as previously reported for single-subject cases (Sandmann et al., 2009). In our study, we observed in some datasets a small residual CI artifact at onset latencies. However, this residual artifact, which demonstrates that the ICA approach is not perfect, was restricted to latencies before the early P1 AEP component. The P1 component has been used as the main marker in the investigation of auditory cortical maturation in implanted children (Gilley et al., 2008; Sharma et al., 2002), but has a longer latency in this population. Our results also showed that N1 and P2 latencies were not affected by the correction procedure, while the amplitudes were slightly reduced. However, the correlation between original and corrected amplitudes was very high, confirming that individual differences in amplitudes were preserved and ICA overattenuation affected all datasets similarly. Moreover, similar results were obtained when

![Figure 6. Evaluation of ICA specificity by simulating cochlear implant (CI) artifacts in normal hearing (NH) participant datasets. (A) Grand average CI artifact for 18 NH datasets after addition of CI artifact templates for the electrode highlighted. Topographies at mean N1 and P2 peak latencies are shown. (B) Original grand average auditory evoked potentials (AEPs) (blue) and reconstructed grand average AEPs after ICA-based CI artifact attenuation (red) for the highlighted electrode. Topographies at N1 and P2 peak latencies are shown for the original and the reconstructed data, respectively. (C) Topographical similarity assessed using the mean spatial correlation between original and reconstructed datasets at all time frames. Gray shadow represents the average time window of CI artifact onset.](image)

![Figure 7. Evaluation of ICA specificity by comparing visual evoked potentials (VEPs) from cochlear implant (CI) users with and without ICA-based CI artifact attenuation. Left, root mean square (RMS) amplitudes across all channels for original (blue) and for corrected (red) VEPs after ICA-based CI artifact attenuation. Right, topographies at mean P1, N1, and P2 peak latencies are shown for the original and corrected data, respectively.](image)

<table>
<thead>
<tr>
<th>Group</th>
<th>RMS VEP latency [ms]</th>
<th>RMS VEP amplitude [μV]</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>P1</td>
<td>N1</td>
</tr>
<tr>
<td>CI</td>
<td>117.6 ± 12.2</td>
<td>167.9 ± 14.0</td>
</tr>
<tr>
<td>CI corr</td>
<td>119.7 ± 10.5</td>
<td>168.1 ± 13.7</td>
</tr>
</tbody>
</table>

**Notes.** CI = cochlear implant users’ original datasets; CI corr = ICA-based corrected datasets; RMS = root mean square; VEP = visual evoked potential.
comparing original and CI-corrected VEPs. The VEP P1, N1, and P2 peak amplitudes and latencies were similar for original and CI-corrected data, and the topographical similarity at peak latencies of these components was also high. Thus, the combination of simulated NH AEPs results and the analysis of VEPs from CI users provides strong evidence of high specificity. We conclude that high-quality AEPs can be recovered from CI users.

When evaluating the quality of the AEPs from CI users, a large variability in terms of component morphology, amplitude, and latency was found. Since the sample tested had a considerable age variance and also varied in clinical parameters, these differences were to be expected. Nevertheless, more than half of the sample of CI users was characterized by AEPs with an SNR generally similar to that of age-matched NH individuals. Although the CI users had significantly smaller N1 and P2 amplitudes, individual features of the population such as aging effects that are evident in NH participants were preserved in the recovered AEPs. This result strongly suggests that the ICA-based CI artifact attenuation procedure did not eliminate individual differences. The subgroup analysis based on SNR values revealed an interesting clinical result. It was found that the CI users with larger SNR, that is, more robust N1 responses, had been deaf for significantly shorter periods. Similar findings have been previously reported, where duration of deafness has been linked to both auditory cortical activations as revealed by functional neuroimaging (Green, Julyan, Hastings, & Ramsden, 2005) and mismatched negativity amplitudes (Sandmann et al., 2010).

Taken together, these findings imply that the longer the auditory cortex was deprived of sensory input, the more difficult auditory rehabilitation with a CI becomes.

The duration of deafness, therefore, seems to be a key clinical parameter. However, onset of deafness is not always easy to determine, as hearing loss can be progressive with a formal diagnosis only at later stages. In this study, the duration of deafness was established using the information provided by CI users at interviews done prior to EEG recording. One neuroimaging study (Green et al., 2005) and another EEG study (Kelly, Purdy, & Thorne, 2005) found a correlation between duration of deafness and scores in speech recognition. Speech scores have been used as the main evaluation measurement of rehabilitation in clinical settings. However, previous studies reported that an isolated speech recognition score can be misleading and therefore recommended that the rehabilitation assessment should be done using a combination of tests (Donaldson et al., 2009; Gifford, Shallop, & Peterson, 2008). Other authors have supported the use of AEPs as a complementary assessment tool (Kileny, 2007; McNeill, Sharma, & Purdy, 2009). In some cases, AEPs may even replace such tests, for instance, when dealing with prelingually deafened CI users (McNeill et al., 2009) or paediatric populations (Kileny, 2007). In any case, making the evaluation of CI users' AEPs easy and reliable seems to be a key factor for a better monitoring of implantation outcome in the future. The available evidence from longitudinal AEP studies is based on a few single cases (Pantev et al., 2006) but clearly demonstrates that AEPs can provide important insights into auditory rehabilitation.

In conclusion, the results presented here complement previous studies showing that ICA can successfully attenuate the electrical CI artifact in EEG data from CI users, thus allowing the recovery of AEPs. ICA is an adequate tool for this task, demonstrating good sensitivity and specificity. In addition, the recovered AEPs from CI users reflected the expected correlations with aging and clinical parameters. Overcoming practical limitations of component selection would help to establish multichannel AEPs in response to speech and musical sounds as a useful research tool. Speech-evoked AEPs could provide an important tool for the objective monitoring of auditory cortical rehabilitation after implantation, and appear particularly promising in cases where conventional measures of speech perception and auditory performance cannot be used. A complementary use of objective measurements of auditory cortex function may help to shape rehabilitation programs and thus improve the quality of life for CI users.

References


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