Epidural recordings in cochlear implant users

S Haumann1,3,4, G Bauernfeind1,3,4, M J Teschner1,3, I Schierholz1,3, M G Bleichner2,3, A Büchner1,3 and T Lenarz1,3

1 Department of Otolaryngology, Hannover Medical School, Hannover, Germany
2 Department of Psychology, University of Oldenburg, Oldenburg, Germany
3 Cluster of Excellence ’Hearing4all’, Hannover & Oldenburg, Germany

E-mail: haumann.sabine@mh-hannover.de

Received 19 October 2018, revised 10 April 2019
Accepted for publication 1 May 2019
Published 30 July 2019

Abstract
Objective. In the long term it is desirable for CI users to control their device via brain signals. A possible strategy is the use of auditory evoked potentials (AEPs). Several studies have shown the suitability of auditory paradigms for such an approach. However, these investigations are based on non-invasive recordings. When thinking about everyday life applications, it would be more convenient to use implanted electrodes for signal acquisition. Ideally, the electrodes would be directly integrated into the CI. Further it is to be expected that invasively recorded signals have higher signal quality and are less affected by artifacts. Approach. In this project we investigated the feasibility of implanting epidural electrodes temporarily during CI surgery and the possibility to record AEPs in the course of several days after implantation. Intraoperatively, auditory brainstem responses were recorded, whereas various kinds of AEPs were recorded postoperatively. After a few days the epidural electrodes were removed. Main results. Data sets of ten subjects were obtained. Invasively recorded potentials were compared subjectively and objectively to clinical standard recordings using surface electrodes. Especially the cortical evoked response audiometry depicted clearer N1 waves for the epidural electrodes which were also visible at lower stimulation intensities compared to scalp electrodes. Furthermore the signal was less disturbed by artifacts. The objective quality measure (based on data sets of six patients) showed a significant better signal quality for the epidural compared to the scalp recordings. Significance. Altogether the approach revealed to be feasible and well tolerated by the patients. The epidural recordings showed a clearly better signal quality than the scalp recordings with AEPs being clearer recognizable. The results of the present study suggest that including epidural recording electrodes in future CI systems will improve the everyday life applicability of auditory closed loop systems for CI subjects.

Keywords: auditory evoked potentials (AEPs), auditory cortex (AC), epidural recordings, cochlear implant (CI)

(Some figures may appear in colour only in the online journal)
1. Introduction

Cochlear implants (CIs) are an established treatment for severe to profound deafness for more than 30 years [1–4]. A CI can restore the hearing ability to some extent by mimicking the function of the inner ear. A CI system consists of an external and an internal part. The internal part is implanted into the ear. It comprises the receiver/stimulator placed in the temporal bone and an electrode array inserted into the cochlea. The external part subdivides into a speech processor and a transmitter coil which is fixed by a magnet onto the skin of the patient.

While early CI systems at the beginning of the 1980s could only provide some assistance for lip-reading, today worldwide candidates can expect open speech perception in the majority of cases [4–10]. Even in more sophisticated tasks like speech perception in noise [4, 6–8, 11] as well as music perception [12–15] or speaker discrimination [16] CI subjects can expect perception in noise [4, 6–10]. Even in more sophisticated tasks like speech perception in noise [4, 6–10].

A BCI system provides users with the possibility to communicate or interact through thought/brain activation processes alone. The technology provides the user with an auditory evoked potentials (AEPs) might be used in the context of brain–computer interface (BCI) technology.

A BCI system provides users with the possibility to communicate or interact through thought/brain activation processes alone. The technology provides the user with an artificial output channel that utilizes the information from neuronal activity of the brain and does not rely on the normal output pathways of peripheral nerves and muscles (for an overview see [17]).

Brain signals like AEPs can be recorded using either surface or invasive electrodes. Late AEPs are mainly generated in the auditory cortex thus electrodes optimally should be placed over these brain areas. BCIs using implanted electrodes do have a better signal to noise ratio, and could become integrated parts of CIs, without the need of additional devices. The site of implantation could either be subdermal [18], intracranial [19–23] (either epidurally or subdurally, whereas the difference of these placing levels was found to be negligible [24, 25]) or even intra-cortically [26, 27]. The best signal to noise ratio is expected for intracranial and intra-cortical electrodes.

In the future a wide array of applications will use brain signals as an important source of information and possible application scenarios are diverse [28]. A trend setting approach in this regard is the continuous monitoring of auditory neural responses in hearing devices, especially in CIs. Therewith the fitting could be adapted accordingly, without the user having to exert active control (see appendix D in [28]). The idea is to use late AEPs for parameter optimization in the CI fitting procedure. In our EEG laboratories, we currently work on using an automated P300 detection in the context of frequency discrimination in CI users. Using an oddball paradigm, the detection of a sound deviating for instance in frequency from a standard sound can be objectively classified by looking at the P300 response (see Finke et al [29]). We currently use the technique to evaluate different pre-set frequency maps and intracochlear electrode sets in CI subjects, to objectively determine the best map regarding frequency discrimination. While these laboratory experiments are only the beginning, one can imagine that the integration of an EEG system into a CI (using epidural electrodes for example) gives lots of opportunities for the design of closed loop fitting approaches with a CI. But also other applications in hearing devices might be possible, as BCIs could provide an active and passive control over the hearing devices by closing the auditory loop in everyday life listening (closed loop CI system).

Most strategies, so far, use EEG electrodes mounted on the users scalp. This approach is not favorable for use in daily life situations, as it is clearly visible, uncomfortable and prone to artifacts [31]. Accordingly, recent efforts aimed at reducing discomfort and improving wearability of the EEG sensors [32]. Here, implantable electrodes might be the more convenient solution, especially if the electrodes are integrated into the CI, making no additional surgical intervention for the implantation of the electrodes necessary.

First invasive recordings during CI surgeries were already performed in the early years of CIs [33, 34]. Intraoperative studies like these provided deep insights into the mechanisms of peripheral stimulation of the auditory system and therewith allowed for large developments in the field of CIs and auditory brainstem implants (ABIs). However, under the influence of anesthesia only ABRs can be recorded reliably. Thus, these intraoperative studies were limited to ABR recordings, whereas for the cognitive control of the CI cortical potentials are needed. These potentials however can only be reliably recorded when the patient is awake [35]. Also other invasive recordings of cortical activity in response to CI stimulation have been performed. A single case study in a bilateral CI patient undergoing an epilepsy surgery showed that auditory event related potentials could be reliably recorded over the posterolateral superior temporal sulcus, using a full electrode grid [36].

For an integrated closed loop CI system, however, the recording electrode has to be included in the CI. A first approach was performed by using the extracochlear reference electrode of the CI for the recording of AEPs [37]. Here the signal was recorded telemetrically by using the CI signal processing. Besides the intracochlear electrodes, some CI systems use an external reference electrode, which is placed under the skin close to the temporal lobe which is a promising recording position. Other CI systems only use one reference electrode, which is integrated into the housing. However, these internal recording implementations are optimized for short latency potentials (electrically evoked compound action potentials, ECAPs) with short recording time windows of 1.6 ms to a maximum of 50 ms depending on the brand, whereas cortical potentials require time windows of up to 500 ms. In general this can be handled by recording the same signal multiple times with shifted time windows and recomposing the signal [38]. However, this approach is very time consuming and not feasible for an online application.

Thus, todays CI systems have to be developed further, integrating an additional electrode that is positioned at a location that enables it to record with an adequate time window. In the meantime, studies have to rely on additional recording
electrodes that are connected to conventional EEG amplifiers to evaluate feasibility and efficacy of the applied method of cortical AEP (EEG) recording.

Recently we started a first feasibility study [39, 40] to investigate the possibilities of recording auditory induced neuronal activity via epidural electrodes in CI users. We used a clinical AEP recording device, which allowed for quick and reliable recordings. The aim of this work was to assess the feasibility of epidural electrodes to record AEPs in CI users and to evaluate their performance in comparison to standard surface electrodes. Results might provide ideas on how to design auditory closed loop systems more suitable for everyday life application and help to optimize signal quality and hence function of auditory BCIs. In more detail we aimed to investigate the feasibility of invasive AEP recordings in CI users with a restricted number of recording channels. Furthermore we wanted to investigate how the electrical CI artifact influences the recording, especially when stimulating in the daily routine setup, that is, via the sound-processor. Finally, we aimed to analyze objectively to which extent invasive recordings are superior to conventional surface electrode recordings.

2. Material and methods

The current project investigated the feasibility of recording neuronal signals in CI subjects via epidural electrodes implanted along with the CI over a course of several days. For this purpose, three epidural electrodes were implanted temporarily within the course of the CI surgery. With these electrodes AEP recordings were performed and compared to recordings using our standard clinical setup. The recordings were performed intraoperatively and postoperatively during the first days of CI use. After 4–5 days the epidural electrodes were removed.

The study was approved by the local ethics committee (approval number 6863) and is in accordance with the ethical standards of the Declaration of Helsinki. The participants were not paid for their participation. All participants gave informed written consent to the study.

2.1. Subjects

10 subjects were included in this study (45–80 yrs, 5× male, 5× female, 7× implantation on left side, 3× implantation on right side). All major brands were represented (4× Cochlear, 3× MED-EL, 2× Advanced Bionics, 1× Oticon). More details about the subjects and implants in use are provided in the supplementary table 1 (stacks.iop.org/JNE/16/056008/mmedia).

All data was analyzed by experienced audiologists. For the quality analysis, only the data sets of six patients were included (45–80 yrs, 5× implantation on left side, 2× implantation on right side). The data of the following patients had to be excluded: Epi01: slightly different recording settings and no single trial data; Epi03: strong artifacts, epidural electrodes presumably too close to the implant; Epi08: patient was deaf for a long time, thus no stimulus responses could be seen at all; Epi10: unintentional removal of two epidural electrodes during dressing changes.

2.2. Electrodes

For this study epidural electrodes from AD-Tech Medical Instrument Corporation (Racine, WI, USA) were used. The electrodes consist of a 5 mm long round stainless steel tip on a flexible Teflon™ lead which is 1.0 mm in diameter. The Teflon lead attaches to a 1.0 m silicone lead, terminating in a 1.5 mm touch proof connector. The electrodes are shown in figure 1(B).

2.3. Placement of the epidural electrodes

The electrodes were positioned over the temporal cortex, where we expected to capture the largest signal amplitude of activity originating in the auditory cortex. CI implantation was performed according to our clinical standards. After the implant was positioned the clinical standard CI evaluation was conducted which consists of impedance measurements, electrically evoked acoustic reflex thresholds and electrically evoked compound action potentials. Subsequently three additional holes (cranial posterior, cranial medial and cranial anterior the external auditory canal) were drilled down to the surface of the dura and the epidural electrodes were placed and fixed with first a collagen sponge coated with fibrinogen and thrombin and second with bone wax (figures 1(D) and (E)).

The respective cables of the three epidural electrodes were marked according to the electrode positions. After closing the wound the cable lines were twisted and the intraoperative epidural recordings were conducted. Subsequently, the bundle of cables was stored into a polythene bag and attached to the head bandage.

In all cases a cone beam CT scan (figures 1(F) and (G)) was conducted in order to evaluate the correct position of the CI electrode array and the epidural electrodes. The scan also belongs to our clinical standard evaluation and was either conducted intraoperatively or postoperatively.

2.4. Intraoperative recordings

Intraoperatively electrically evoked auditory brainstem responses (eABRs) were recorded. The stimulation was provided directly via the CI by using the clinical steering hard- and software from the respective CI manufacturer. For recording the Nicolet Viking EDX was used (Natus Medical Incorporated, Pleasanton, CA, USA). The recorded data was compared to our clinical standard setup using sub-dermal needle electrodes (Medtronic, Minneapolis, MI, USA).

2.5. Postoperative recordings

At the test switch-on appointment (1–3 days after surgery) the subjects received their CI processor, which was fitted as
Table 1. Stimulus responses visually evaluated by experienced audiologists for different potentials and recording settings: individual data.

| Recording setting | Epi01 | Epi02 | Epi03<sup>a</sup> | Epi04<sup>b</sup> | Epi05<sup>c</sup> | Epi06 | Epi07 | Epi08<sup>d</sup> | Epi09 | Epi10 |
|-------------------|-------|-------|-------------------|-------------------|-------------------|-------|-------|-------------------|-------|-------|
| cABR (intraop)    | ++    | −     | Art.              | +                 | ++                | +     | −     | Art.              | ++    | ++    | Art.              | Art.  | n/a   |
| ABR (postop)      | n/a   | −     | +                 | ++                | Art.              | Art.  | Art.  | Art.              | +     | −     | −                 | −     | Art.  | n/a   |
| MLR (postop)      | n/a   | Art.  | Art.              | +                 | +                 | Art.  | Art.  | n/a               | Art.  | −     | +                 | n/a   |       |
| CLR (postop)      | −     | ++    | +                 | ++                | Art.              | +     | ++    | ++                | +     | −     | −                 | −     | +++   | +++   |
| MMN (postop)      | −     | +     | +                 | ++                | ++                | +     | +     | −                 | −     | −     | n/a               | +     | +++   |
| P300 (postop)     | −     | +     | +                 | +                 | +                 | ++    | +     | +                 | +     | ++    | ++                | n/a   | n/a   | +     |

<sup>a</sup> Leads of epidural electrodes too close to CI.
<sup>b</sup> Subject had extremely thick bone.
<sup>c</sup> Bad impedances of adhesive electrodes due to strong perspiration.
<sup>d</sup> Subject had a long duration of deafness.

Evaluation: ++ ..., excellent responses, + ... moderate to good responses, − ... no responses.

Abbreviations: epi—epidural, Art.—recording disturbed by artifacts or intraoperatively also by electrical noise, n/a—not available.
standardized by our clinical fitting engineers. This included determination of the hearing thresholds and the maximum comfortable levels for each stimulation channel of the implant, as well as choosing the speech coding strategy and other necessary parameters. After the fitting session we performed the first AEP recording session. At the next session day the CI processor was re-fitted and the recording session was repeated.

During the postoperative recording sessions a variety of AEPs were recorded [41]. These measurements included auditory brainstem responses (ABR) via brainstem evoked response audiometry (BERA), 0–10 ms, middle latency responses (MLR) via middle latency response audiometry (MLRA), 10–50 ms, auditory cortical responses (ACR) via cortically evoked response audiometry (CERA), 50–400 ms, mismatch negativity (MMN) and P300. The data was recorded at a range of intensity levels.

A Nicolet Synergy EDX system (Natus Medical Incorporated, Pleasanton, CA, USA) was used to generate all stimuli and to record the evoked potentials. For the measurements the CI was stimulated acoustically by means of a 1 m long sound tube using clicks and tone bursts at different intensity levels. The sound tube was not particularly calibrated with regard to the applied stimuli, as with the CI sound processor, especially when applying Automatic Gain Control and Input Dynamic Range, calibration is affected anyway. Nevertheless, we included the set values of the intensities in order to represent differences between these levels. The recording setup consisted of five monopolar channels, which were placed in the additional holes (cranial posterior, cranial medial and cranial anterior the external auditory canal), as well as two adhesive electrodes (Asmuth GmbH Medizintechnik, Minden, Germany), placed at the ipsilateral & contralateral mastoid (M1, M2). The reference and ground electrode for all channels were additional adhesive electrodes placed at the forehead (Fpz and approximately Fp2, respectively), see figure 2. The details of the stimuli in use as well as the recording parameters can be found in the supplementary table 2.

For each patient the CERA was measured using different frequencies (500 Hz, 1000 Hz, and 2000 Hz) at different individual intensity levels on different days. Also BERA and MLR sessions (using clicks) and MMN and P300 sessions (500 Hz as standard and 1500 Hz as deviant) were performed. Due to their general health condition not all patients were able to perform all sessions at one day. Also the number of possible runs per session was varying due to the general health condition. However, all patients were able to perform at least two different sessions with two or more runs per session. During the CERA runs, patients were instructed to count, if possible, the number of tones (50 plus) presented, during the P300 runs, patients were instructed to count the number of the rare deviant tones (100 plus). The automatic online artifact rejection procedure of the Nicolet Synergy EDX system works as follows: the overall signal level during the artifact rejection time window (here: 1 ms) is checked in one recording channel (here: contralateral mastoid). If the level exceeds the artifact rejection threshold (here: 120 µV) the trial is rejected for all channels. Artifact afflicted trials are discarded until the predefined number of clean averages is reached. Due to this procedure the number of presented tones differed between runs. After each run patients were asked if they could perceive the tones at all, how many tones they counted and how loud they subjectively evaluate the loudness of the presented tones.

2.6. Removal of the epidural electrodes

After all recording sessions were finished, which was typically 4–5 days after surgery, the epidural electrodes were removed by pulling them out on the awake patient followed by a sterile bandage. No further sequelae have been noted.
2.7. Visual analysis

In a first step all recorded potentials were evaluated by two experienced audiologists independently. The applied criteria for presence or absence of waveforms were our clinical standards. The primary criterion was the clear presence of a waveform with the morphology of a typical waveform obtained with our standard scalp electrodes at roughly the same latency, as these were expected to be similar. Another criterion was that the identified waveform faded out with decreasing stimulation level. If the evaluation of both audiologists differed on a certain wave, the poorer evaluation result was used.

2.8. Data preprocessing & quality calculation

In a second step a quality investigation was developed in order to give an objective analysis of the differences between the recording settings. For these analyses we concentrated onto the postoperative CERA datasets. For different reasons only data from 6 subjects could be used here.

Data preprocessing and analysis was performed with MATLAB 2015a (The Mathworks, Inc., Natick, Massachusetts, USA) using custom written tools. The filtered data were exported from the recording system and imported into MATLAB. Afterwards three bipolar epidural recording channels (Medial-Anterior, Medial-Posterior and Posterior-Anterior) were calculated using the epidural recordings. In a next step the data of each separate channel was corrected individually for artifacts using the procedure and values (Artifact rejection level: 120 µV; Artifact rejection time: 1 ms) as implemented in the Nicolet Synergy EDX system. Trials with artifacts were discarded for the next processing steps. Afterwards all artifact free trials were averaged for each channel over a single run. Subsequently the channel specific N1 peaks were detected by means of an automatic procedure (search for local minima/maxima in a time window from 60 to 180 ms after stimulus onset) and the associated mean peak amplitude (A) and standard error (SE) were determined. The SE (standard deviation divided by the square root of the number of trials) was chosen to take also into account the different residual numbers of artifact free trails as a higher number of artifact free trials of a specific derivation indicate a better quality of the recording.

The given time window has been chosen based on our experience in recording CERA. In order to give an objective analysis of the differences between the recording settings a quality measure (Q) was introduced. Q is defined as the quotient of the absolute value of the amplitude and the standard error (a larger amplitude and a low SE indicates better signal quality of the N1). For a better understanding figure 2 depicts a schematic illustration of the whole procedure.

Figure 2. Schematic illustration of the preprocessing steps. Subsequent to the recording of 5 monopolar channels (M1, M2, Anterior, Medial and Posterior), three bipolar derivations (Medial-Anterior, Medial-Posterior and Posterior-Anterior) were calculated offline. Additionally the numbers of averages used are given. In a next step the channel specific N1 peak values were extracted (amplitude (A) and standard error (SE)) and used to calculate the quality measure (Q). Finally these values were used to perform the statistical evaluation.
Table 2. Averaged data (amplitude (A) standard error (SE) and calculated quality (Q)) of an example subject (Epi07) for a 1000 Hz left side (L)-CERA recording with intensity levels of 100, 90 and 80 dB (nHL).

| Intensity (dB) | Epi07 M1 | M2 | Anterior | Medial | Posterior | Med.-Ant. | Med.-Post. | Post.-Ant. |
|---------------|----------|----|----------|--------|-----------|-----------|------------|------------|
|               | A (µV)   | SE (µV) | Q (a.u.) | A (µV) | SE (µV) | Q (a.u.) | A (µV) | SE (µV) | Q (a.u.) | A (µV) | SE (µV) | Q (a.u.) | A (µV) | SE (µV) | Q (a.u.) | A (µV) | SE (µV) | Q (a.u.) |
| 100           | −9.53    | 6.98 | 1.37     | −8.58 | 2.62     | −15.59   | 3.73      | −13.34    | 3.41     | 3.91     | 4.39    | 3.73     | 1.18      | 8.78    | 0.98     | 8.97      | −9.84    | 0.73     | 13.40    | 18.36    | 1.33     | 13.79    |
| 90            | −8.34    | 7.20 | 1.16     | −6.64 | 3.15     | −10.57   | 4.51      | −8.94     | 4.35     | 2.06     | 8.29    | 4.16     | 1.99      | 6.11    | 0.69     | 8.83      | −9.31    | 0.66     | 14.17    | 15.22    | 1.08     | 14.16    |
| 80            | −1.75    | 7.96 | 0.22     | −7.18 | 2.55     | −11.74   | 3.98      | −8.38     | 4.00     | 2.10     | 14.62   | 4.06     | 3.60      | 7.25    | 1.48     | 4.90      | −10.47   | 1.55     | 6.75     | 17.46    | 1.51     | 11.58    |
| Mean          | −6.54    | 7.38 | 0.91     | −7.47 | 2.77     | −12.64   | 4.08      | −10.22    | 3.92     | 2.69     | 9.10    | 3.98     | 2.26      | 7.38    | 1.05     | 7.57      | −9.87    | 0.98     | 11.44    | 17.01    | 1.30     | 13.18    |
2.9. Statistical analysis

Statistical analysis of the data of the six patients was performed only on the runs where patients reported an auditory sensation (non heard runs were excluded). The statistical analysis was done using SPSS 24.0 (IBM, Armonk, NY). In general, p-values of .05 or less were regarded as statistically significant. Bonferroni correction was applied to correct for multiple comparisons. To evaluate differences between the three electrode configurations, we defined three clusters of electrodes:

(i) Scalp, including the electrodes M1 and M2
(ii) Scalp-Epi, containing the electrodes Anterior, Medial and Posterior, referenced against the adhesive reference
(iii) Epi, including the bipolar channels Medial-Anterior, Medial-Posterior and Posterior-Anterior.

Experimental conditions, including day of measurement, frequency and sound pressure level, varied between the different subjects, which was related to factors as subject status and weekday of the surgery. However, as we were primarily interested in the comparison of the different electrode configurations, we used a pure within-subject design and included all recorded data. The prerequisite however was that the respective combination of measured factors (subject × day of measurement × frequency × sound pressure level) was available for all electrodes. If this was not the case, these data points were excluded. By this it was ensured that no electrode configuration was biased by differences in the factors subject, day, frequency or loudness. In total, 80 different data points each were included for the analysis of amplitude, SE and quality measures of the auditory N1 component. As polarity of the N1 changed between the different electrodes we used the absolute values of the amplitudes. To contrast the different electrode clusters (Scalp, Scalp-Epi, Epi), separate paired sample t-tests were conducted for amplitude, SE and quality measures.

3. Results

The results of the visual evaluation of the experienced audiologists are given in table 1. The morphologies of the epidurally recorded waveforms were detected to be very similar to the ones recorded on the scalp, but in some recordings the polarity was reversed.

Figure 3 and table 2 show the averaged data and the quality measure of one example patient (Epi07) for a 1000 Hz CERA recording using different intensity levels at day 1. For further information the averaged data of the remaining patients for 1000 Hz CERA recording are illustrated in supplementary figures 1–5. Figure 4 depicts the specific clustered (according to the electrode configurations) A, SE and Q values of the six subjects in scatter plots.
Mean N1 amplitudes for the different electrode clusters were 4.43 µV (SD: 2.84 µV), 9.42 µV (SD: 4.83 µV) and 8.46 µV (SD: 5.73 µV) for Scalp, Scalp-Epi and Epi, respectively (figure 5(A)). The paired samples t-tests (see figure 5) revealed significant enhanced N1 amplitudes for the Scalp-Epi and the Epi cluster when compared to the Scalp cluster (Scalp-Epi: \( t(79) = -10.03, p < .001 \); Epi: \( t(79) = -5.97, p < .001 \)). The difference between the Scalp-Epi and the Epi cluster did not survive correction for multiple comparisons (\( p = .06, \text{corr.} \)).

Mean SE values for N1 amplitudes were 3.87 µV (SD: 1.11 µV), 4.08 µV (SD: 1.08 µV) and 2.40 µV (SD: 0.98 µV) for Scalp, Scalp-Epi and Epi, respectively (figure 5(B)). Pairwise comparisons showed smaller SE values for the Epi cluster compared to both, the Scalp and the Scalp-Epi cluster (Scalp: \( t(79) = 8.08, p < .001 \); Scalp-Epi: \( t(79) = 12.45, p < .001 \)). When correcting for multiple comparisons, no difference was observed between the clusters Scalp and Scalp-Epi (\( p = 1.08, \text{corr.} \)). Mean values of quality, that is, the ratio of N1 amplitude to SE of the N1 amplitude, were 1.31 (SD: 0.96), 2.47 (SD: 1.51) and 4.06 (SD: 3.12) for Scalp, Scalp-Epi and Epi,

Figure 4. 2D plots of standard error (SE (µV)), amplitude values (A (µV)) and quality measure (Q (a.u.)) of the three different clusters of electrodes (Scalp: M1, M2; left panel; Scalp-Epi: Anterior, Medial, Posterior; middle panel; Epi: Medial-Anterior, Medial-Posterior and Posterior-Anterior; right panel). Data are shown separately for 500 Hz, 1000 Hz and 2000 Hz CERA recordings averaged over two days of recording sessions.
respectively (figure 5(C)). Statistical analysis revealed quality to be significantly higher for the Epi cluster compared to both, the Scalp and the Scalp-Epi cluster (Scalp: \(t(79) = -7.95, p < .001\); Scalp-Epi: \(t(79) = -5.02, p < .001\)). In addition, quality was as well significantly higher for the Scalp-Epi compared to the Scalp cluster (\(t(79) = -10.13, p < .001\)).

4. Discussion

In the present study we recorded AEPs invasively by means of epidural electrodes in CI users and compared the results to our clinical standard recording setting using scalp electrodes. For the invasive recording, three epidural electrodes were placed temporally over the auditory cortex. Our study showed the principal feasibility of this approach. In nine out of ten cases, the epidural electrodes could be placed and fixed such that they did not migrate within the course of measurements. After 4–5 days the epidural electrodes could be removed safely and without pain for the patients.

Intraoperative eABR recordings as evaluated by experienced audiologists revealed better response thresholds with our clinical scalp electrode set-up than with the epidural electrodes. However, the positions of the epidural electrodes were optimized for cortical response recordings, not for ABR measurements, which might explain this finding. Slutzky et al showed that the spatial resolution of invasive epidural recordings is better than with surface recordings [24]. On the other hand, Schwartz et al stated that the accuracy and specificity is enhanced for cortex compared to surface settings [42].

Following, the generators of the evoked potentials in the brain have to be targeted more exactly when recording directly on the dura. This would explain the superiority of the clinical scalp set-up for measuring eABR responses.

Postoperatively, the CI was fitted with an individual clinical map, and the stimulation was performed acoustically to include the preprocessing of the CI and mimic everyday listening situations as close as possible. However, in this recording mode the CI artifact is not stimulus locked, making ipsilateral recordings of early potentials challenging. This was observed in the present ABR and MLR recordings, contaminated by the artifact, however, in two cases we obtained clear MLRs with response thresholds being lower with the epidural compared to the clinical standard scalp set-up. A future CI system, fully integrating electrodes for the measurement of event-related potentials, would have the full knowledge about the characteristics of the electrical artifact. This information could be used by the system to better deal with the artifact and to further optimize the signal quality.

A better signal quality could be observed for the late cortical potentials, using the epidural compared to the surface electrode setup. The N1 was much clearer than with the adhesive electrodes and already visible at lower stimulation intensities in all cases. Furthermore, the signal was less disturbed by artifacts e.g. due to movements.

With our quality measure we were able to show objectively the superiority of epidurally recorded compared to scalp recorded potentials. This result was to be expected as the electrodes are closer to the signal source, and not dampened by the skull.

![Figure 5. Boxplots displaying the distribution of (A) amplitude values (A (µV)), (B) standard error (SE (µV)) and (C) quality (Q (a.u.)) for the three different electrode clusters: monopolar adhesive (scalp, cyan), monopolar epidural (scalp-Epi, green) and bipolar epidural (epi, magenta). Significant differences are indicated with respective p-values, corrected for multiple comparisons.](image_url)
Also for MMN and P300 we detected clearer potentials with the epidural settings compared to the scalp electrodes. Further investigations on the feasibility of recording late cortical potentials like the P300 and the MMN by means of epidural electrodes will be performed in an upcoming study, where we will use a full EEG amplifier and additional stimulus material (e.g. speech).

Nourski et al showed the feasibility of recording auditory cortical responses intracranially in an experienced CI subject using an electrode grid, implanted contralateral to the stimulus side in the course of an epilepsy surgery. Stable and clear auditory cortical responses were obtained for a diverse range of auditory stimuli (pure tones, click trains, amplitude-modulated noise, speech). In this respect it has to be mentioned that numerous studies using scalp recordings in humans as well as intracranial recordings in animals have shown evidence that auditory sensation induced cortical responses are predominantly contralateral or at least larger than ipsilateral. In contrast, our recordings were performed directly after CI activation and were restricted concentrated to three electrodes on the ipsilateral side that can be implanted easily during a standard CI surgery without the need of a larger surgical intervention [36]. However, anecdotally it was possible to perform additional contralateral recordings (presented in supplementary figure 6) which depict, as expected, the above mentioned larger contralateral cortical response.

Sandmann et al showed a rapid adaptation of the activation of the auditory cortex after implant activation, with the most rapid changes occurring within the first eight weeks after CI activation [43]. Accordingly, it might be likely that the signal quality of all recording settings would be even better after the full rehabilitation of the auditory cortex. After each recording session our subjects were urged to practice hearing with the implant. Furthermore the CI processor was re-fitted directly before each recording session to take into account neural changes. Over the course of recording sessions, improvements were observed in some, but not all subjects, with some subjects even showing a deterioration in activation. As our recording protocol for each session was quite long, one reason for these deteriorations might be fatigue of the subjects. However, even in the very first days after CI activation we could record clear cortical responses in all subjects except one. This subject however was deaf for more than 20 years and was not able to count the number of stimuli reliably which likely explains the absence of clear stimulus responses right after the initial activation of the CI.

There are various applications of an auditory closed loop system in CI users, for example active and passive CI adaptation, automated fitting of the CI or CI evaluation in children. Obviously, the setup has to be optimized and tailored, to be used in an everyday life context. Here, implanting the recording electrodes would be a convenient solution and therefore constitutes a promising approach. Most invasive recordings for BCI applications aim for motor control [19–23], and to our knowledge there is currently no study using intracranial recordings for control of a hearing device. Our study shows the potential of epidural AEP recordings, which might find their way into future BCI systems for the control of hearing devices.

A possible further step towards a closed loop CI system would be the implementation of automatic classification algorithms that are based on single trial data and that could for example be used for an automated fitting procedure. Here, an objective classification of whether or not specific sound characteristics—e.g. frequency or loudness—of different tones can be discriminated could help especially in patient groups not capable of an active participation. A promising approach in this regard is the use of AEP, including also the P300 wave. Previous studies have already shown the possible suitability of different auditory paradigms [44–46]. First results in CI users by Finke et al showed that EEG data can be automatically classified on a single-trial level using scalp mounted electrodes, providing information on whether a difference in, for example, tone frequency could be perceived or not [29]. In a next step, the feasibility of epidural recordings for single trial EEG classification will be evaluated, including studies with auditory discrimination tasks.

5. Conclusion

In this study, epidural recordings of AEPs in CI users were performed and compared to scalp recordings. This approach was chosen to improve data acquisition in CI users with the aim to get new insights into how to make auditory closed loop systems more applicable for daily life. Accordingly, the study protocol provides the use of daily routine CI settings, a realistic number of recording channels and recording sites which are realistically accessible during CI surgery. The epidural recordings yielded waveforms similar to common surface recordings, suggesting the suitability of epidurally recorded potentials in an auditory closed loop system.

The quality measure analysis showed a significantly higher signal quality for bipolar compared to monopolar (referenced to surface electrode) epidural recordings, with the latter, in turn, showing a better signal quality than pure surface recordings. This suggests that the quality and accuracy of auditory closed loop systems could be improved by using epidural electrodes.

In order to realize the striven wearing comfort, a set of epidural recording electrodes would have to be fully integrated into a CI system. This, however, also calls for technological progress, optimizing the CI itself to be used as an AEP recording device.

Acknowledgments

This work was supported by the German Research Foundation (Deutsche Forschungsgemeinschaft, DFG), Cluster of Excellence Hearing4all (EXC 1077/1). This paper only reflects the authors’ views and funding agencies are not liable for any use that may be made of the information contained herein.
We would like to thank our patients who participated in this study. Also we would like to thank our surgeons and our staff from the operation theatre as well as our CI engineers for their support. We are grateful to S Debener for accompanying discussions that proved helpful.

**Author’s Statement**

The authors report no declarations of interest. Informed consent has been obtained from all individuals included in this study. Ethical approval: the research related to human use has been complied with all the relevant national regulations, institutional policies and in accordance the tenets of the Helsinki Declaration, and has been approved by the authors’ institutional review board or equivalent committee.

**ORCID iDs**

S Haumann [https://orcid.org/0000-0002-9138-0277](https://orcid.org/0000-0002-9138-0277)

G Bauernfeind [https://orcid.org/0000-0002-5205-8920](https://orcid.org/0000-0002-5205-8920)

**References**

[1] Büchner A and Gärtner L 2017 Technical advancements in cochlear implants: state of the art *HNO* 65 276–89

[2] Lenarz T 2017 Cochlear implant—state of the art. GMS. Curr. Top Otorhinolaryngol. Head Neck Surg. 16 Doc04

[3] Loizou P C 1999 Signal-processing techniques for cochlear implants *IEEE Eng. Med. Biol. Mag.* 18 34–46

[4] Wilson B S and Dormann M F 2008 Cochlear implants: a remarkable past and a brilliant future *Hea. Res.* 242 3–21

[5] Büchner A, Frohne-Büchner C, Gärtner L, Lesinski-Schiedat A, Battmer R-D and Lenarz T 2006 Evaluation of advanced bionics high resolution mode *Int. J. Audiol.* 45 407–16

[6] Krüger B, Joseph G, Rost U, Strauss-Schier A, Lenarz T and Büchner A 2008 Performance groups in adult cochlear implant users: speech perception results from 1984 until today *Otol. Neurotol.* 29 509–12

[7] Lenarz M, Sönmez H, Joseph G, Büchner A and Lenarz T 2012 Long-term performance of cochlear implants in postlingually deafened adults *Otolaryngol. Head Neck Surg.* 147 112–8

[8] Lenarz T et al 2013 European multi-centre study of the nucleus hybrid L24 cochlear implant *Int. J. Audiol.* 52 838–48

[9] Teschner M, Polite C, Lenarz T and Lustig L 2013 Cochlear implantation in different health-care systems: disparities between Germany and the United States *Otol. Neurotol.* 34 66–74

[10] Wilson B S and Dormann M F 2008 Cochlear implants: current designs and future possibilities *J. Rehabil. Res. Dev.* 45 695–730

[11] Haumann S, Lenarz T and Büchner A 2010 Speech perception with cochlear implants as measured using a moving-roving-adapter test method *ORL—J. Oto-Rhino-Laryngol. Head Neck Surg.* 72 312–8

[12] Bruns L, Mürbe D and Hahn A 2016 Understanding music with cochlear implants *Sci. Rep.* 6 32026

[13] Caldwell M, Jian N and Limb C 2017 Assessment and improvement of sound quality in cochlear implant users *Laryngoscope Investigative Otolaryngol.* 2 119–24

[14] Haumann S, Mühler R, Ziese M and von Specht H 2007 Discrimination of musical pitch with cochlear implants (Diskriminierung musikalischer Tonhöhen bei Patienten mit Koecheimplantat) *HNO* 55 613–9

[15] McDermott H J 2004 Music perception with cochlear implants: a review *Trends Amplification* 8 49–82

[16] Müller R, Ziese M and Rostalski D 2009 Development of a speaker discrimination test for cochlear implant users based on the oldenburg logatome corpus *ORL—J. Oto-Rhino-Laryngol. Head Neck Surg.* 71 14–20

[17] Wolpaw J R, Birbaumer N, McFarland D J, Pfurtscheller G and Vaughan T M 2002 Brain–computer-interfaces for communication and control *Clin. Neurophysiol.* 113 767–91

[18] Välsäinen J, Wendel K, Seemann G, Malmiuvo J and Hyytin J 2009 Sensitivities of bipolar subcutaneous and cortical EEG leads *World Congress on Medical Physics and Biomedical Engineering* (Munich, Germany, 7–12 September 2009) (IFMBE Proc. vol 25) ed O Düssel and C Schlegel pp 267–70

[19] Bleichner M, Freudenburg Z, Jansma J, Aarnoutse E, Vansteensel M and Ramsey N 2016 Give me a sign: decoding four complex hand gestures based on high-density ECoG *Brain Struct. Funct.* 221 203–16

[20] Leuthardt E C, Schalk G, Wolpaw J R, Ojemann J G and Moran D W 2004 A brain–computer interface using electrocorticographic signals in humans *J. Neural Eng.* 1 63–71

[21] Schalk G, Kubanek J, Miller K, Anderson N, Leuthardt E, Ojemann J, Limbrick D, Moran D, Gerhardt L and Wolpaw J 2007 Decoding two-dimensional movement trajectories using electrocorticographic signals in humans *J. Neural Eng.* 4 264–75

[22] Wang W et al 2013 An electrocorticographic brain interface in an individual with tetraplegia *PLoS One* 8 e55344

[23] Vansteensel M, Hermes D, Aarnoutse E, Bleichner M, Schalk G, van Rijen P, Leijten F and Ramsey N 2010 Brain–computer interfacing based on cognitive control *Ann. Neurosci.* 67 809–16

[24] Slutzky M, Jordan L, Krieg T, Chen M, Mogul D and Miller L 2010 Optimal spacing of surface electrode arrays for brain-machine interface applications *J. Neural Eng.* 7 26004

[25] Valderrama A, Oostenveld R, Vansteensel M, Huiskamp G and Ramsey N 2010 Gain of the human dura in vivo and its effects on invasive brain signal feature detection *J. Neurosci. Methods* 187 270–9

[26] Hochberg L, Serruya M, Frieds G, Mukand J, Saleh M, Caplan A, Branner A, Chen D, Penn R and Donoghue J 2006 Neuronal ensemble control of prosthetic devices by a human with tetraplegia *Nature* 442 164–71

[27] Maynard E M, Nordhausen C T and Normann R A 1997 The utah intracortical electrode array: a recording structure for potential brain–computer interfaces *Electroencephalogr. Clin. Neurophysiol.* 102 228–39

[28] Müller-Putz G R et al 2015 The Future of Brain/Neural Computer Interaction: Horizon 2020 (Graz: Graz University of Technology) (https://doi.org/10.3217/978-3-85125-379-5 BNCI Horizon 2020 Project ID: 609593 Funded under FP7-ICT )

[29] Finke M, Billinger M and Büchner A 2017 Toward automated cochlear implant fitting procedures based on event-related potentials *Ear Hear.* 38 e118–27

[30] Zander T O and Kothe C 2011 Towards passive brain–computer interfaces: applying brain–computer interface technology to human-machine systems in general *J. Neural Eng.* 8 025005

[31] Bleichner M, Lundbeck M, Selisky M, Minow F, Jäger M, Emkes R, Debecker S and De Vos M 2015 Exploring miniaturized EEG electrodes for brain–computer interfaces. An EEG you do not see? *Physiol. Rep.* 3 e12362
[32] Bleichner M and Debener S 2017 Concealed, unobtrusive ear-centered EEG acquisition: cEEGrids for transparent EEG Frontiers Human Neurosci. 11 163
[33] Frohne C, Lesinski A, Battmer R D and Lenarz T 1997 Intraoperative test of auditory nerve function Am. J. Otol. 18 893–4
[34] Frohne C, Lesinski A, Battmer R D and Lenarz T 1997 Monitoring the electrode position during acoustic neuroma surgery Am. J. Otol. 18 895–6
[35] Heinke W and Koelsch S 2005 The effects of anesthetics on brain activity and cognitive function Curr. Opin. Anesthesiol. 18 625–31
[36] Nourski K, Etler C, Brugge J, Oya H, Kawasaki H, Reale R, Abbas P, Brown C and Howard M 2013 Direct recordings from the auditory cortex in a cochlear implant user J. Assoc. Res. Otolaryngol. 14 435–50
[37] McLaughlin M, Lu T, Dimitrijevic A and Zeng F-G 2012 Towards a closed-loop cochlear implant system: application of embedded monitoring of peripheral and central neural activity IEEE Trans. Neural Syst. Rehabil. Eng. 20 443–54
[38] McLaughlin M, Lopez Valdes A, Reilly R and Zeng F-G 2013 Cochlear implant artifact attenuation in late auditory evoked potentials: a single channel approach Hear. Res. 302 84–95
[39] Haumann S, Bauernfeind G, Bleichner M, Teschner M, Debener S and Lenarz T 2016 Epidural recordings of auditory evoked potentials in cochlear implant users: first experiences J. Otol. Rhinol. 5 5
[40] Haumann S, Bauernfeind G, Bleichner M G, Teschner M J, Debener S and Lenarz T 2016 Track U. Epidural recordings of auditory evoked potentials in cochlear implant users-first cases Biomed. Tech. 61 s240
[41] Picton T W, Hillyard S A, Krausz H I and Galambos R 1974 Human auditory evoked potentials. I: evaluation of components Electroencephalogr. Clin. Neurophysiol. 36 179–90
[42] Schwartz A, Cui X, Weber D and Moran D 2006 Brain-controlled interfaces: movement restoration with neural prosthetics Neuron 52 205–20
[43] Sandmann P, Plotz K, Hauthal N, de Vos M, Schönfeld R and Debener S 2015 Rapid bilateral improvement in auditory cortex activity in postlingually deafened adults following cochlear implantation Clin. Neurophysiol. 126 594–607
[44] Bauernfeind G, Horki P, Kurz E-M, Schöpfling W, Pichtler G and Müller-Putz G R 2015 Improved concept and first results of an auditory single-switch bci for the future use in disorders of consciousness patients Conf. Proc. IEEE Eng. Med. Biol. Soc. 2015 1902–5
[45] Kübler A, Furdea A, Halder S, Hammer E M, Nijboer F and Kotchoubey B 2009 A brain computer interface controlled auditory event-related potential (P300) spelling system for locked-in patients Ann. New York Academy Sci. 1157 90–100
[46] Pokorny C et al 2013 The auditory P300-based single-switch brain–computer interface: paradigm transition from healthy subjects to minimally conscious patients Artif. Intell. Med. 59 81–90