A behavioral method to estimate charge integration efficiency in cochlear implant users

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A B S T R A C T

Background: In cochlear implants, pulse amplitude (PA) or pulse phase duration (PPD) can be used to increase loudness. Loudness grows more slowly with increasing PPD, resulting in a larger dynamic range (DR), possibly reflecting “leaky” charge integration associated with neural degeneration due to hearing loss. Here, we propose a method to estimate charge integration efficiency for CI users.

New method: The DR was measured with increasing PA or PPD, relative to a common threshold anchor with a short PPD (25μs/ph); DRs were converted to the common unit of charge (nC). Charge integration efficiency was calculated as the dB difference in DR with increasing PPD or PA. Loudness growth functions were also compared as PA or PPD was increased relative to the common threshold.

Results: Ten CI ears were tested; all participants were adult users of Cochlear® devices. DR was significantly larger when PPD was increased, requiring (on average) 70 % more charge than when PA was increased. A significant correlation (p = 0.007) was observed between duration of deafness and charge integration efficiency, largely driven by a participant with long auditory deprivation in both ears. Loudness growth was slower when PPD was increased, consistent with previous studies.

Comparison to Existing Methods. The present method offers a quick behavioral test with which to measure charge integration efficiency, which may be a useful measure of neural health.

Discussion: Charge integration efficiency may be used to probe neural health independent of absolute detection thresholds, which mostly reflect the proximity of electrodes to neural populations.

1. Introduction

Cochlear implants restore hearing sensation in individuals with severe-to-profound hearing loss by stimulating surviving auditory neurons with electric current. Typically, biphasic pulse trains are used for electric stimulation. For each pulse, the pulse amplitude (PA) specifies the amount of current and the pulse phase duration (PPD) specifies the duration of that current within each phase of the pulse. The total charge delivered by the stimulus is the product of PA and PPD, typically expressed in terms of nanocoulombs (nC). While increasing PA or PPD can both increase the charge delivered and loudness, the parameters do not trade off proportionally. Shannon (1985) found that loudness grows much more steeply with increasing PA for pulse trains with short PPD than with long PPDs. Zeng et al. (1998) reported that the “equal charge = equal loudness” assumption does not hold for threshold or supra-threshold stimulation. In that study, a loudness-balance method was used to equate loudness between stimuli with relatively short or long PPDs by adjusting the PA. The authors reported that a greater amount of charge was required for longer PPDs to match loudness to shorter PPDs, and found that a power function best described the loudness relationship between PA and PPD. Chatterjee et al. (2000) found a similar loudness relationship between PA and PPD using a loudness scaling method.

These results all indicate that, given a fixed amount of charge, long PPDs effectively produce less neural excitation than short PPDs. This is thought to reflect “leaky” charge integration of the auditory nerve, where the neural membrane loses the injected charge over time (Shannon, 1985; Shepherd and Javel, 1999). If integration were perfect, doubling the PPD would improve thresholds by 6 dB. However, the slope of threshold versus PPD functions tends to be much shallower, and is highly variable among implant listeners. Pfingst and Sutton (1983) suggested that this variability may reflect differences in neural survival and/or health among cochlear implant patients, which was later confirmed by physiological studies (e.g., Van den Honert and...
A number of factors resulting from neural degeneration following hearing loss can change the integrating property of the auditory nerve, including the loss of peripheral processes, demyelination of the cell body, and/or cell loss. These factors may result in fewer ion channels available for integration, or a change in the spike initiation site towards the extensively myelinated central axon which has a lower membrane capacitance and shorter integration time constant. The efficiency of excitation can also depend on stimulation parameters unrelated to neural degeneration, such as the stimulation mode or electrode-neural interface (the position of the electrode relative to neural populations). The stimulation mode specifies the active and return electrodes, allowing for relatively broad (monopolar stimulation with extra-cochlear return electrodes) or narrow stimulation (e.g., bipolar stimulation between two intra-cochlear electrodes). Modeling and physiological studies suggest that more efficient integration (e.g., a steeper threshold versus PPD function) occurs when a greater number of nodes fall into the depolarized region of the function, as occurs with broad monopolar stimulation (Rattay, 1989; Smith and Finley, 1997).

Chatterjee et al. (2014) found steeper loudness growth with increasing PA for a broad stimulation mode and long PPDs, compared to a narrow stimulation mode and short PPDs. These results suggest that recruiting a larger neural population with a broad stimulation mode may overcome disadvantages associated with long PPDs. In contrast, narrow stimulation or perimodiolar placement of electrodes may better target the peripheral processes of auditory neurons, resulting in a better integration time constant (Van den Honert and Stypulkowski, 1984). Taken together, these studies suggest that interactions among stimulation parameters can be complex.

The goal of the present study was to evaluate a novel behavioral measure with which to characterize the efficiency of charge integration in human cochlear implant users, which in turn could be used to probe the neural health among implant users. To investigate charge integration efficiency, we measured loudness growth functions with increasing PA or PPD, both anchored to a common threshold. Slower loudness growth and a larger dynamic range (DR) would suggest greater leakiness or less efficient integration. As noted above, growth of excitation may also be affected by factors unrelated to neural health. To minimize these factors, we defined “charge integration efficiency” as the dB difference between DRs measured with increasing PPD and increasing PA, relative to a common anchor threshold. Using dB units (converted from electric charge) reduces the contribution of non-neural factors (e.g., the electrode-neural interface; Bierer et al., 2011) when estimating neural health. We further examined the relationship between charge integration efficiency and the duration of deafness before implantation to determine whether charge integration efficiency could be a worthwhile tool with which to probe neural health in cochlear implant users. We predicted that DRs would be larger with increasing PPD than PA, and that longer duration of deafness would be associated with poorer charge integration efficiency. We were also interested in characterizing loudness growth within the DRs, and observing whether loudness grew similarly for different portions of the DR as PA or PPD was increased.

## 2. Methods

### 2.1. Participants

Ten CI ears (3 bilateral CI users, 4 unilateral CI users) were tested in this experiment. All participants except for bilateral CI user S16 were adult, post-lingual, users of Cochlear® devices; S16 was prelingually deaf, with duration of deafness > 45 years in either ear. All test ears had at least 3 years of CI experience and all were native speakers of American English. The mean age at testing was 67.8 years, the mean duration of deafness was 12.4 years, and the mean CI experience was 9.7 years. Demographic information for CI test ears is shown in Table 1. All participants provided informed consent and the study was approved by the local Investigational Review Board (UMCIRB 13-001783).

### 2.2. Stimuli

All stimuli were charge-balanced, symmetrical, biphasic pulse trains presented to El 11 in monopolar (MP1 + 2) stimulation mode. The stimulation rate was 1000 pulses-per-second and the interphase gap was 8 μs. The pulse train duration was 300 ms. The stimulation parameters were selected to be similar to those used in clinical fitting of Cochlear® devices. All stimuli were presented via research interface (NIC II) connected to a Nucleus® Freedom processor (Cochlear Corporation, Englewood, CO), which bypassed participants’ clinical processors and allowed for precise control of stimulation parameters.

### 2.3. Dynamic range estimation and calculation of charge integration efficiency

Electrode DR was estimated for stimuli where PA or PPD was increased relative to a common threshold anchor (i.e., the same threshold was used to measure DR with increasing PA or PPD). For all stimuli, threshold was estimated using Bekesy tracking. First, PA threshold was measured for a short PPD (25 μs/phase). As illustrated in Fig. 1, this...
Threshold was used as the common anchor with which to measure DRs with increasing PA (1) or PPD (2). Maximum acceptable loudness (MAL) was estimated by slowly increasing PA ("PA stimulus") or PPD ("PPD stimulus"). The DRs for the PA and PPD stimuli were calculated as the difference between MAL and threshold, in terms of charge (nano-coulombs, or nC).

After estimating DRs, charge integration efficiency was calculated as:

\[
\text{Charge integration efficiency} = 20 \log \left( \frac{\text{PPD DR}}{\text{PA DR}} \right)
\]  

where DR = the dynamic range with increasing PPD or PA (in charge) relative to the common threshold anchor. By calculating the log difference in DR relative to a common threshold, the contribution of absolute threshold levels and absolute DR values to charge integration efficiency was reduced. It is important to note that the calculation depends on a common threshold anchor for the PPD and PA DRs, and that all units be converted to charge before calculation.

2.4. Loudness growth and loudness balancing

To measure loudness growth, loudness rankings were obtained for stimuli presented at 10, 20, 30, 40, 50, 60, 70, 80, and 90% within the PA or PPD DR, as in Chatterjee et al. (2000); the percent DR for increasing PA was calculated in terms of linear \( \mu \)A. Loudness rankings were collected in separate test blocks for the PA and PPD stimuli. Each test block contained 10 presentations at each percent DR presented in random order; the order of the test blocks was randomized across test ears. During testing, a stimulus would be played and the participant indicated the loudness on a scale of 1–9. Loudness ratings were averaged across the 10 presentations for each percent DR.

Stimuli with relatively short PPDs (probe) or long PPDs (reference) were also loudness-balanced at relatively soft, medium, and loud presentation levels using a method of adjustment. Reference stimuli (with a fixed low PA) were presented at 30, 50, and 70% of the PPD DR. During loudness balancing, the reference and probe stimuli were presented in sequence repeatedly, and the PA of the probe stimulus was adjusted until achieving equal loudness; this procedure was repeated two times.

3. Results

3.1. Dynamic range

Table 2 shows threshold, MAL, and DR for each test ear as PA or PPD was increased. When PA was increased, the DR across test ears ranged from 9.0–21.8 nC. When PPD was increased, the DR across test ears ranged from 10.6–44.0 nC. Fig. 2 shows DR with increasing PA or PPD for individual test ears. The difference in DR with increasing PPD or PA ranged from 0.5–31.7 nC. The mean difference between the PPD and PA DRs was 9.0 ± 9.5 nC. A Wilcoxon signed rank test showed that DR was significantly larger when PPD rather than PA was increased.
3.2. Loudness ratings

Fig. 5 shows loudness ratings as a function of charge as the PA or PPD was increased. In general, loudness grew much more quickly with increasing PA than PPD. As a function of charge level, loudness ratings were similar for the lower portion of the DR when PA or PPD was increased, and began to diverge for the upper portion of the DR. Power functions were fit to the data shown in Fig. 5, in the form of \( L = a x^{b} \) (as in Chatterjee, 1999; Chatterjee et al., 2000), where \( L \) is loudness, \( a \) is the steepness of the function, \( x \) is the intensity in nC, and \( b \) is the exponent of function.

Fig. 6 shows the values for exponent \( b \) of the power functions fit to the data in Fig. 4 for PPD loudness growth as a function of PA loudness growth. The mean exponent \( b \) values were 2.2 and 1.6 for the PA and PPD stimuli, respectively. Paired \( t \)-tests showed that exponent \( b \) values were significantly greater for the PA than for the PPD stimulus \([t (9) = 4.3, p = 0.002] \), indicating steeper loudness growth.

3.3. Loudness balancing

Probe stimuli with a short PPD (25 μs/ph) (i.e., PA stimuli) were loudness-balanced to reference stimuli with longer PPDs (i.e., PPD stimuli) presented at 30, 50, and 70 % of the PPD DR, representing relatively soft, medium, and loud presentation levels. During testing, the PA of the probe stimulus was adjusted until achieving equal loudness. Fig. 7 shows the charge needed to maintain equal loudness as the PA or PPD was increased. The symbols show loudness-balancing data. The numbers in each panel show estimated loudness ratings extrapolated from the power function fits to the data in Fig. 5; note that the estimated loudness versus charge values may sometimes deviate from the real values (symbols) shown in Fig. 5. In general, greater charge was needed for a long PPD, relative to a short PPD. A two-way RM ANOVA was performed on the data shown in Fig. 7, with loudness level (1, 2, 3, 4, 5, 6, 7, 8, 9, soft, medium, loud) and PPD (relatively short or long) as factors. Results showed significant effects for loudness level \([F (11,99) = 40.3; p < 0.001]\) and PPD \([F(1,99) = 6.2; p = 0.034]\); there was a significant interaction \([F(11,99) = 8.1; p < 0.001]\). Paired \( t \)-tests showed that long PPDs required significantly greater charge than short PPDs at loudness ratings 5 \((p = 0.016)\), 6 \((p < 0.001)\), 7 \((p < 0.001)\), 8 \((p < 0.001)\), 9 \((p < 0.001)\), and at the loud presentation level \((p = 0.017)\). There was no significant difference in charge between short and long PPDs at loudness ratings 1, 2, 3, 4, or at the soft and medium presentation levels \((p > 0.05\) in all cases).

The estimated loudness ratings and loudness balance data shown in Fig. 7 appear to align well, suggesting that both methods produced similar patterns of results in terms of the loudness relationship (in charge) between the PA and PPD stimuli. Linear slopes were fit to the data shown in Fig. 5, in the form of \( x(b) \) for loudness-balanced test ears (the symbols in Fig. 7). A paired \( t \)-test showed no significant difference in slope between short and long PPDs at loudness ratings 1, 2, 3, 4, or at the soft and medium presentation levels \((p > 0.05\) in all cases).

4. Discussion

In this paper, we evaluated a novel method to estimate charge integration efficiency for electrode stimulation sites in cochlear implant users. Charge integration efficiency was defined as the dB difference (converted from charge unit) between DRs with increasing PPD or PA, relative to a common threshold anchor. Different from previous studies, a fixed threshold with a short PPD (25 μs/ph) was used to estimate DRs with increasing PA or PPD (Fig. 1), and DRs were compared in terms of charge. Using a common threshold anchor and converting units to charge removed the contribution of the absolute threshold values and allowed for more direct comparison of the effects of increasing PA or PPD on DRs. Such direct comparison would be computationally more efficient.
Fig. 5. Individual loudness ratings with increasing PA or PPD as a function of charge. The solid lines show power functions fit to the data. The fixed PPD used for the PA loudness growth function and the fixed PA for the PPD loudness growth function are shown in the figure legend for each ear tested.
Individual exponents from the power functions fit to the data from Fig. 4. Values below the diagonal line indicate that loudness growth was steeper with increasing PA than with increasing PPD.

Fig. 6. Individual exponents from the power functions fit to the data from Fig. 4. Values below the diagonal line indicate that loudness growth was steeper with increasing PA than with increasing PPD.

difficult if a common threshold in charge units was not used. For example, Zeng et al. (1998) and Chatterjee et al. (2000) used power functions to estimate the tradeoff between PA and PPD for loudness growth. While such an approach may similarly characterize the effects of increasing PA or PPD on loudness, it is less intuitive, more time consuming, and more computationally complex. Using a short PPD also reduced the contribution of leakiness to threshold estimates, as the charge integration window was sufficiently short. This is borne out by the similar loudness growth functions with increasing PA or PPD for the lower portion of the DR (Fig. 5). Comparing the ratio between the PPD and PA DRs (i.e., the log difference between DRs in charge units) reduced the influence of absolute DRs on estimates of charge integration efficiency. For example, for electrodes with overall larger DRs, simply subtracting the PA DR from the PPD DR would result in a greater absolute difference. Such large DRs and slow loudness growth may not be exclusively due to neural degeneration, and may also reflect the electrode-neural interface and/or impedance at the stimulation site. As such, comparing DRs on log scale reduces such effects. This is further supported by the lack of correlation between thresholds (which largely relate to the electrode-to-neuron interface) and charge integration efficiency. Charge integration efficiency was measured using clinically relevant stimulation parameters (e.g., monopolar stimulation, relatively high stimulation rates, relatively small range of PPDs), as opposed to previous studies that typically measured loudness growth with bipolar stimulation, relatively low rates, and a wide range of PPDs (Shannon et al., 1983; Zeng et al., 1998; Chatterjee et al., 2000). Using clinically relevant stimulation parameters allowed charge integration efficiency to be contextualized within the clinical fitting parameter space.

4.1. Effects of increasing PA or PPD on DR

The absolute PA and PPD DRs (in terms of charge) were correlated, suggesting that there may be common non-neural factors underlying PA and PPD DRs. For example, greater impedance or perimodiolar placement of electrodes may result in slower growth of excitation. It is also possible that ears that demonstrate slower loudness growth (and presumably increased leakiness) with increasing PPD may also exhibit such effects at very short PPDs (e.g., the 25 μs/ph PPD used in this study and for Cochlear © clinical processors). However, as shown in Fig. 5, loudness grew similarly in many cases for the lower portion of the PA and PPD DRs. In general, PPD needed to be greater than 80 μs/ph to diverge from the PA loudness growth functions, suggesting that PA loudness growth with a fixed short PPD may not have been affected by leaky charge integration. It is also possible that for ears with severe neural degeneration, the surviving auditory neurons may be less responsive to increasing charge (i.e., reduced excitability), whether by increasing PA or PPD. However, the loudness growth function with increasing PA for S16L (which had a long period of deprivation and presumably greater degeneration) appears to be comparable to other test ears. Taken together, it is likely that the underlying mechanism for the correlation between the absolute PPD and PA DRs values may be non-neural factors common to both DRs. This further emphasizes the importance of comparing PPD and PA DRs in terms of ratio, rather than in terms of absolute DRs or in terms of loudness growth function.

On average, the PPD DR was 1.7 times larger than PA DR (3.6 dB), but with substantial inter-subject variability (Figs. 2 and 3). Test ears S16L and S16R exhibited much larger PPD DRs and a much poorer charge integration efficiency than the remaining test ears. These test ears had much longer duration of deafness (> 45 yrs) than the remaining test ears (range: 0.1–12 years), although implant experience (> 9.5 years in either ear) was comparable (range: 3.6–17.1 years). It is likely that the early onset of hearing loss and the long period of hearing deprivation resulted in substantial neural degeneration. The same long duration of deprivation would result in much more substantial neural degeneration with early versus late hearing loss onset (Hardie and Shepherd, 1999). Neural degeneration can result in a reduction in both neural density and function; both factors can reduce the auditory nerve’s ability to hold onto charge over long PPDs. A significant correlation was observed between duration of deafness and the charge integration efficiency when S16L and S16R data were included (p = 0.007), but not when these data were excluded (0.906). The lack of correlation among postlingually deaf test ears suggests that the proposed charge integration efficiency measure may be sensitive only to very large differences in neural health. Note that duration of deafness, as defined in the present study, represents the duration between diagnosis of severe-to-profound sensorineural hearing loss and cochlear implantation. Neural degeneration may begin long before diagnosis of severe-to-profound hearing loss, and may begin with the onset of mild to moderate hearing loss. While duration of progressive hearing loss (beginning at onset of mild or moderate hearing loss) relative to cochlear implantation may be a more appropriate demographic variable with which to characterize the degree of hearing deprivation, these data are sometimes difficult to reliably obtain. It would be interesting to compare charge integration efficiency with other measures that purport to reflect neural health, such as the effect of increasing inter-phase gap (Prado-Guitierrez et al., 2006; Ramekers et al., 2014; He et al., 2019).

4.2. Loudness growth with increasing PA or PPD

It is difficult to directly compare the present loudness growth data to those from Zeng et al. (1998) or Chatterjee et al. (2000) as a slightly different methodology was used in this study. Zeng et al. (1998) measured strength-duration functions at threshold and MAL for increasing PA (with fixed PPD) or PDD (with fixed PA), and derived a function to predict loudness according to the slopes of the threshold and MAL strength-duration functions. Chatterjee et al. (2000) measured loudness growth with increasing PA for a wide range of fixed PPDs. In this study, a common threshold anchor was used to more directly compare loudness growth functions in terms of the common unit of electric charge (nC). Despite differences in methodology, the present data are similar to those in Zeng et al. (1998) and Chatterjee et al. (2000), in that equal charge did not result in equal loudness, and that loudness generally grew more slowly with increasing PPD than PA.

The present method of measuring charge integration efficiency can quickly capture differences in loudness growth with increasing PA or PPD, since measuring DR necessarily incorporates the two extremes of a
Fig. 7. Individual charge levels for the PPD and PA stimuli to maintain equal loudness. The numbers indicate estimated loudness ratings extrapolated from the power function fits to the PA and PPD data from Fig. 4. Thus, each number represents a target loudness and the x- and y-axes represent the charge needed for the PA and PPD stimuli to maintain that loudness, respectively. The symbols represent the presentation levels for the loudness balance data. Thus, each symbol represents a target loudness and the x- and y-axes represent the charge needed for the PA and PPD stimuli to maintain that loudness, respectively.
loudestness growth function (threshold and MAL). However, note that loudness grew similarly in many cases for the lower portion of the DR. Across all test ears, the mean midway point of the PPD DR was 79.3 µs/ph, above which the PPD loudness growth functions started to diverge from the PA functions, needing greater charge for loudness to grow. This suggests that differential loudness growth with increasing PA or PPD may be observed for different portions of the DR. This observation is facilitated by using a common threshold anchor and by comparing the loudness growth functions in terms of charge.

4.3. Clinical and research implications

A significant portion of the variance in detection thresholds can be accounted for by electrode-neural interface and can vary across stimulation sites and/or cochlear implant patients, especially when using focused stimulation modes (e.g., Bierer et al., 2011). The present data suggest that charge integration efficiency may be potentially used to assess neural health at a stimulation site, independent of absolute detection thresholds. The present method is straightforward and could easily be implemented if clinical fitting systems were modified to allow PPD to be continuously increased (currently, only Oticon devices allow for this). Further research is needed to better understand the extent to which charge integration efficiency may explain variability in cochlear implant outcomes.

In Cochlear © devices (as tested in this study), 40–60 dB of the entire 120 dB acoustic input DR is compressed within the relatively small DR with electric hearing (typically 10–20 dB). For each channel of cochlear implant signal processing, changes in acoustic intensity are mapped within the electrode DR (“intensity coding”). Many contemporary implant devices (e.g., Cochlear© and Advanced Bionics) use short, fixed PPDs (e.g., ≤ 25 µs/ph) to accommodate high stimulation rates with multi-channel stimulation, and intensity is encoded by adjusting PA. Other devices encode intensity by varying PPD with a fixed PA (Oticon) or co-varying PA and PPD (MED-EL). To the extent that charge integration efficiency may reflect underlying neural health at a stimulation site, processors may be optimized according to PPD and/or stimulation rate (which is limited by PPD, inter-phase gap, and inter-pulse interval). For example, for sites with poor charge integration efficiency, short PPDs may be advantageous. For sites with good integration efficiency, longer PPDs may be used, which might reduce current spread (via lowering PA) and improve electrode selectivity. Currently, Oticon is the only device that offers explicit intensity coding with increasing PPD and a fixed PA. The slower loudness growth and a larger DR with PPD may translate into better intensity resolution. Further studies are needed to assess perceptual differences short or long PPDs, and whether PPD offers an advantage over PA for intensity coding.

4.4. Limitations to the study

The range of PAs and PPDs was markedly smaller than in previous studies where relatively high current and long phase durations were tested (e.g., Zeng et al., 1998; Chatterjee et al., 2000; Chatterjee and Kulkarni, 2014). In this study, monopolar stimulation and a relatively high stimulation rate were used to keep within the current CI clinical fitting parameters. As such, the DRs, loudness growth functions, and tradeoffs between PA and PPD are somewhat different from previous studies, although the general effects are similar (e.g., slower loudness growth with increasing PPD). However, it is interesting that the effects of increasing PPD could be observed within the relatively small range used in this study (range across all test ears: 25–198 µs/ph). To the extent that charge integration efficiency could be used to probe neural health across stimulation sites and/or CI patients, it is useful that PPD effects could be observed within the typical CI fitting parameter space. Also, only one mid-array electrode (EI 11) was tested in this study. However, our follow-up studies will further explore the differences in charge integration across sites and determine whether the mid-array electrode examined here would be representative of the entire array.

ChARGE integration was measured in a limited number of cochlear implant patients in this study. More data from cochlear implant users with a wider range of duration of deafness and sampling from a greater number of electrodes might elucidate whether charge integration efficiency may reflect the degree of neural degeneration across patients and/or stimulation sites. Charge integration efficiency might be especially illuminating in children, where early implantation may reduce auditory deprivation and slow neural degeneration.

4.5. Conclusions

In this study, we evaluated a new method to estimate charge integration efficiency at an electrode stimulation site in cochlear implant patients. Charge integration efficiency was defined as the dB difference in DR when PPD or PA was increased, anchored by a common threshold. Major findings include:

1. DR was significantly larger with increasing PPD than with PA, but with substantial variability across test ears.
2. Charge integration efficiency varied greatly across test ears, with a significant correlation between charge integration efficiency and duration of deafness, largely driven by two test ears with a long period of auditory deprivation.
3. In terms of charge, loudness grew more slowly with increasing PPD than with PA, particularly within the upper portion of the DR.
4. The present method may be a quick way to clinically evaluate charge integration efficiency and identify stimulation sites with poorer neural health and/or explain differences in cochlear implant outcomes across patients.

CRediT authorship contribution statement

Ning Zhou: Conceptualization, Methodology, Software, Formal analysis, Investigation, Resources, Data curation, Writing - original draft, Writing - review & editing, Visualization, Supervision, Project administration, Funding acquisition. Lixue Dong: Formal analysis, Investigation, Writing - original draft, Writing - review & editing. John J. Galvin: Conceptualization, Methodology, Formal analysis, Writing - original draft, Writing & editing, Visualization.

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References

Bierer, J.A., Faulkner, K.F., Tremblay, K.L., 2011. Identifying cochlear implant channels with poor electrode-neuron interfaces: electrically evoked auditory brain stem responses measured with the partial tripolar configuration. Ear Hear. 32, 436–444. Chatterjee, M., Kulkarni, A.M., 2014. Sensitivity to pulse phase duration in cochlear implant listeners: effects of stimulation mode. J. Acoust. Soc. Am. 136, 829–840. Chatterjee, M., Fu, Q.J., Shannon, R.V., 2000. Effects of phase duration and electrode separation on loudness growth in cochlear implant listeners. J. Acoust. Soc. Am. 107, 1637–1644. Hardie, N.A., Shepherd, R.K., 1999. Sensorineural hearing loss during development: morphological and physiological response of the cochlea and auditory brainstem. Hear. Res. 128, 147–165. He, S., Xu, L., Skidmore, J., Chao, X., Jeng, F.C., Wang, R., Luo, J., Wang, H., 2019. The effect of interphase gap on neural response of the electrically stimulated cochlear nerve in children with cochlear nerve deficiency and children with normal-sized cochlear nerves. Ear Hear. https://doi.org/10.1097/AUD.0000000000000815. Pfingst, B.E., Sutton, D., 1983. Relation of cochlear implant function to histopathology in monkeys. Ann. N. Y. Acad. Sci. 405, 224–239. Prado-Guitierrez, P., Fewster, L.M., Heasman, J.M., McKay, C.M., Shepherd, R.K., 2006.

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Effect of interphase gap and pulse duration on electrically evoked potentials is correlated with auditory nerve survival. Hear. Res. 215, 47–55.

Ramekers, D., Versnel, H., Strahl, S.B., Smeets, E.M., Klix, S.F., Grolman, W., 2014. Auditory-nerve responses to varied inter-phase gap and phase duration of the electric pulse stimulus as predictors for neuronal degeneration. J. Assoc. Res. Otolaryngol. 15, 187–202.

Rattay, F., 1989. Analysis of models for extracellular fiber stimulation. IEEE Trans. Biomed. Eng. 36, 676–682.

Shannon, R.V., 1985. Threshold and loudness functions for pulsatile stimulation of cochlear implants. Hear. Res. 18, 135–143.

Shepherd, R.K., Javel, E., 1999. Electrical stimulation of the auditory nerve: II. Effect of stimulus waveshape on single fibre response properties. Hear. Res. 130, 171–188.

Shepherd, R.K., Hardie, H.A., Baxi, J.H., 2001. Electrical stimulation of the auditory nerve: single neuron strength-duration functions in deafened animals. Ann. Biomed. Eng. 29, 195–201.

Smith, D.W., Finley, C.C., 1997. Effects of electrode configuration on psychophysical strength-duration functions for single biphasic electrical stimuli in cats. J. Acoust. Soc. Am. 102, 2228–2237.

van den Honert, C., Stypulkowski, P.H., 1984. Physiological properties of the electrically stimulated auditory nerve. II. Single fiber recordings. Hear. Res. 14, 225–243.

Zeng, F.G., Galvin 3rd, J.J., Zhang, C., 1998. Encoding loudness by electric stimulation of the auditory nerve. Neuroreport 9, 1845–1848.