35th Annual International Conference of the IEEE EMBS
Osaka, Japan, 3 - 7 July, 2013

Multipolar current focusing increases spectral resolution in cochlear implants*

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Abstract— **Cochlear implants are highly successful neural prostheses that restore hearing in the deaf, often resulting in high levels of speech understanding in quiet listening conditions. In more challenging conditions, however, cochlear implant subjects often score much lower than their normal-hearing peers, possibly reflecting limits of the electrode-neural interface. In this study, we compare monopolar stimulation versus focused stimulation, using multipolar channels, to test if current focusing can increase spectral resolution. Psychophysical results show that current focusing significantly improves subjects' ability to discriminate spectral features and detect dynamic modulations in sound stimuli. These results suggest that focused stimulation can successfully increase the number of effective channels with a cochlear implant and may lead to improved hearing in noisy conditions.**

I. INTRODUCTION

Cochlear implants (CIs) restore sound perception to patients with sensorineural hearing loss by bypassing the normal actions of the outer, middle, and inner ears and directly stimulating the auditory nerve with electrical currents. A series of electrodes are surgically placed along the tonotopic length of the cochlea, thereby allowing the conveyance of multiple sound frequencies at the different electrode sites. Many CI recipients experience high levels of speech understanding in favorable listening conditions and can converse on the telephone. However, in more challenging situations, such as listening to a talker in the presence of multiple competing voices, there can be large performance gaps between CI and normal acoustic hearing. While good speech understanding in quiet is possible with only a small (as low as 4) number of spectral channels in CI simulations with normal-hearing subjects [1], the addition of background noise requires additional spectral channels to maintain good speech scores [2]. It has been demonstrated that speech understanding with a CI improves with increasing number of electrodes until about 7-8 electrodes, where speech-in-noise performance saturates, while the performance of normal-hearing subjects listening to CI processing simulations increases up to at least 20 bands for the same material [3]. This is despite CI devices containing 12 to 22 electrodes, each coding a different part of the sound spectrum.

*Research supported by Cochlear Limited.

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One potential cause for the saturation of CI speech-in-noise performance at about 8 electrodes, which has also been observed in tests of spectral resolution [4], may be spread of excitation from the relatively broad current flow produced by the monopolar stimulation modes used in today's clinical CIs. Current spread may lead to large overlaps in the neural excitation patterns elicited by nearby electrodes, thus limiting the number of effective channels and spectral resolution. In this study we investigate current focusing techniques that attempt to limit current spread and excite more restricted populations of auditory neurons.

Previous investigations of spectral resolution, including the discrimination of spectral ripple phase and spectral ripple detection, have shown a high correlation between spectral sensitivity and speech understanding under various conditions (e.g. phonemes [5], speech in quiet [6], speech in noise [7]), thus making spectral resolution measures a potential tool for predicting speech outcomes with experimental CI processing strategies. In this study we used similar measures of spectral resolution to test the hypothesis that current focusing, using multipolar electrode configurations, increases spectral resolution in CI subjects compared to monopolar stimulation.

II. METHODS

A. Subjects and channels

Nine Percutaneous Contour Advance [8] cochlear implant recipients served as research subjects. All nine subjects participated in the first experiment; seven of these also participated in a second experiment, as the first two were no longer available for testing at the time of the second experiment. The percutaneous device allows for more flexible stimulation beyond what is available in the commercial device. For example, commercial devices from Cochlear Ltd have a single current source and the focused-multipolar channels tested here are not possible on these devices. Other electrode configurations are possible on Nucleus-brand implants, such as bipolar and common ground, but the focused-multipolar channels used here are expected to lead to more selective stimulation.

In these experiments, we compared two channel configurations: monopolar channels and focused-multipolar channels. Each monopolar channel delivers current via a

978-1-4577-0216-7/13/$26.00 ©2013 IEEE

single active electrode paired with a far-field return electrode. Focused-multipolar channels are based on subject-specific measures of voltage spread and use all available electrodes in each channel to focus current [8]. The focused-multipolar channels used in this study were further optimized to minimize simultaneous channel interactions in each subject [9]. Fig. 1 shows the average weighting, across all subjects, of stimulation current on each electrode for the focused-multipolar channels centered on electrode 12. This compares to the monopolar configuration where all current is passed between electrode 12 and an extra-cochlear return electrode (ECE1).

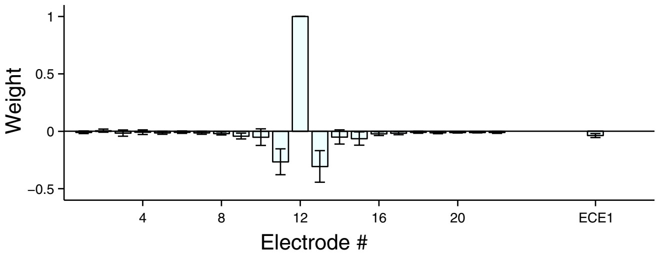

Figure 1. Across-subject weighting of stimulation current for focused-multipolar channels centered on electrode 12 (mean ± s.d.). ECE1 is an extra-cochlear electrode.

B. *Ripple stimuli*

In the first experiment, spectral ripple stimuli were used to measure spectral ripple phase discrimination thresholds. Random pink noise tokens were generated and their spectrums modified in the frequency domain with sinusoidal modulations applied across log-frequency modifying the log-amplitude (dB SPL) of the spectrum. Stimuli had a bandwidth of 6 octaves (120–7680 Hz), which approximately covers the full input frequency range of the CI processing, which filters incoming sounds into 22 frequency bands. Four ripple densities were tested: 0.25, 0.5, 1.0, and 2.0 cycles/octave. Example spectra are shown in Fig. 2 for spectral ripple stimuli with 0.5 cycles/octave ripple density and 10 dB ripple depth.

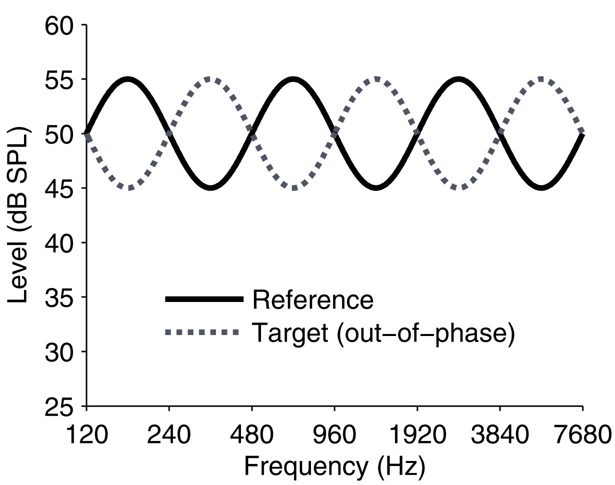

Figure 2. Example spectral ripple stimuli with a ripple density of 0.5 cycles/octave and ripple depth of 10 dB. These stimuli have a fixed spectrum thoughout the duration of each token. The baseline pink noise spectrum (1/*f*) has been removed in this figure for clarity.

In the second experiment, dynamic ripple stimuli were used to measure spectro-temporal modulation detection thresholds. Here, the baseline pink noise stimulus was a sum of random-phase sinusoids logarithmically spaced at 16 sinusoids/octave over the 6 octaves (120–7680 Hz) with magnitudes following the pink noise spectrum (1/*f*). To produce temporal modulations, each sinusoid was modulated with a log-amplitude sinusoid at a specified ripple velocity. Modulation phase across sinusoids was arranged to create the specified ripple density (spectral modulations). As in the first experiment, spectral ripples were applied across log-frequency. Ripple velocity in these stimuli was positive for ripple peaks increasing in frequency over time (upward moving) and negative for ripple peaks decreasing in frequency over time (downward moving). Fig. 3 shows examples of the combinations of ripple densities (0.5, 1.0, and 2.0 cycles/octave) and ripple velocities (±4, ±12, and ±32 Hz) tested. Note that in this figure only either upward or downward ripple velocities are shown for each density/velocity combination.

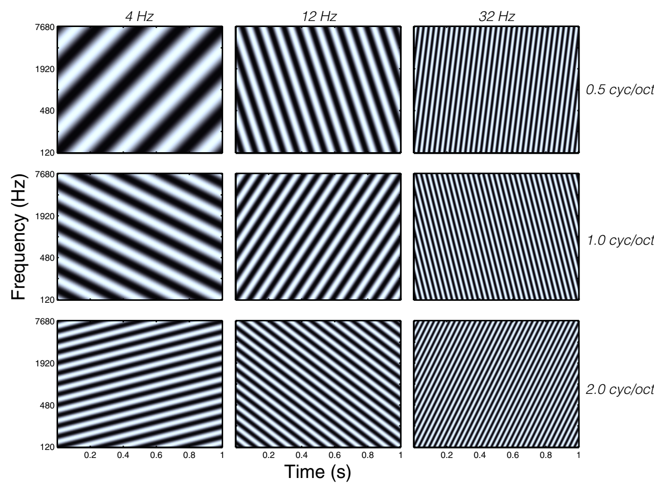

Figure 3. Example dynamic ripple stimuli. Left, center, and right columns have ripple velocities of 4, 12, and 32 Hz. Top, middle, and bottom rows have ripple densities of 0.5, 1.0, and 2.0 cycles/octave. The instantaneous spectrum of any of the 0.5 cycles/octave stimuli (top row) would be similar to that shown in Fig. 2.

C. *Psychophysical procedures*

In the spectral ripple phase discrimination experiment, a cued, three-alternative forced choice (cued 3AFC) task was used. Each trial consisted of four presentations of a spectral ripple noise stimulus, each with the same ripple density and ripple depth. Within each trial, reference stimuli had the same random ripple phase and the target stimulus was π radians out of phase. Subjects reported which of the last three intervals contained the target stimulus. In the dynamic ripple detection experiment, a 3AFC task was used with no cuing. Each trial consisted of three intervals, two of which contained steady noise (i.e. no ripples) and one with dynamic spectro-temporal ripples. For both experiments, a 2-down 1-up adaptive staircase procedure [10] with 8 reversals was used to adaptively find the ripple depth corresponding to 71% correct (3–4 runs per condition). For each run, the starting ripple depth was 20 dB and the step size was 0.5 on a $\log_2$(dB ripple depth) scale. Stimuli were processed by the ACE sound coding strategy [11] and were delivered using either monopolar or focused-multipolar channels.

D. Mapping

Stimulation levels for the different CI maps were set for each channel configuration separately. In Cochlear devices, sound levels between 25 dB SPL and 65 dB SPL within each frequency band are mapped between threshold (T) current levels and comfort (C) current levels following the loudness growth function (roughly linear on log-log sound pressure by current axes). Briefly, T-levels were set near perceptual threshold at a level just high enough that subjects could reliably count the number of stimuli. Once T-levels were found for all electrodes, C-levels were set just above this profile and speech at 65 dB SPL was presented through the map while increasing C-levels using equal clinical current level steps (log current scale) until the speech was perceived to be at a conversational level. Next, C-levels were checked across channels and adjusted to equalize any unevenness in loudness. Finally, speech was again presented through the map, and any final overall loudness adjustment was made. This process was used for both monopolar and focused-multipolar channels to create two maps per subject. For each subject, overall loudness was checked between maps and loudness balanced by making small adjustments to the C-level profile.

E. Data analysis

While stimuli were created with sinusoidal ripples in log amplitude (dB SPL) to better represent the statistics of natural sounds such as speech [12], results were analyzed as a function of modulation depth of the linear amplitude modulations, defined by equation (1):

$$md = 20 \cdot \log_{10}\left(\frac{p_{lin} - v_{lin}}{p_{lin} + v_{lin}}\right) \tag{1}$$

where *md* is modulation depth, p_{lin} is the linear amplitude of the modulation peak of the stimulus, and v_{lin} is the linear amplitude of the modulation valley of the stimulus. Using this definition, modulation depth is 0 dB for a fully modulated signal and increasingly negative for smaller modulations. In cases where modulation discrimination was impossible above 40 dB ripple depth (> -0.17 dB modulation depth), a modulation depth of 0 dB was used in the analysis. Note that ripple depths greater than ~40 dB approach and are asymptotic towards 0 dB modulation depth.

III. Results

Spectral ripple phase discrimination thresholds are shown in Fig. 4 for all nine subjects. Thresholds (dB modulation depth) for monopolar (closed squares) and focused-multipolar (open triangles) configurations are shown as a function of ripple density. As ripple density increases, thresholds increase for both configurations, indicating the increasing task difficulty at the higher ripple densities and the need for larger spectral modulations before subjects could discriminate spectral phase. With monopolar stimulation, one and three of the nine subjects could not perform the discrimination task at 1.0 and 2.0 cycles/octave, respectively, while all subjects could perform the task with focused-multipolar stimulation at all ripple densities tested. At the lower ripple densities, there is no significant difference in thresholds for the two configurations, but as the ripple density increases, monopolar thresholds increase more rapidly than focused-multipolar thresholds. Threshold is 3.3 dB lower at 1.0 cycles/octave (paired *t*-test, p=0.0587) and 5.7 dB lower at 2.0 cycles/octave (paired *t*-test, p=0.0060) for focused stimulation.

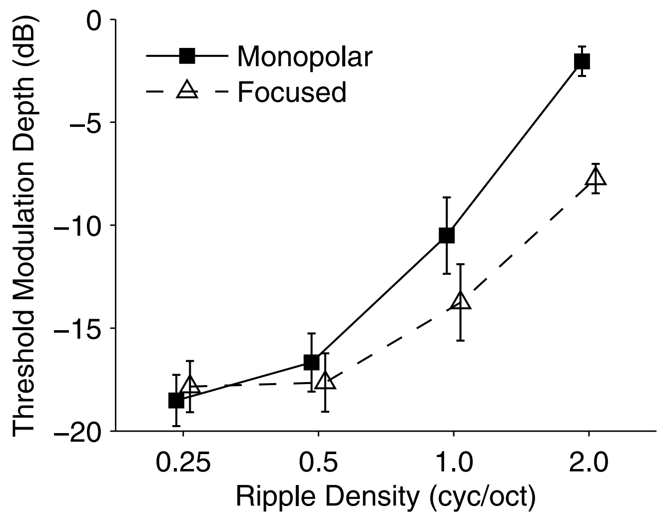

Figure 4. Spectral ripple phase discrimination thresholds (mean ± s.e.m.), as a function of ripple density, for monopolar (closed squares) and focused-multipolar (open triangles) channels. Data for each configuration in this figure are slightly offset horizontally for clarity.

Dynamic ripple detection thresholds are shown in Fig. 5 for the seven subjects tested. Thresholds (dB modulation depth) for monopolar (closed squares) and focused-multipolar (open triangles) configurations are shown as a function of ripple density in each panel for the three ripple velocities tested (4, 12, and 32 Hz). Data for upward and downward moving ripples were similar and were thus combined in the analysis and in the figure. As in the first experiment, thresholds increase with increasing ripple density, but here, thresholds also increase with increasing ripple velocity. Detection of the dynamic ripples was highly dependent on the electrode configuration used, with all thresholds lower for focused-multipolar compared to monopolar, and significantly lower for all cases except the combination of *density*=0.5 cycles/octave and *velocity*=4 Hz (paired *t*-tests, significance level p<0.050). A multi-factor analysis of variance was performed with subject, electrode configuration, ripple density, and ripple velocity as factors. While all factors had highly significant effects on detection threshold, the only factor that had a significant interaction with electrode configuration was ripple density (p=0.0012) and not ripple velocity (p=0.8614). This suggests that the increased modulation sensitivity from monopolar to focused-multipolar stimulation was based on spectral amplitude modulations rather than temporal amplitude modulations. When the detection threshold data is collapsed across ripple velocity, mean thresholds are 3.0, 4.9, and 5.9 dB lower for focused-multipolar stimulation at 0.5, 1.0 and 2.0 cycles/octave respectively.

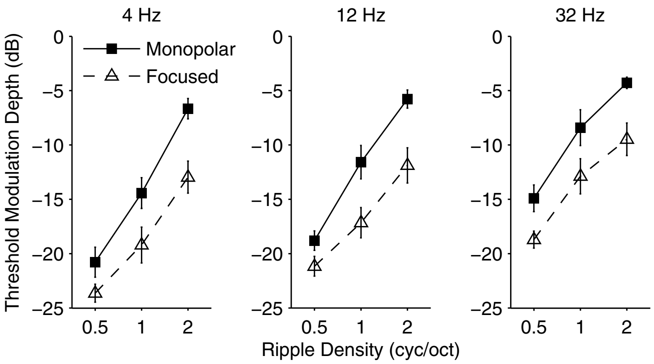

Figure 5. Dynamic ripple detection thresholds (mean ± s.e.m.), as a function of ripple density, for monopolar (closed squares) and focused-multipolar (open triangles) channels. Each panel shows threshold data for a different ripple velocity (left = 4 Hz, center = 12 Hz, right = 32 Hz).

IV. Discussion

In previous studies, current focusing techniques have been shown to reduce current spread in the scala tympani [8] and elicit more narrow neural excitation in both animal models [13] and human subjects [14]. In this study, optimized multipolar focused stimulation was added to a full CI sound coding strategy and resulted in significantly improved spectral ripple phase discrimination thresholds at the highest ripple density and lower detection thresholds of dynamic ripple stimuli across a range of ripple densities and velocities when compared to monopolar stimulation. This is in contrast to previous work, using partial tripolar channels in CI subjects, where only a very slight improvement in spectral resolution was observed [15].

The addition of temporal amplitude modulations in the second experiment, by using dynamic ripple stimuli, increased the effect of current focusing at the intermediate ripple densities (0.5 and 1.0 cycles/octave) when compared to the results of the first experiment with steady spectral ripples. The dynamic ripple stimuli are more relevant to speech since the spectrum of speech is dynamic (e.g. formants often shift frequency over time) and dynamic ripples can be used as basis functions to recreate speech spectrograms [16].

These results suggest that current focusing can increase spectral resolution in CIs, thus increasing the number of effective channels. This technique may also be helpful in other applications of neural stimulation where more selective stimulation is desired. Future work in take-home CI experiments is needed to further investigate focused stimulation and to test the hypothesis that increased spectral resolution will result in advantages for speech understanding in noise.

Acknowledgments

We thank Chris van den Honert for hardware and software contributions that made this work possible, Sara Nulle for assistance with data collection, and David Kelsall and Clough Shelton for surgical implantation and clinical care of the percutaneous subjects.

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