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Disorders of Peripheral Implications for the study of auditory adaptation to spectral shift: vowel spaces

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Abstract

Cochlear implant (CI) users differ in their ability to perceive and recognize speech sounds. Two possible reasons for such individual differences may lie in their ability to discriminate formant frequencies or to adapt to the spectrally shifted information presented by cochlear implants, a basalward shift related to the implant's depth of insertion in the cochlea. In the present study, we examined these two alternatives using a method-of-adjustment (MOA) procedure with 330 synthetic vowel stimuli varying in F1 and F2 that were arranged in a two-dimensional grid. Subjects were asked to label the synthetic stimuli that matched ten monophthongal vowels in visually presented words. Subjects then provided goodness ratings for the stimuli they had chosen. The subjects' responses to all ten vowels were used to construct individual perceptual "vowel spaces." If CI users fail to adapt completely to the basalward spectral shift, then the formant frequencies of their vowel categories should be shifted lower in both F1 and F2. However, with one exception, no systematic shifts were observed in the vowel spaces of CI users. Instead, the vowel spaces differed from one another in the relative size of their vowel categories. The results suggest that differences in formant frequency discrimination may account for the individual differences in vowel perception observed in cochlear implant users.

INTRODUCTION

Although cochlear implants allow profoundly deaf people to hear, cochlear implant users show a very wide range of speech perception skills. The most successful cochlear implant users can easily hold a face-to-face conversation, and they can even communicate on the telephone, a difficult task because there are no visual cues available and because the acoustic signal itself is highly degraded (Gstoettner et al., 1997). On the other hand, the least successful cochlear implant users have a difficult time communicating even in a face-to-face situation, and can barely perform above chance on auditory-alone speech perception tasks (Dorman, 1993).

One long term goal of our research is to understand the mechanisms that underlie speech perception by cochlear implant (CI) users and, in so doing, gain an understanding of the individual differences in psychophysical characteristics which may explain individual differences in speech perception with a CI. It is important to remember that electrical hearing as provided by a cochlear implant is quite different from normal acoustic hearing. One important difference lies in listeners' ability to discriminate formant frequencies. Kewley-Port and Watson (1994) report difference limens between 12 and 17 Hz in the F1 frequency region for highly practiced normal hearing listeners. In the F2 frequency region, they found a frequency resolution of approximately 1.5%. For cochlear implant users, formant frequency discrimination depends on two factors: the frequency-to-electrode map that is programmed into their speech processor, and the individual's ability to discriminate stimulation pulses delivered to different electrodes. It is not uncommon for some cochlear implant users to have formant frequency difference limens that are one order of magnitude larger than those of listeners with normal hearing, or even more (Nelson et al., 1995; Kewley-Port and Zheng, 1998). It is reasonable to hypothesize that cochlear implant users with such limited frequency discrimination skills will find it quite difficult to identify vowels accurately because formant frequencies are important cues for vowel recognition.

Another important difference between acoustic and electric hearing is related to the finding that cochlear implants do not stimulate the entire neural population of the cochlea but only the most basal 25 mm at best, because the electrode array cannot be inserted completely into the cochlea. Therefore, cochlear implants stimulate cochlear locations that are more basal and thus elicit higher pitched percepts than normal acoustic stimuli. For example, when the input speech signal has a low frequency peak (e.g., 300 Hz), the most apical electrode is stimulated, regardless of the particular frequency-to-electrode table employed. The neurons stimulated in response to this signal may have characteristic frequencies of 1000 Hz or even higher. This represents a rather extreme modification of the peripheral frequency map. To the extent that the auditory nervous system of CI users is adaptable enough to successfully "re-map" the place frequency code in the cochlea, the basalward shift provided by a CI should not hinder speech perception. On the other hand, an inability to adapt and re-map the place frequency code may severely limit speech perception in CI users and may be an important source of individual differences in speech perception (Fu and Shannon, 1999a, b). Although there is consensus in the literature that cochlear implants stimulate neurons with higher characteristic frequencies than those stimulated by the same sound in normal ears, there is controversy about the amount of this possible basalward shift. For example, Blamey et al. (1996), in a study where CI users with some residual hearing were asked to match the electrical percepts in one ear to the acoustic percepts in the other ear, concluded that the electrode positions that matched acoustic pure tones were more basal than predicted from the characteristic frequency coordinates of the basilar membrane in a normal human cochlea. However, Blamey et al. (1996) also acknowledge that "the listeners may have adapted to the sounds that they hear through the implant and hearing aid in everyday life so that simultaneously occurring sounds in the two ears are perceived as having the same pitch." On the other hand, Eddington et al. (1978) conducted the only experiment we are aware of, where a unilaterally deaf volunteer received a cochlear implant and was asked to match the pitch of acoustic and electric stimuli while still in the operating room, before he had much of a chance to adapt to the basalward shift in the way described by Blamey et al. (1996). Eddington et al. (1978) concluded that their pitch-matching data were "consistent with frequency versus distance relationships derived from motion of the basilar membrane."

Several previous studies have addressed the issue of adaptation to changes in frequency-to-electrode assignments for cochlear implant users. Skinner et al. (1995) showed that users of the SPEAK stimulation strategy identified vowels better with a frequency-to-electrode table that mapped a more restricted acoustic range into the subject's electrodes than the default frequency-to-electrode table. The experimental table that resulted in better vowel perception represented a more extreme basalward shift in spectral information than the default table, suggesting that listeners with cochlear implants can indeed adapt to such shifts, at least within certain limits. Another study that demonstrates the adaptation ability of human listeners in response to spectral shifts was conducted by Rosen et al. (1999), who used acoustic simulations of the information received by a hypothetical cochlear implant user who had a basalward spectral shift of 6.5 mm on the basilar membrane (equivalent to 1.3–2.9 octaves, depending on frequency). Initially, the spectral shift reduced word identification in normal-hearing subjects (1% correct, as compared to 64% for the unshifted condition), but after only three hours of training, subjects' performance improved to 30% correct. Whether or not such performance represents the maximum possible by CI users was not addressed in this study, given the relatively short time spent in training.

Recently, Fu et al. (submitted) performed an experiment in which the frequency-to-electrode tables of three cochlear implant users were shifted one octave with respect to the table they had been using daily for at least three years. It is important to note that this one-octave shift was in addition to the original shift imposed by the cochlear implant. After three months of experience with the new table, it was apparent that adaptation was not complete because, on average, subjects did not reach the same levels of speech perception that they had achieved before the table change. Taken together, these previous studies show that auditory adaptation to a modified frequency map is possible but it may be limited, depending on the size of the spectral shift that listeners are asked to adapt to.

In the present study, we investigated the adaptation of human listeners to a basalward shift using a new paradigm, a method-of-adjustment (MOA) procedure. This methodology was used to obtain maps of the perceptual vowel spaces of adult, post-lingually deafened cochlear implant users. Similar tasks have been used with normal hearing listeners as well as CI users (Johnson et al., 1993; Hawks and Fourakis, 1998). In this task, subjects select the region of the F1 – F2 plane that sounds to them like a given vowel, and the procedure is repeated for ten English vowels. This task simultaneously assesses a cochlear implant user's auditory adaptation ability, by comparing the locations of his/her selected regions to those selected by normal hearing listeners, and his/her frequency discrimination skills, by examining the spread of the selected regions. More specifically, a listener who was unable to adapt to the basalward spectral shift introduced by their cochlear implant would be expected to select regions whose centers are systematically shifted to lower frequencies with respect to the regions of the vowel space selected by normal hearing listeners in mapping their vowel categories. The extent to which cochlear implant listeners show relatively normal vowel category centers could be used as a measure of their adaptation to basalward spectral shift.

An alternative to this hypothesis predicts that the spread of selected regions (category sizes), as well as category centers, would increase (or "smear") as a result of perceptual adaptation. The resulting vowel categories may be larger (show greater spread) reflecting the cochlear implant listener's need to map a greater range of frequencies to a given vowel. To differentiate between the spectral smearing hypothesis and the frequency discrimination explanation would require a longitudinal study of the changes in vowel spaces and frequency discrimination. For the purposes of this study, a simple shift hypothesis limited to vowel category centers was tested and compared with the frequency discrimination hypothesis.

In addition to the MOA task, two other perceptual tests were administered to CI users, an F1 jnd test with synthetic vowel stimuli and a closed set identification test with naturally produced vowels. Taken together, these measures were intended to investigate the role of formant frequency discrimination and auditory adaptation in vowel perception by CI users.

EXPERIMENT

Methods

Participants

Forty-three Indiana University undergraduates with no reported history of speech or hearing problems and eight adult cochlear implant (CI) users, all monolingual speakers of English, participated in this experiment. The normal-hearing participants consisted of 20 males and 23 females ranging in age between 18 and 28. The normal-hearing participants were recruited to represent the dialect of American English spoken in central Indiana with a common inventory of vowels. Only normal-hearing listeners who reported living their entire lives in central Indiana were included in this experiment. Central Indiana was defined in terms of a 60-mile radius around Indianapolis, roughly covering the Midland dialect as described by Wolfram and Schilling-Estes (1998). This criterion was used to exclude two other regional dialects found at the northern and southern extremes of the state. These other regional dialects are reported to differ from the Midland dialect in terms of vowel quality and degree and type of dipthongization (Labov, 1991). For participating in two 1-h sessions, the participants received either $7.50 per hour or two credits towards their research requirement if they were enrolled in an undergraduate psychology class.

The CI users were recruited from the population of adult patients served by the Department of Otolaryngology-Head and Neck Surgery at the Indiana University School of Medicine in Indianapolis. The demographics of the CI users are given in Table I, while information concerning their cochlear implants is provided in Table II. All of the CI users were native speakers of American English, with the exception of CI1, who was a native speaker of British English. British and Midland American English are not reported to differ substantively from one another in vowel quality for the ten vowels used in this study (Gimson, 1962; Pilch, 1994). Thus the American English vowel spaces were deemed an acceptable benchmark for CI1 as well as the other CI users.

All of the CI users had received their cochlear implants at least one year prior to participating in this study. The CI users differed from one another in terms of the type of co-chlear implant they received: Five were users of the Nucleus-22 or Nucleus-24 device, programmed with either the SPEAK strategy or the MPEAK strategy, while three were users of the Clarion device, programmed with the CIS strategy. The SPEAK strategy (Skinner et al., 1994) filters the incoming speech signal into a maximum of 20 frequency bands, which are associated with different intracochlear stimulation channels. Typically, six channels are sequentially stimulated in a cycle, and this cycle is repeated 250 times per second. The channels to be stimulated during each cycle are chosen based on the frequency bands with the highest output amplitude. In contrast, the CIS strategy (Wilson et al., 1991) as implemented in the Clarion device filters the signal into eight bands, one for each stimulation channel. All channels are sequentially stimulated with pulses whose amplitudes are determined by the filters' outputs. The stimulation cycle is repeated at a rate of at least 833 times per second. The CIS strategy differs from the SPEAK strategy in its use of a different stimulation rate, fewer stimulation channels, and in its stimulation of all channels in a cycle rather than only a subset of the available channels.

The CI users also differed from one another in terms of the depth of insertion of the electrode array in the cochlea. The array's depth of insertion, in turn, determines the magnitude of the basalward spectral shift induced by the implant. It is possible to roughly estimate the size of this basalward shift for an individual CI user with three pieces of information: the location of the electrodes, the frequency to electrode mapping used by the cochlear implant's speech processor, and estimates of the characteristic frequency of the neurons stimulated by a given electrode pair. The speech processors of the participants in this study divide the acoustic frequency spectrum into channels. Each channel is specified by an acoustic frequency range that is assigned to a pair of intra-cochlear electrodes. Low frequencies are mapped to the apical electrodes, while high frequencies are mapped to the basal electrodes.

In the present study, the basalward spectral shift was calculated for two channels for every subject, the channels corresponding to 475 Hz and 1500 Hz. These frequencies correspond to the F1 and F2 of a neutral vowel for an average male speaker. First, the place of stimulation for a specific channel was defined as occurring half way between the electrodes for that channel. When stimulation was bipolar (i.e., both electrodes were intracochlear), the place of stimulation was defined as occurring half way between the electrodes for that channel. For patients who received monopolar stimulation (i.e., the return electrode was extracochlear) the location of electrical stimulation was considered to be at the intracochlear electrode. Second, the intraoperative report of insertion depth was used to adjust this place estimate based on the depth of electrode insertion. Cochlear lengths of the subjects were not measured individually. Instead, the basalward shift was calculated assuming that all subjects had average sized cochleas with a length of 35 mm (Hinojosa and Marion, 1983). Third, the electrically assigned frequency for this location was calculated as the geometric mean of the frequency boundaries defining the channel being studied in the speech processor. Finally, the characteristic frequency of the neurons stimulated by a given electrode was calculated from Greenwood (1961) and compared to the frequency calculated in step three. This discrepancy for each CI user, measured in Hz and octave, is shown in Table III. Alternatively, the shift is also reported in terms of the location difference between electrical and acoustic stimulation, in mm. The use of Greenwood's equation is based on the assumption that the average characteristic frequency of neurons stimulated by an electrode pair placed x mm from the round window is the frequency that would cause maximum displacement of the basilar membrane at the same x mm from the round window. Clearly, these are only rough estimates of the amount of basalward shift. In particular, the estimate of a 35-mm cochlea may lead to substantial overestimates or underestimates of the actual basalward shift. Future studies may improve the precision of these estimates by measuring the length of each individual cochlea using 3D reconstructions of CAT scans (Skinner et al., 1994; Ketten et al., 1998); by using the same 3D reconstructions to obtain more precise estimates of electrode location in the cochlea; and by obtaining physiological and behavioral data that may help determine the characteristic frequency of the neurons stimulated by different electrode pairs.

Stimulus materials

Method-of-adjustment task

The stimulus set consisted of 330 synthetic, steady-state isolated vowels that varied from one another in their first and second formants in 0.377 Bark increments. The vowels were generated using the Klatt 88 synthesizer (Klatt and Klatt, 1990). The Bark increment size was chosen as a close approximation of the just noticeable difference for vowel formants of Flanagan (1957). The F1 and F2 values for this stimulus set ranged between 2.63 Z (250 Hz)–7.91 Z (900 Hz) and 7.25 Z (800 Hz)–15.17 Z (2800 Hz). These ranges were chosen to represent the full range of possible values for speakers of American and British English, and were successfully used in piloting the present experiment and in an earlier method-of-adjustment study of vowel perception in normal-hearing listeners (Johnson et al., 1993). All of the other synthesis parameters for this stimulus set also followed Johnson et al. (1993). The formulas for calculating the values of the most relevant synthetic parameters are summarized in Table IV. The F0 parameter was varied to generate two sets of the 330 stimuli, one representing a male voice and one representing a female voice. All of the synthetic speech sounds were presented at a 70 dB C-weighted SPL listening level. The stimuli were presented over Beyer Dynamic DT-100 headphones for normal-hearing listeners, and over an Acoustics Research loudspeaker for CI users.

Vowel identification task

The vowel identification task employed a closed-set procedure that used nine vowels in an "h-vowel-d" format. The stimuli were digitized from the female vowel tokens of the Iowa laserdisc (Tyler et al., 1987). Only steady-state vowels (no diphthongs) were used from this stimulus set. There were three separate productions of each vowel. Listeners were administered three lists that consisted of five repetitions of each vowel and three practice tokens. The stimuli were presented at a level of 70 dB C-weighted SPL over an Acoustics Research loudspeaker.

Vowel identification task

The vowel identification task, administered only to the CI users, was a closed-set speech perception task in which three separate tokens of each of nine /hVd/ tokens were presented in random order, one at a time. The CI users had to say which one of the nine stimuli they thought they heard by responding verbally. They were instructed to guess if they did not know which vowel was presented. All subjects heard a total of at least 15 presentations of each vowel (except CI5 who heard 10). The subjects' responses were tabulated and scored for total percentage of correct responses.

F1 jnd task

The F1 jnd task, administered only to the CI users, required listeners to make an absolute judgment. The stimuli for this task were labeled "1" through "7" in order of increasing F1. In this task, all stimuli were played in sequence several times, so the subjects could become familiar with the stimuli. The stimuli were then presented ten times each in random order and subjects were asked to identify the stimulus that was presented using one of the seven responses. The subject's response and the correct response were displayed on the computer monitor before the presentation of the next stimulus. After each block of 70 trials (10 presentations of each of 7 stimuli), the mean and standard deviations of the responses to each of the 7 stimuli were calculated. The d′ for each pair of successive stimuli was calculated as the difference of the two means divided by the average of the two standard deviations. These d′ measurements were then cumulated to calculate a cumulative d′ curve, which provided an overall measure of the subject's ability to discriminate and pitch rank the seven stimuli (Durlach and Braida, 1969; Levitt, 1972). To calculate the cumulative d′ curves, we followed the common assumption that the maximum possible value of d′ was three. The average jnd (defined as the mean stimulus difference resulting in d′=1) was calculated based on the cumulative d′ curve. Given that the F1 range spanned by the seven stimuli was 600 Hz, the jnd was defined as (600/cumulative d′). At least eight blocks of 70 trials were administered in order for all subjects to reach a plateau in performance as measured by the cumulative d′. The cumulative d′ reported here is the average of the best two blocks for each subject. The normative mean cumulative d′ from this procedure was calculated to be 53 Hz, based on a pilot study with six normal-hearing listeners.

Results

Normal-hearing participants

The normal-hearing listeners were expected to select vowel category centers with first and second formant values that corresponded to those typical in vowel production in American English. Figure 1 shows the mean vowel categories obtained from the group of normal-hearing subjects, for the male-voice stimulus set.2 Each category center is shown along with error bars denoting the "size" of each category, that is, the relative spread of the category in both formant dimensions. The error bars represent the standard deviation from the category centers in both formant dimensions. In panel (A) of this figure, on the left, all of the ratings have been used to calculate the center and size of all ten vowel categories. Panel (B) of this figure, on the right, shows the vowel space of normal-hearing subjects calculated using only ratings of four and above. The rating of four was chosen because it was the highest rating that still allowed for category sizes to be calculated for all ten vowels of all of the normal-hearing and CI participants. The center of each category was determined by weighting the Bark values (in each dimension) of all synthetic stimuli that were chosen by their goodness ratings, and then averaging the weighted values.

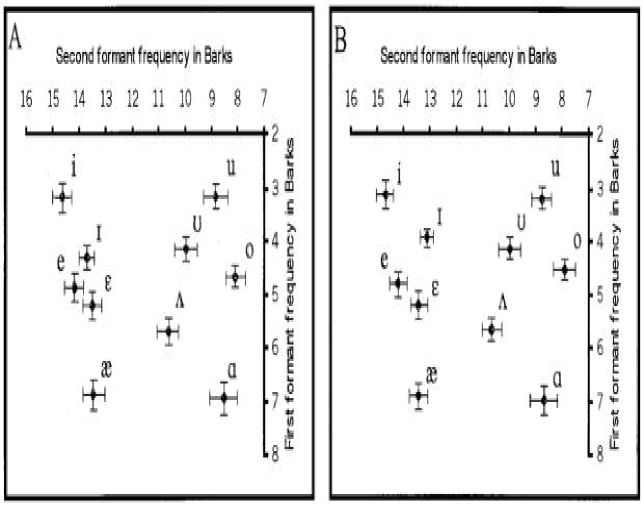

FIG. 1: The mean vowel space of normal-hearing listeners, calculated (a) using all of the ratings provided and (b) using only ratings of four or above.

The perceptual spaces shown in Fig. 1 for the normal-hearing listeners demonstrate that the method-of-adjustment technique for measuring vowel categories can be successfully used to generate vowel spaces that display the typical intervowel relationships that have been observed in F1 by F2 spaces generated from vowel production data (Peterson and Barney, 1952; Hillenbrand et al., 1995).

presentation condition, suggests to us that these children relied primarily on visual-spatial encoding of the target sequence to perform the task. These results were obtained despite the fact that many of these cochlear implant children did well on the auditory WISC digit span task and on the auditory-only presentation condition of the memory game.

In summary, the present results suggest that even those cochlear implant children who are able to accurately identify speech signals in isolation, may not have phonological working memory mechanisms or processing strategies that are developed to a point equivalent to chronologically age-matched normal-hearing children. This outcome would not exactly be surprising, as many important milestones in the development of speech perception and memory are reached during the first 2 yr of life (Aslin, Jusczyk, & Pisoni, 1998; Jusczyk, 1997). Despite their prelingually deafened status, most of the cochlear implant users reported on in this paper received their implant at a point in time when the FDA did not permit implantation of children under 2 yr of age. Additionally, because the implantation procedure requires that candidates show a demonstrated failure to benefit from conventional hearing aids, we can be fairly certain that most of these 8- and 9-yr-old children had received only minimal auditory input for at least one quarter to one third of their lives. It should not be surprising, then, that the encoding strategies and working memory mechanisms of pediatric cochlear implant users seem to differ measurably from those of normal-hearing children.

Ongoing research in our lab is attempting to describe in more detail how these encoding/rehearsal mechanisms differ, and what kind of developmental changes can be observed or effected in these children. Increasingly, clinicians are beginning to see pediatric cochlear implant users that have reached ceiling levels of performance on the traditional standardized measures of speech perception and spoken word recognition that are typically used with this population—and yet these children are still clearly having problems with reading and other more advanced language skills that are based on listening, phonological encoding, and other metalinguistic abilities. Further investigation of how pediatric cochlear implant users engage in cognitive processing of information originating from this reintroduced sensory input modality may help us develop new assessment and treatment techniques (Pisoni, 2000). Eventually we would like to answer the question of whether individual differences in the function of particular components of working memory within the pediatric cochlear implant pop ulation might have a meaningful causal relation to the level of verbal language skill attained by individual children. The present research begins to address this important issue because it provides some of the first behavioral data on working memory in pediatric cochlear implant users involving tasks in which the potential contribution of each available sensory modality was varied.

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