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

Improving Hearing Aid Fitting Using the Speech-evoked Auditory Brainstem Response*

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Abstract—**Measuring brain responses to speech may help improve the process of hearing aid fitting, especially in young children. Speech-evoked auditory brainstem responses (sABR) may be particularly useful because they provide a spectro-temporal representation of auditory neural activity in response to speech. However, use of the sABR in evaluating hearing aid performance has not been explored. This paper reviews recent work on measuring brain responses to speech, illustrates how sABR can provide insights into internal auditory processing, and proposes ways in which these responses may be used to improve hearing aid fitting.**

I. Introduction

Despite major advances in the flexibility and power of modern digital hearing aids, users report variable levels of satisfaction with their performance, with only around 50% of users satisfied in noisy environments (e.g. [1], [2]). Moreover, except for algorithms that improve directional focus, most signal processing approaches in these devices have provided very few gains in speech intelligibility when tested in the field. Surprisingly, this even applies to speech enhancement algorithms that apparently improve the signal-to-noise ratio (SNR) at the output of the hearing aid [3].

One of the problems associated with hearing aid use is that of obtaining a best fit, which involves adjusting multiple settings (e.g. frequency-dependent gain, compression related attack and release times) in the device in order to optimize the listening experience of the user. A common target is better speech intelligibility, but sound quality is also frequently taken into account. Hearing aid fitting is usually based on pure tone thresholds, if these can be determined behaviorally. However, pure tone thresholds are not well correlated with speech perception in noise [4]. Moreover, the fitting process (including fine-tuning) is a labor intensive task that requires significant interaction between the user and an audiologist, often necessitating multiple fitting sessions, and often yielding less than optimum results. This task is particularly difficult with infants and young children, where feedback from the user is limited or unavailable.

*Research supported by the University of Ottawa and the Natural Sciences and Engineering Research Council of Canada

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Given the difficulties involved in hearing aid fitting, particularly in young children and infants, there have been attempts to guide the fitting process by using the brain's electrophysiological responses to sound stimuli (e.g. [5]). Electrophysiological responses would complement perceptual measures, when these are available. In the ideal case, access to the activity of the auditory system at different stages, peripheral, brainstem, and cortical, could provide an effective objective approach to evaluating hearing aid performance. More broadly, decoding the internal electrophysiological activity of the auditory system could not only help to improve the process of fitting, but the entire process of hearing aid design as well.

As part of the general effort to develop brain computer interfaces and more specifically silent speech interfaces, there has been some recent research work to decode the brain's electrical activity related to speech perception. This has included attempts to determine imagined or heard speech with electrodes implanted on the surface on the cortex (e.g. [6], [7]), with magnetoencephalography or multiple-electrode electroencephalography (e.g. [8]-[10]), and with functional magnetic resonance imaging [11]. The research techniques used in these studies, which have met with variable success, are not widely available and currently not practical in the audiology clinic due to the associated invasiveness and/or technical complexity, as well as the associated costs.

In audiological practice, hearing thresholds can be estimated based on the auditory brainstem response (ABR) to tone bursts or more recently with amplitude modulated pure tones (auditory steady-state responses, ASSRs) [12]. Just as with the behavioral detection of pure tones, the mere presence of an ABR to such stimuli only suggests audibility of highly artificial sounds that do not relate to everyday sounds. Moreover, audibility does not necessarily imply capacity for good speech comprehension [13].

The measurement of cortical auditory evoked responses (CAEPs) to artificial stimuli such as tone bursts has also been investigated for a number of years in the evaluation of hearing aids, with some recent work described in [5] and [14]. However, the utility of CAEPs remains controversial because there is still little understanding of how hearing aid processing affects the responses. In particular, it appears that changes in stimulus characteristics, and in particular the SNR at the output of the hearing aid, can affect the CAEP in ways that do not obviously reflect changes in amplification [5].

At the core of the problem of hearing aid fitting is the general finding that measures based on the SNR at the output of the hearing aid do not correspond well to the intelligibility

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of speech [3]. Edwards [3] proposes that there must be an internal representation better suited to play this role. This paper explores the suitability of an internal representation in the form of the speech-evoked ABR (sABR) as a tool for hearing aid fitting.

II. The Outlook for Speech-evoked Auditory Brainstem Responses

A. The Speech-evoked Auditory Brainstem Response

The sABR is a signal that could aid in characterizing the internal representation of sounds after they are processed by the hearing aid. This signal represents the compound activity of populations of auditory neurons that follow, through phase-locking, different components of the speech stimulus. In fact, it is sufficiently "speech-like" that if it is played back as sound, it can be intelligible as speech [15]. Recent evidence suggests that different components of the sABR, relating to stimulus fine structure and envelope, may predict listening performance [13].

The sABR is typically measured with two surface electrodes and one ground electrode in response to a speech stimulus presented repeatedly and typically in alternating polarity, and with the subject usually not attending to the stimulus. The coherent average of an equal number of responses to the stimulus in one polarity and in its inverted polarity will follow the fundamental frequency of the envelope of the signal and its harmonics [16], [17], and so can be referred as an envelope following response (EFR). The EFR is appropriate for evaluating the components of the response at the fundamental frequency (F0) and its harmonics. If the coherent average is taken between the responses to one polarity and the negative of the responses to the inverted polarity, then the resulting response will be dominated by components that directly follow the harmonic content of the stimulus, and so can be referred to as a frequency following response (FFR). The FFR is appropriate for evaluating the components of the response at the formants and their surrounding frequencies. As a result, the sABR contains components that reflect the harmonic content of speech, including F0 and components in the region of the first formant (F1) and possibly the second formant (F2) if the frequency of F2 does not exceed the phase-locking limit of auditory neurons. The sABR can be distinguished from any electrical artifacts generated by the hearing aid based on the 5 – 10 msec stimulus to response latency which would not be found in the artifacts [17].

The sABR is thought to originate mainly from the upper brainstem [18] and so it does not simply replicate the spectro-temporal characteristics of the stimulus. Rather, the sABR conveys auditory processing that takes place in the periphery and the brainstem. For example, it may reflect phenomena such as formant capture, in which activity near the formant frequency is enhanced while that at frequencies surrounding it is suppressed. Figure 1 illustrates apparent formant capture in the sABR with the word /eardrum/ used as stimulus, recorded in a normal hearing subject (details of the recording methodology can be found in [17], [19], and [20]). As can be seen, the harmonic content of the response surrounding the first formant in /ear/ is suppressed (Fig. 1b) compared to the same harmonic content in the acoustic stimulus (Fig. 1a). Recently, we have described a similar phenomenon that we refer to as F0-capture and which occurs when a moderate amount of noise is added to the speech stimulus [20]. Another example of the effect of neural processing on the spectral content of sABR is the robust component found at F0, even when it is suppressed in the stimulus [21].

The sABR therefore offers the possibility of developing an internal representation based on the measured responses of neural populations at an intermediate stage of auditory processing. Because this representation is related to phase-locking in neural populations, but whose components can be differentiated depending on their tonotopic locations, it combines both place and synchrony auditory activity. Therefore, the sABR likely reflects major components of the auditory code in response to speech stimuli. Moreover, because sABR retains speech-like signal characteristics, it may allow detailed characterization of the effects of the complex processing performed by modern hearing aids, such as wideband and narrowband compression, and attack and release times of compression. This is not really possible and may not be very informative with the highly abstracted cortical responses nor with ABRs to non-speech stimuli [16].

B. Evaluating Hearing Aid Performance Using the Speech-evoked Auditory Brainstem Response

The use of sABRs to evaluate hearing aid performance has not been explored. In the literature, there is only one report on one subject in which it was demonstrated that the sABR varied with different settings of the hearing aid, with one particular setting producing the best stimulus to response

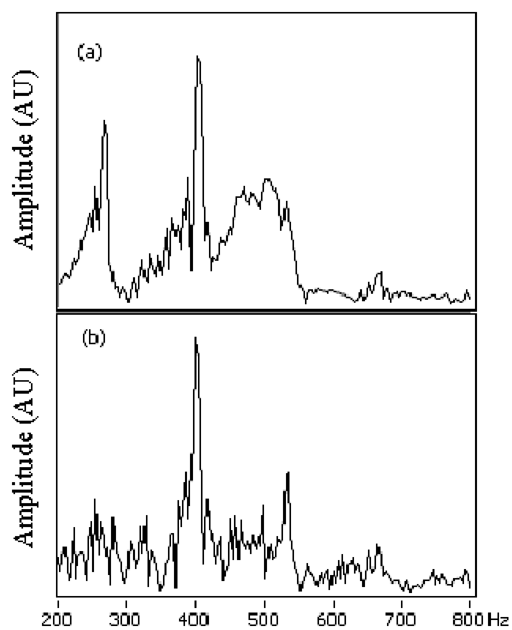

Figure 1. (a) Amplitude spectrum of the first portion of the word /eardrum/ (i.e. /ear/) spoken by a male, and focusing on the first formant (around 400 Hz). (b) Amplitude spectrum of the sABR to this stimulus in a normal hearing subject, based on coherently averaging responses to 10,800 stimulus repetitions. In both plots, the amplitude is in arbitrary units.

correlation [22]. In reality, the similarity of the response to the stimulus may not be the best indicator of the quality of the internal representation since, as mentioned earlier, the sABR reflects transformations in the auditory periphery and in the brainstem.

An important question is to identify the main features of the internal auditory representation, as reflected in the sABR, that could be used to assess the performance of the hearing aid. Response amplitude at the harmonics of the speech stimulus is one feature that is likely related to the perceptual salience of the stimulus. Therefore, changes in the harmonic amplitude could be useful for adjusting the gain of the hearing aid and its compression settings, as has been proposed with the ASSR to amplitude modulated pure tones [12]. The dependence of the amplitude of the sABR on stimulus level has not been previously studied. Fig. 2a shows the amplitude at F0 and the effective amplitude at the first 4 harmonics of F0 in the EFR in a normal hearing subject with a synthetic vowel /a/ stimulus presented at different intensities, while Fig. 2b shows the amplitude at F1 and the effective amplitude at the two harmonics surrounding F1 in the FFR. As can be seen, in this subject, the response amplitude at F0 and F1 did not experience the expected loudness growth. However, the effective amplitude at stimulus components near F0 and F1 did experience loudness growth, suggesting that the spectrum of the response is richer at higher levels.

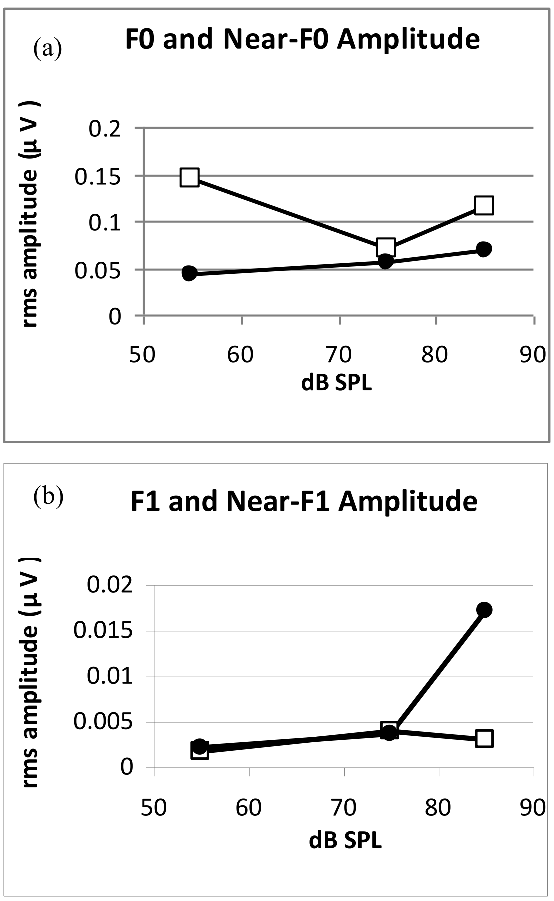

Figure 2. sABR amplitude with a synthetic vowel /a/ in a normal hearing subject as a function of stimulus level based on 4000 stimulus repetitions in each condition. The fundamental frequency F0 is at 100 Hz and the first formant F1 is at 700 Hz. (a) *Open Squares:* rms amplitude at F0. *Full Circles:*Effective rms amplitude at first four harmonics of F0. (b) *Open Squares:* rms amplitude at F1. *Full Circles:*Effective rms amplitude at the two harmonics surrounding F1.

The internal SNR of neural activity, which can be estimated in the sABR [20], is also expected to be correlated with the performance of the hearing aid. In a previous study, we measured the SNR of the harmonic components of the sABR with a synthetic vowel stimulus in different levels of background noise [20]. The SNR in the response was estimated based on the power at F0 or at F1 relative to neural noise at surrounding frequencies. Then based on the difference between the stimulus and response SNRs, we were able to estimate the internal SNR gain with different levels of background noise.

Another potentially useful feature in the sABR is the initial transient complex, whose morphology and latency is affected by the addition of background noise [23], and which has been shown to depend on the initial consonant in consonant-vowel speech stimuli [24]. It is likely that the latency of the initial transient of the sABR will also be affected by the compression time constants of the hearing aid [22]. Multi-band compression in hearing aids can also reduce the spectral contrasts between the peaks and troughs of the speech spectrum, while the rise and fall time constants of compression will the affect the depth of temporal modulation [3]. Both of these effects of compression can be studied in the sABR. Coding of temporal modulation in the auditory system can be studied at different rates, ranging from modulation at F0 to slower modulations at phoneme and word level, in which case the electrophysiological response is likely to originate at higher levels of the auditory system [25].

III. CONCLUSION

The use of the sABR in hearing aid fitting is promising since these responses can provide direct insights into auditory neuronal processing in hearing aid users. There is therefore a need to characterize sABRs in aided and unaided hearing impaired listeners, with various speech stimuli and in various noise conditions.

There are, however, known limitations to the use of sABRs to characterize hearing aid performance. One limitation already mentioned above is that sABRs are constrained by the upper frequency limit of phase-locking in auditory neurons, although higher frequency components can be differentially encoded via relative timing differences, which can be captured via the sABR [24]. Another limitation is that a recording time of several minutes is often required to obtain acceptable sABRs, when responses to individual stimulus repetitions are averaged coherently. With natural speech, even longer recording times may be required because stimulus energy is not concentrated at harmonic frequency components, unlike in synthetic steady-state vowels. Work is ongoing in our group to develop adaptive signal processing algorithms that utilize knowledge about the expected response and the background noise in order to reduce the required recording time [26].

Future work could also include the development of objective measures based on the response SNR, borrowed from other fields related to speech intelligibility, such as the articulation index (AI) and speech intelligibility index (SII) which have been proposed for hearing aid fitting [27]. These measures divide the signal spectrum into frequency bands, and weigh the SNR in each band depending on the importance of the bands for speech intelligibility. An analogous approach may allow the development of an objective "physiological" measure of speech intelligibility based on the weighting of the estimated internal SNR of sABR. This would require further research into the relative importance of the frequency components of the response for speech intelligibility and sound quality.

Finally, we proposed another approach to evaluating hearing aid performance using sABRs, one that would rely on the automatic classification of responses [28]. The assumption is that the hearing aid user would find it easier to perceptually identify or discriminate among different stimuli, such as confusable consonants in consonant-vowel syllables, when the hearing aid is tuned to produce evoked responses that can be clustered into distinct and maximally-separated classes. We speculate that this approach would be especially beneficial in individuals with profound hearing impairment, but this requires further investigation.

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