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CHAPTER 13

Auditory Steady-State Responses and Hearing Device Fitting

Part A

The Role of Auditory Steady-State Responses in Fitting Hearing Aids

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The advent of universal hearing screening programs has ensured early detection of hearing impairment at birth. This involves the fitting of hearing aids at the earliest stage possible. Hearing aid fittings constitute the most frequent method of hearing impairment habilitation in newborns and very young infants. Early hearing aid fitting in infants contributes to acquisition and development of oral language. For this reason, it is important to have hearing aid fitting protocols specifically designed for very young infants.

These protocols will be dependent on electrophysiological methods, because behavioural audiometry is not viable until the age of 5 to 6 months and, in some infants or young children with developmental delay, not possible at all.

In infants and in the adult population, hearing aid fitting consists of three stages: the assessment of hearing sensitivity, the selection of adjustment parameters to restore the hearing perception, and the verification of the prescribed gain for each patient (Scolie & Seewald, 2001).

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Assessing hearing implies the establishment of hearing thresholds and maximum comfort and discomfort levels at different frequencies for each ear independently. The prescription of adjustment parameters requires the establishment of the gain in such a way that the speech spectrum (Cornelisse, Gagné, & Seewald, 1991; Zenker, Delgado, & Barajas, 2003) is amplified within the dynamic hearing range of the patient (Cornelisse, Seewald, & Jamieson, 1995; Zenker & Barajas, 1999). Finally, the verification of amplification allows the clinician to check that the objectives set in the prescription of adjustment parameters have been achieved (Stelmanchiwicz, Kopun, Mace, Lewis, & Nittrouer, 1995).

For certain difficult-to-test populations, hearing thresholds can be obtained only through electrophysiological measures that do not require any voluntary response from the individual (Goldstein & Aldrich, 1999). Moreover, electrophysiological tests can assist those involved in the adaptation of hearing aids, because these tests can measure auditory function objectively (Picton et al., 2002). Assessment of auditory steady-state responses (ASSRs) provides a quick and objective way to establish electrophysiological hearing thresholds at different frequencies. The ASSRs have several advantages in their application in hearing aid fitting: First, they provide assessment of hearing thresholds at different frequencies. Second, from these measurements it is possible to infer the adjustment parameters of hearing aid devices. Third, the acoustic characteristics of the ASSR's stimuli allow verification that the hearing aid is functioning and that the subject perceives and discriminates sounds at a brain level (Picton et al., 2002).

Recent studies (Cone-Wesson, Parker, Swiderski, & Rickards, 2002; Picton et al., 1998; Zenker, Fernández, & Barajas, 2006) have proposed that assessment of auditory evoked potentials, and specifically ASSRs, could serve as an useful tool in the fitting and verification of the function of hearing aids. The application of electrophysiological techniques in hearing aid fitting requires measures of (1) the stimuli, (2) the acoustic characteristics of external auditory canal, and (3) the frequency-specific hearing threshold assessment. These aspects are reviewed in the following sections.

Estimating Pure-Tone Audiometric Threshold Through Auditory Evoked Potentials

Hearing aid amplification is based on the individual characteristics of each patient. In this regard, individual amplification is established by gain prescription methods. Cornelisse and colleagues (1995) defined this prescription process as "a function that prescribes hearing devices' gain at different frequencies related to the patient's audiometric values." The first gain prescription methods were established by Watson and Knudsen (1940) and later by Lybarger (1944). At present, prescription methods can be provided either by hearing aid manufacturers or through independent procedures such as the "Desired Sensation Level" (Seewald, Ross, & Spiro, 1985) or the National Acoustic Laboratory approach (Byrne & Tonnisson, 1976).

The manufacturer modules for fitting hearing aids require knowledge of the

patient's hearing sensitivity. Once this information is obtained from electrophysiological procedures, an estimation of behavioural thresholds must be carried out. In the case of ASSRs, the estimation of hearing thresholds can be influenced by certain measurement conditions such as the stimulus frequency, the degree of hearing loss, the age of the subject, and the duration of the electrophysiological test. Some studies indicate that the physiological thresholds may be more accurate in hearing-impaired individuals than in normal-hearing subjects (Dimitrijevic et al., 2002; Rance & Briggs, 2002). Two main procedures are used to estimate audiogram thresholds: The first consists of obtaining the average difference between ASSRs and behavioural thresholds; the second procedure consists of determining the regression function between the physiological and the psychoacoustic thresholds for different values of hearing loss (Rance & Briggs, 2002; Rance et al., 1995). The ways in which behavioural hearing thresholds can be predicted from ASSR findings are discussed in detail in Chapter 7.

Frequency Specificity

The adjustment of hearing aids requires frequency specific information at hearing threshold level. The thresholds obtained from the ASSR are at least as accurate and have the same frequency-specificity as that for tone-burst evoked auditory brainstem responses (ABRs) (Herdman, Picton, & Stapells, 2002). Moreover, they offer the advantage that thresholds for several carrier frequencies can be measured simultaneously (Lins & Picton,

1995). Up to eight thresholds can be obtained in a significantly shorter time than is typical with use of techniques based on sequential testing using one stimulus at a time (John et al., 2002).

Stimulus Calibration

The prescription of amplification from behavioural testing must take into account the calibration of the stimuli employed. In this regard, electrophysiological and behavioural responses can significantly differ for the same patient. These dissimilarities can be ascribed to differences in the type of stimulus and physiological generators involved in the detection of tone.

Modulated tones used during ASSR testing are similar to warble tones used in behavioural testing (Rance et al., 1998). This implies that calibration corrections associated with tone burst and clicks (used for ABR) are not required, and the stimuli can therefore be presented at levels that extend to 120 dB HL. Calibration of ASSR stimuli can be carried out with the same standards used for pure tones employed in audiometers. This procedure makes it possible to obtain the thresholds in dB HL and thereby to directly introduce these values into hearing aid fitting modules. It should be noted that the values used in hearing aid fitting algorithms are based on behavioural thresholds. An additional step, therefore, is needed to convert the ASSR thresholds (in dB HL) to estimated behavioural thresholds (also in dB HL). This can be done by using regression formulas or correction factors. These methods and their benefits and limitations are reviewed in Chapter 7.

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Ear Canal Acoustics

Sininger and coworkers (1997) have demonstrated that part of the differences found between the physiological and behavioural thresholds in infants and adults may be due to the large ear canal resonances associated with small infant ear canals. The effect of the ear canal resonance on the gain prescription of the hearing aid can be minimized by the measurement of the real ear to couple difference (RECD). Bagatto and associates (2005) proposed the following equation to obtain the finally estimated dB SPL thresholds at the ear canal level.

$$\begin{aligned} \text{dB SPL threshold (ear canal level)} = \\ \text{dB HL threshold} + \text{insert earphone} \\ \text{RETSPL} + \text{RECD} \end{aligned}$$

Table 13-1 provides an example of how hearing threshold, in dB SPL, is calculated from ASSR threshold. In this case, the estimated behavioural thresholds at the eardrum in dB HL can be converted into dB SPL by using the *real-ear coupler-difference* (RECD) plus the *reference equivalent threshold sound pressure levels*

(RETSPLS) according to ANSI S3.6-1996. In the example case, values reflect the RECD for a newborn 6 months of age.

Applications of Auditory Evoked Potentials in Hearing Device Fittings

Several studies have proposed the application of electrophysiological techniques at the different stages of hearing aid fittings. An essential contribution of the electrophysiological technique consists of the verification of the prescribed gain and the calculations of the adjustment parameters of the hearing aid derived from the amplitude or latency of the AEPs.

The Amplitude Projection Procedure

Kiessling (1982, 1983) published several articles describing the clinical applications of the ABRs in hearing aid fitting. He established an objective method based on the amplitude of the ABR especially indicated for noncooperative subjects

Table 13-1. Example worksheet to calculate the hearing threshold, in dB SPL, at the ear canal from the ASSR thresholds in dB HL

	500 Hz	1000 Hz	2000 Hz	4000 Hz
Thresholds (dB HL) (estimated from ASSR)	40.0	45.0	45.0	50.0
RECD	6.0	13.0	14.0	18.0
RETSPL correction	5.5	0.0	3.0	5.5
dB SPL threshold (ear canal)	51.5	58.0	62.0	73.5

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and infants with multiple disabilities. The *amplitude projection procedure* (APP) is an attempt to establish the adjustment parameters of hearing aids such as the dynamic range, gain, or compression factor. These parameters are derived from the wave V amplitude intensity function of the ABRs. The methodology is based on the assumption proposed, and never demonstrated, by Kiessling, that wave V amplitude of the ABR correlates with the sensation of loudness. The steepness of the amplitude intensity function of the ABR described the amount of compression needed for a listener with hearing loss. For example, a patient with a severe-to-profound hearing loss will have elevated ABR (or ASSR) thresholds, but the amplitude growth of the response above threshold will be minimal. This would correspond to the limited dynamic range of hearing, that is, the difference between threshold and uncomfortably loud sounds, associated with severe-to-profound hearing loss. In such cases, greater compression would be indicated, compared with that for the patient who has a mild-to-moderate loss, with a considerably greater dynamic range, and who demonstrates some growth of ABR (or ASSR amplitude) over a 30- to 40-dB stimulus range. In a subsequent study, Davidson, Wall, and Goodman (1990) studied the relationship between loudness and the ABR wave V amplitude. In this study, great individual variation in the amplitude was found. As expected, variability was reduced as soon as the number of sessions increased. This is because variability in background EEG noise causes great variability in the ABR amplitude. If background noise levels are held constant (and low), such as by measuring the estimated noise from a statistic like F_{sp} (Don

et al., 1996), then ABR amplitude data become less variable and more reliable. Davidson and colleagues reported no significant differences in the prescription of the hearing aid gain and compression settings for two out of three hearing-impaired subjects when the APP fitting procedure was compared with conventional prescription methods. Given that hearing aid compression technologies today are much different than those existing in 1990, it may be difficult to replicate this result. Yet the underlying physiological principles are sound: The evoked potential threshold can be used to estimate the amount of gain needed, and the amplitude growth function provides information relevant to the listener's dynamic range of hearing, for which compression can be used as compensation.

Electrophysiological Gain Verification

Functional gain is one of the most used procedures in the verification of the prescribed parameters of hearing aids. Measurement of the amount of functional gain is calculated as the difference between aided and unaided thresholds at each specific frequency obtained through free field testing (Mueller, Hawkins, & Northen, 1992) and is defined as the relative decibel difference between the aided and unaided thresholds. Because this technique is based on voluntary behavioural procedures, the inherent degree of variability to which behavioural thresholds measurements are subjected will also influence functional gain measurements (Mueller et al., 1992). Interaction between the test stimuli, the transducer employed, the hearing aid and the test

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room acoustics are well known limitations that interfere with the accurate determination of hearing thresholds. Moreover, verifying the hearing aid in the free field does not provide ear specific information. Several attempts have been made in order to reproduce the functional gain test through electrophysiological procedures. Electrophysiology testing has been seen as a procedure that may provide an objective method to verify the adjustment of a hearing aid.

Mokotoff and Krebs (1976) published a pioneer study in which ABR measurements were obtained from adult hearing aid users. Results indicated that aided ABR measurement compared favorably with aided audiological data. Several authors have used the ABR with click or tone burst in order to obtain an objective measurement of the hearing aid response. As described by Mahoney (1985), however, obtaining ABR in the sound field with amplification is a complex issue. In general, the brief nature of the stimulus employed during ABR testing shows high susceptibility to distortion in both: the sound field speaker and the hearing aid amplifier (Hall & Ruth, 1985).

Picton and colleagues (1998) demonstrated that ASSRs can be recorded when multiple stimuli are presented simultaneously through a sound field speaker and amplified with a hearing aid. Therefore, this procedure seems more useful than procedures using transient stimuli. Its application is limited, however, because ASSR cannot provide information regarding how well the nonlinear processing of the hearing aid is benefiting the listener (Picton et al., 2002). Even though ASSR can be used to assess suprathreshold hearing, it does not provide information regarding to how the aided sound is perceived, following processing in the brain (Picton et al., 2002).

Fitting Hearing Aids from the Auditory Steady-State Response Testing

Loudness and Auditory Steady-State Responses

Loudness measurement serves two important clinical functions in audiological practice: to determine the adjustment of hearing aids (Fabry & Schum, 1994) and to distinguish the site-of-lesion in sensorineural hearing loss (Hall, 1991). Subjective judgements of loudness are often obtained to define the *most comfortable level* (MCL) or the *loudness discomfort level* (LDL). Another method often employed is to obtain loudness judgement over a range of stimulus levels providing information on how listeners perceive the growth of loudness. Finally, a loudness growth function can be derived from the intensity values of the stimulus by the loudness magnitude.

Nonlinear circuits found in digital or wide dynamic range compression (WDRC) hearing aids have incorporated new fitting strategies that provide information about the loudness growth function over the range of intensities amplified (Serpanos, O'Malley, & Gravel, 1997). The adjustment of non linear hearing aids involves the concept of loudness growth normalization, where hearing aid features would be adjusted for a particular hearing loss in order to normalize the perception of loudness (Byrne et al., 2001). From this kind of loudness judgment it is possible to derive electroacoustic characteristics of hearing aids such as average gain, maximum output, compression ratio and onset level (Kiessking, 1982). Much effort has been made in order to develop adequate methods to measure

loudness, and a series of different loudness scaling procedures have been especially proposed for hearing aid fitting, rather than for diagnostic purposes.

Objective methods for estimating loudness growth have been proposed using electrophysiological measures. Several studies have revealed that loudness growth could be estimated using click-evoked ABR (Davidson et al., 1990; Galambos & Hecox, 1977; Picton et al., 1977; Thornton, 1987). In these studies, the wave V latency (Rosenhamer et al., 1981), the slope of the latency-intensity function (Galambos & Hecox, 1977; Picton et al., 1977; Thornton, 1987), interaural latency differences (Rosenhamer et al., 1981), and thresholds of the ABR (Conjin, Brocaar, & van Zanten, 1990) have been used as indicators of loudness. ABR amplitude has not been used as latency since it has greater intersubject variability (Schwartz, Morris, & Jacobson, 1994). The major disadvantage of the procedure based on ABR measurements is the lack in frequency specificity of these responses.

Assessment of ASSRs partly overcomes the limitations of ABR testing. Because the amplitude of the ASSR decreases as the intensity of the stimulus decreases, ASSRs can be used as an indicator of loudness. The major limitation of this procedure is the variability of the amplitude of the ASSRs, because recordings at a given intensity vary from subject to subject. The recorded amplitude of the ASSRs depends on several parameters. Among those the most important are the amount of synchronized current in the generators, the orientation of these generators in relation to the recording electrodes, and the impedance of the volume conductor.

An electrophysiological measure of loudness growth could assist audiolo-

gists in estimating discomfort levels and determining hearing aid features. Objective measurement of loudness could be included in the prescription of gain in order to fit hearing aids within the first few month of age.

In a recent contribution of our group (Zenker, Barajas, & Fernández, 2005), an attempt was made to prove whether it is possible to establish a relationship between subjective loudness growth derived from the contour test and the physiological responses obtained from the ASSR. The *contour test* is a clinical method to quantify loudness perception (Cox, Alexander, Taylor, & Gray, 1997). This test was designed to develop a function that describes the growth of loudness perception as input levels increase from near threshold to uncomfortably loud levels. In the contour test, tonal stimuli were 5% warble tones presented at .5, 1, 2, and 4 kHz. Verbal judgement from the subject of the perceived loudness was required by rating the loudness in seven categories ranging from very soft to uncomfortably loud.

The amplitude of the ASSR for each level as a function of frequency is illustrated in Figure 13-1.

There are differences in the slope of the amplitude growth function as a function of frequency. The slopes of the functions vary from 0.002 $\mu\text{V}/\text{dB}$ (right ear, 4000 Hz) to 0.0005 $\mu\text{V}/\text{dB}$ (left ear, 1000 Hz). (Right-versus-left ear ASSR amplitude differences are observed among normal hearing listeners [John, personal communication, 2007], but the application of this knowledge to clinical diagnosis is yet unclear.) Not only do the slopes vary with frequency, but overall amplitude does as well. The largest amplitude responses are found for 500 and 4000 Hz (at 80 dB HL). The variability between

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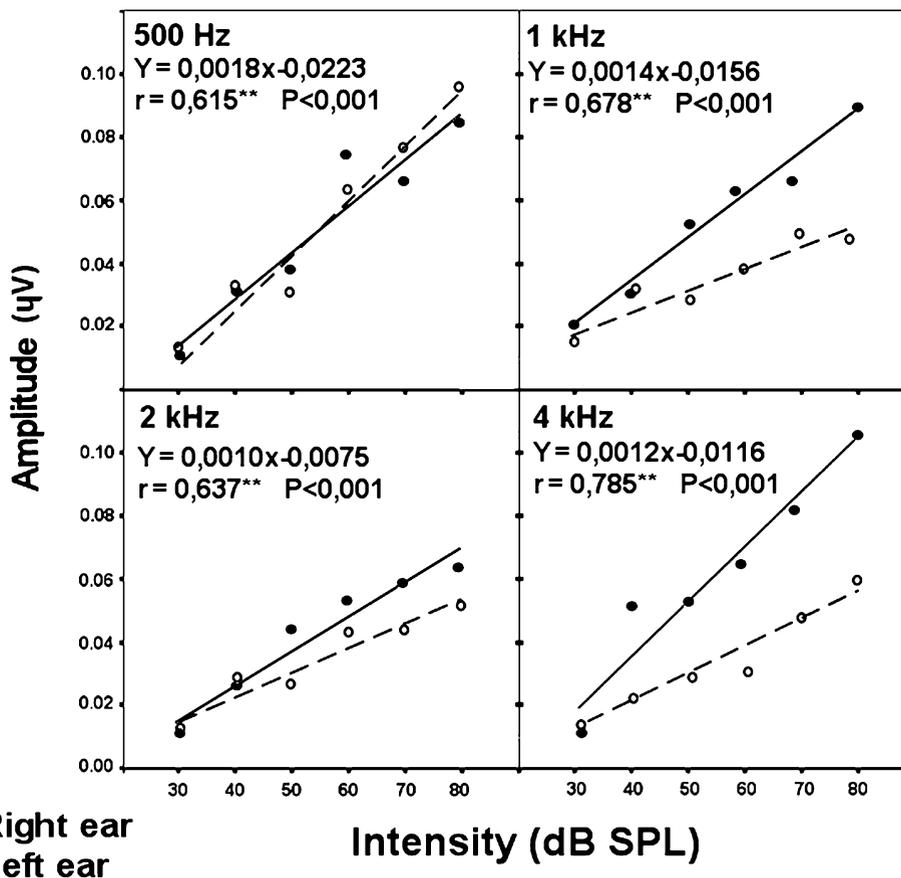


Figure 13-1. ASSR amplitude as a function of level in normal-hearing subjects. Amplitude growth functions are derived by means of linear regression. As expected, the amplitude of the responses increases with increasing level above threshold.

subjects was fairly similar for a given stimulus level across carrier test frequencies. The variability increased as the intensity level increased. Table 13-2 provides the mean amplitude (and standard deviation) of ASSR amplitude as a function of frequency, ear, and level.

Figure 13-2a shows the amplitude spectra of ASSRs and the loudness growth functions (Figure 13-2b) (from the contour test) obtained in a 29-year-old subject with normal hearing. The amplitude growth functions as a function of frequency and the loudness growth functions are correlated.

A multiple regression analysis was performed using the data from the contour test (sensation of loudness), level of the stimulus and amplitudes of the ASSR. The multiple regression analysis resulted in a prediction of loudness estimated from the ASSR amplitude and amplitude growth functions (Figure 13-3a). This relationship can be defined by the equation

$$Y(f) = B_0 + B_1 * \text{level} + B_2 * \text{AMPLITUDE}$$

where B_0 = y-intercept

B_1 = level (dB HL) coefficient

B_2 = ASSR amplitude coefficient

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Table 13-2. Mean amplitude and standard deviations of ASSRs for each carrier frequency and each ear (R and L). The corresponding modulation rate for each carrier frequency also is indicated. Response amplitude increases as the stimulus intensity increases for all carrier frequencies.

Intensity (dB HL)	ASSR Modulated Tones (Hz)											
	500		1000		2000		4000		500		1000	
	R (81 Hz)	L (77 Hz)	R (89 Hz)	L (85 Hz)	R (97 Hz)	L (93 Hz)	R (105 Hz)	L (101 Hz)	R (81 Hz)	L (77 Hz)	R (89 Hz)	L (85 Hz)
80	0.120 (0.089)	0.759 (0.065)	0.074 (0.040)	0.056 (0.034)	0.055 (0.031)	0.077 (0.049)	0.124 (0.066)	0.122 (0.066)	0.120 (0.089)	0.759 (0.065)	0.074 (0.040)	0.056 (0.034)
70	0.110 (0.081)	0.842 (0.068)	0.064 (0.028)	0.056 (0.030)	0.053 (0.024)	0.050 (0.035)	0.084 (0.045)	0.063 (0.036)	0.110 (0.081)	0.842 (0.068)	0.064 (0.028)	0.056 (0.030)
60	0.093 (0.090)	0.065 (0.070)	0.056 (0.023)	0.050 (0.027)	0.047 (0.025)	0.054 (0.031)	0.067 (0.037)	0.059 (0.037)	0.093 (0.090)	0.065 (0.070)	0.056 (0.023)	0.050 (0.027)
50	0.070 (0.067)	0.052 (0.064)	0.047 (0.025)	0.039 (0.025)	0.053 (0.022)	0.043 (0.018)	0.041 (0.020)	0.043 (0.025)	0.070 (0.067)	0.052 (0.064)	0.047 (0.025)	0.039 (0.025)
40	0.044 (0.039)	0.029 (0.030)	0.035 (0.020)	0.026 (0.019)	0.037 (0.017)	0.038 (0.017)	0.030 (0.018)	0.031 (0.019)	0.044 (0.039)	0.029 (0.030)	0.035 (0.020)	0.026 (0.019)
30	0.036 (0.030)	0.020 (0.015)	0.026 (0.019)	0.024 (0.015)	0.033 (0.019)	0.028 (0.018)	0.047 (0.086)	0.025 (0.014)	0.036 (0.030)	0.020 (0.015)	0.026 (0.019)	0.024 (0.015)
20	0.013 (0.011)	0.016 (0.010)	0.025 (0.013)	0.018 (0.011)	0.022 (0.015)	0.021 (0.012)	0.020 (0.009)	0.024 (0.013)	0.013 (0.011)	0.016 (0.010)	0.025 (0.013)	0.018 (0.011)
10	0.020 (0.012)	0.017 (0.014)	0.0218 (0.014)	0.020 (0.009)	0.015 (0.009)	0.014 (0.009)	0.022 (0.012)	0.014 (0.009)	0.020 (0.012)	0.017 (0.014)	0.0218 (0.014)	0.020 (0.009)

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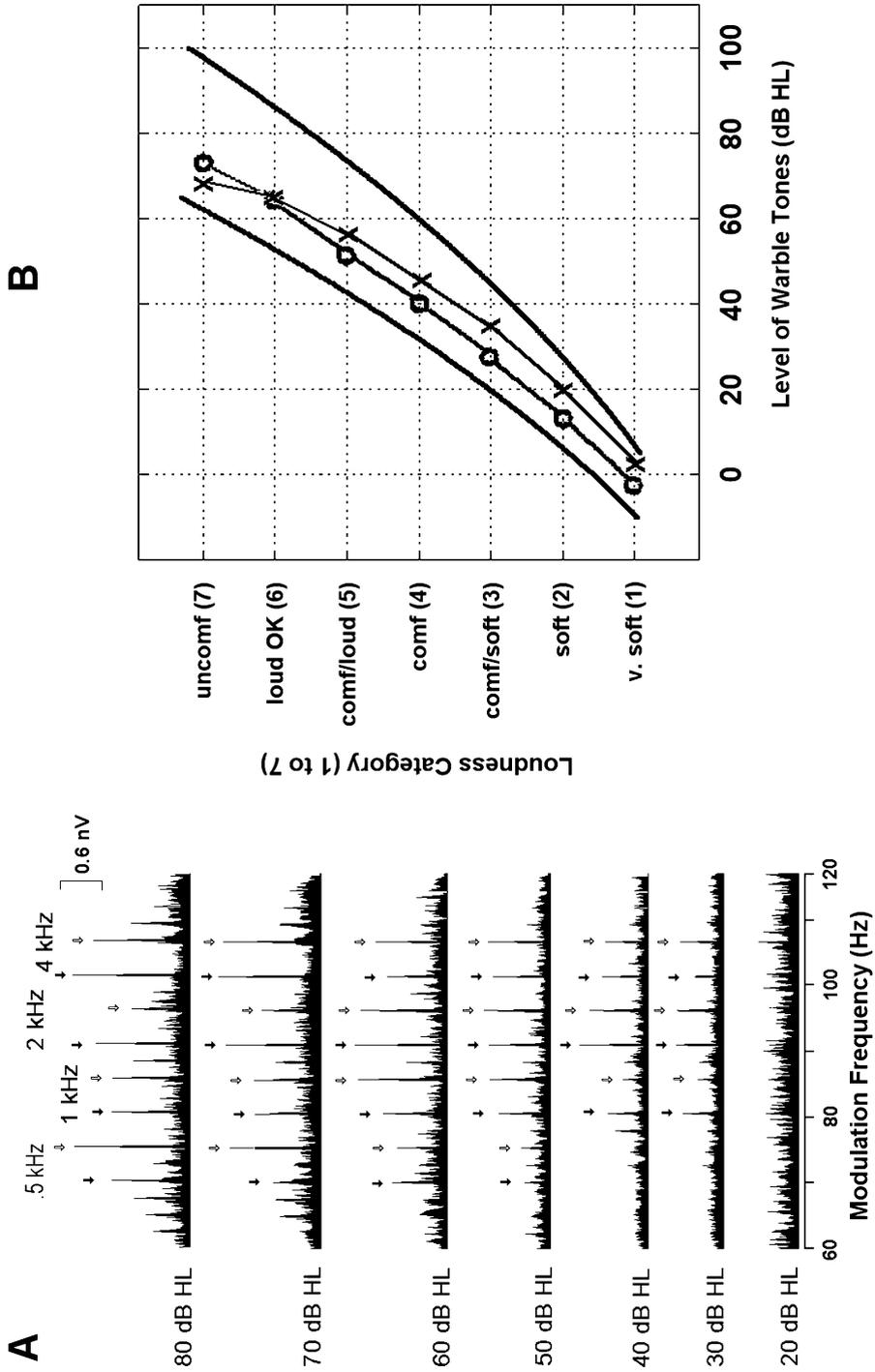


Figure 13-2. A, ASSR amplitude spectra recorded at levels ranging from 80 to 20 dB HL for the four carrier frequencies presented to each ear. Arrows indicate a significant response. As level increases, so does the amplitude of the ASSR. B, Loudness Contour Test results are for the same subject.

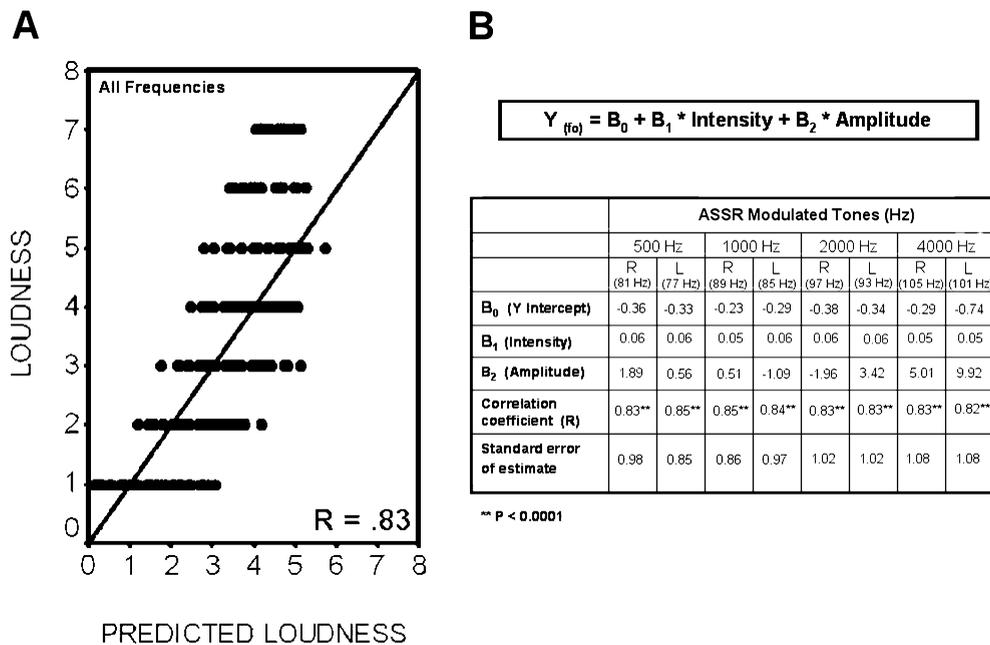


Figure 13-3. **A**, Loudness judgements plotted as a function of predicted loudness from the equation derived from the multiple regression analysis of ASSR amplitude with loudness contour. (The regression equation accounts for 70% of total variance.) **B**, The table shows the result of the multiple regression analysis predicting loudness, with a high correlation index regardless of carrier frequency and ear.

B₀, B₁, and B₂ are coefficient values obtained from the multiple regression analyses (Figure 13-3b). The sensation of loudness showed a significant correlation with ASSR amplitude growth of between 0.82 and 0.85 regardless of the carrier frequency. The regression line for all frequencies has a slope of -0.43 loudness/amplitude * level.

In summary, these data suggest that loudness growth can be reasonably well predicted from ASSR amplitude (at least in subjects with normal hearing).

Hearing Aid Selection

Regardless of the fitting formula employed to adjust a hearing aid, all procedures

must give information about some critical parameters. First, hearing dynamic range must be established from the pure tone hearing loss and loudness discomfort level; second, the hearing aid is supposed to amplify the whole range of speech into the dynamic range of a particular hearing loss; third, the difference between the hearing loss and the lower limit of the speech dynamic range provide the amount of gain required by the hearing aid; fourth, the compression factor will be determined by the degree of hearing loss to relative to the *long-term average speech spectrum* (LTASS) and finally, the *maximum power output* (MPO), the hearing aid power may be established from the amplitude growth function of the electrophysiological response.

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In a recent study, Zenker and colleagues (2006) proposed an ASSR prescription method based on the APP presented earlier in this chapter. Dynamic range, gain, compression, and MPO of the hearing aid were established from the level amplitude function of the ASSRs. ASSRs were obtained in a 27-month-old infant diagnosed with bilateral moderate sensorineural hearing impairment. ASSR thresholds were 50 dB SPL for 0.5 Hz and 1.0 kHz; 60 dB SPL for 2 kHz; and 70 dB SPL for 4 kHz. (Converted to dB HL, the thresholds were 55, 50, 63, and 75 dB for 0.5, 1.0, 2.0, and 4.0 kHz, respectively.)

Dynamic Range

Figure 13-4 shows the APP from which the dynamic range is obtained. In this figure, the previously established amplitude level function from a group of normal-hearing subjects (continuous line) is presented with the amplitude level function obtained from a group of hearing-impaired children (dashed line). In the same figure, the dynamic range of speech (40 to 80 dB) for the 500 Hz is projected upward from the abscissa to the normal amplitude intensity function. Then, horizontally, it is projected to the function obtained from the hearing-impaired individual and, finally, projected vertically upward to yield the output dynamic range. Hence, the equivalent dynamic range for this child with a moderate hearing loss is 27 dB (84 – 57 dB) for a 40-dB input range.

Gain and Compression Factor

The difference between the initial point of the output dynamic range (in this case, 57 dB) and the initial point of the

input dynamic range (40 dB) define the gain (57 – 40 = 17 dB) of the hearing aid. The width of the output dynamic range is determined from the output range (in this case, 57 to 84 dB, or 27 dB). The need for compression is determined by the ratio of the output dynamic range (27 dB) to the input dynamic range (40 dB); that is, $27/40 = 0.67$.

Maximum Power Output

From the ASSRs obtained for this patient, the MPO can be derived from the loudness sensation levels estimated from the amplitude level function. The MPO of the hearing aid is determined from the formula

$$\text{MPO} = [L_{\text{oud}} - B_0 - (B_2 * \text{amplitude})]/B_1$$

The values of the regression coefficients (determined from the multiple regression analysis described previously) are shown in Figure 13-3b. The MPO must be fitted to the category of uncomfortably loud, that is, contour test category 7. The amplitude data (Table 13-3) used in the formula are those obtained at the highest test level (80 dB SPL) for this patient. The MPO settings for this patient are 103, 110, 123, and 121 dB SPL for 0.5, 1.0, 2.0, and 4.0 kHz, respectively (see Table 13-3).

Finally, curves describing hearing aid output as a function of input are shown for each carrier frequency in Figure 13-5. These curves are shown with respect to threshold (estimated from ASSR threshold) and the maximum output (estimated from ASSR amplitude). As shown in the figure, using these fitting procedures, speech is amplified within the dynamic range of this patient's hearing.

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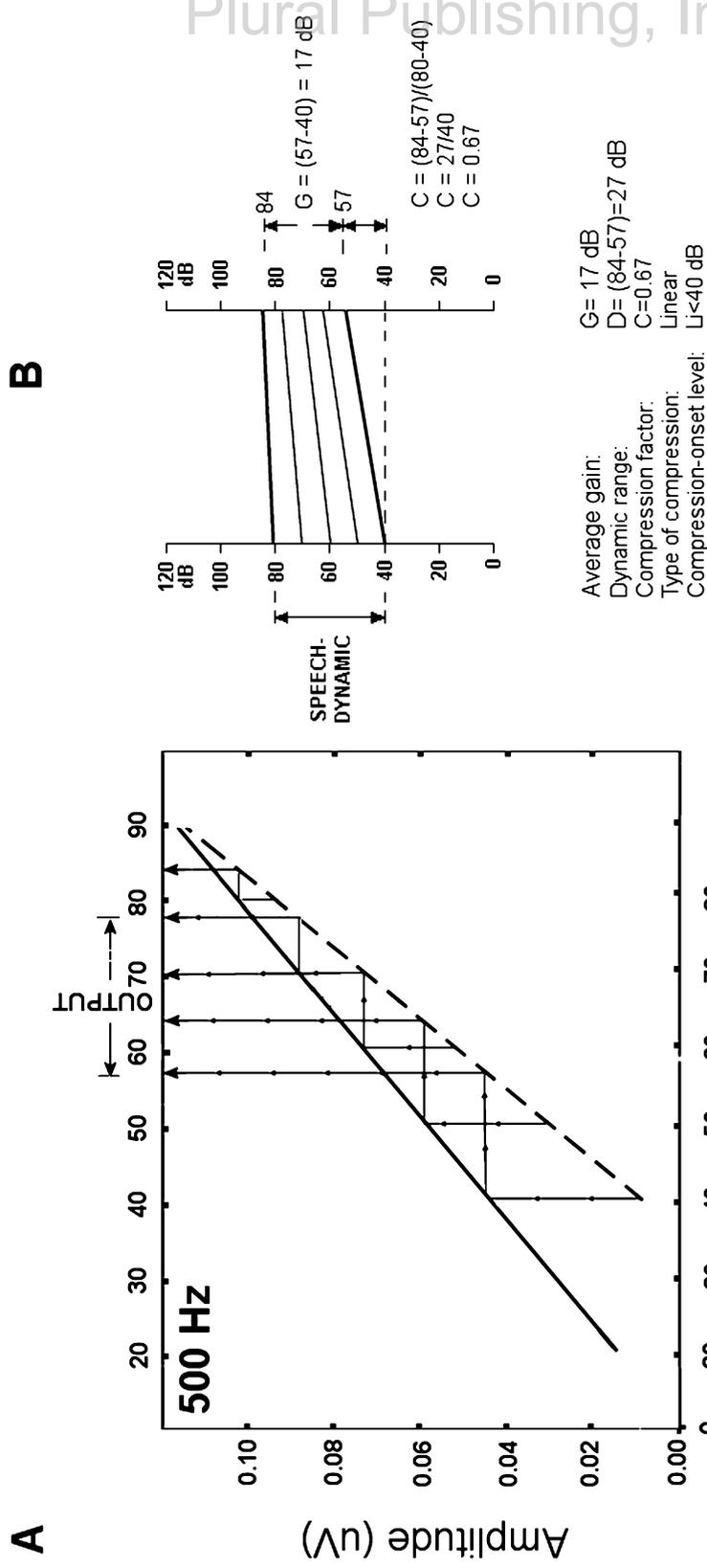


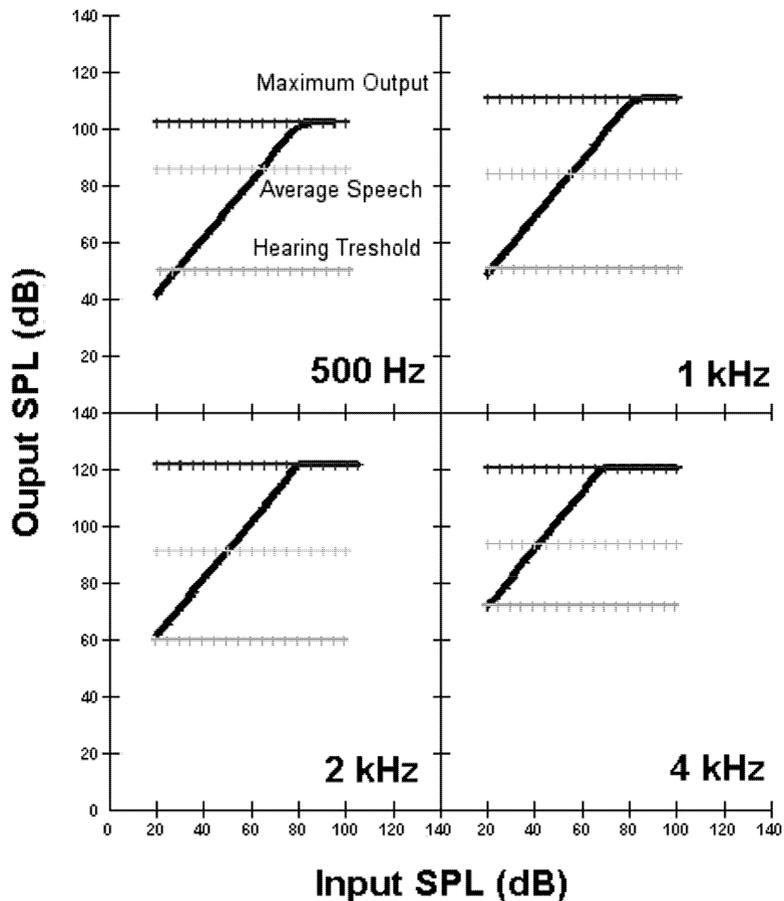
Figure 13-4. A, The amplitude projection procedure (APP). *Solid line* indicates the amplitude level function for normal-hearing subjects; *dashed line* indicates the amplitude level function for infants and children with sensory hearing loss. The input speech dynamic range of 40 to 80 dB is projected upward on the normal curve. Then the projection from the curve for ears with hearing loss yields the output dynamic range for this patient. **B**, The gain requirement is estimated as the difference between hearing loss (57 dB) and the lower limit of the LTASS (40 dB), or $57 - 40 = 17 \text{ dB}$. The compression factor (C) is given by the ratio of the dynamic range ($84 - 57 = 27 \text{ dB}$) of the patient to the normal speech dynamic range ($80 - 40 = 40 \text{ dB}$), so $C = 0.67$.

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Table 13–3. MPO values prescribed from the ASSR for a 27-month-old infant diagnosed with bilateral moderate sensorineural hearing impairment (see text)

Feature	ASSR Modulated Tones (Hz)			
	500 (81 Hz)	1000 (89 Hz)	2000 (97 Hz)	4000 (105 Hz)
Amplitude (μ V)	0.07	0.066	0.061	0.058
Loudness category	7	7	7	7
B0	-0.36	-0.26	-0.38	-0.29
B1	4.8	4	6.18	5
B2	1.89	0.51	-1.96	5.01
B2 \times amplitude	0.13	-0.03	-0.12	0.29
MPO	103	110	123	121

**Figure 13–5.** Input-output curves prescribed for each carrier frequency studied. The patient's hearing threshold, average speech for each carrier frequency, MPO, and final insertion gain calculated are shown.

Summary

The treatment of those with hearing loss involves the selection and fitting of amplification devices. In difficult-to-test individuals, such as young infants, subjective and objective measures such as functional gain and real-ear probe measurements, are not always possible. For those subjects who do not provide reliable responses to behavioural audiometry, the appropriate selection and fitting of hearing aids requires the establishment of accurate hearing thresholds by other means. ASSR can be used in the characterization of hearing loss to estimate the auditory threshold. In addition, the ASSR can provide information at threshold and also at suprathreshold levels. ASSRs can be used to verify and select the adjustment of hearing aids.

The role of ASSR in hearing aid fittings outlined in this chapter are: the estimation of hearing thresholds that can be introduced in the fitting software and the prescription of hearing aid features such as MPO, gain and the compression ratio. The procedures described in this chapter need further empirical demonstration of their effectiveness and accuracy. As well, the the estimation error of hearing threshold from the ASSR can be unacceptably large when applied to hearing aid fittings, but in cases where no other information is available, the ASSR thresholds provide a valuable starting point.

The APP method described provides estimates of loudness sensation derived from the amplitude level function of the ASSR. This procedure provides frequency specific information about features of hearing aids, such as average gain, compression factor, onset level, and

output limitation. This procedure, based directly on the ASSR amplitudes, could deliver an initial adjustment of the hearing aid until other behavioural responses can be obtained.

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Part B

Fitting Cochlear Implants Using Electrically Evoked Auditory Steady-State Responses

BARBARA CONE-WESSON

Infants with severe to profound congenital hearing loss are now undergoing cochlear implant surgery at the age of 12 months or younger. This sophisticated hearing technology requires a fair amount of “input data” about the user’s hearing sensitivity, comfortable and uncomfortable listening levels, and, ideally, speech perception abilities. The limitations of behavioural techniques in the infant population have spurred research aimed at determining auditory abilities from electrophysiological responses including transiently evoked (click and tone-burst) and steady-state evoked brainstem and cortical auditory evoked potentials (CAEPs). A large part of this research has used acoustic stimuli, although there is now an evidence base for using electrical stimulation (through the implant electrode) to evoke these responses for use

in estimating hearing abilities in persons with cochlear implants.

Vander Werff and associates (Chapter 7) discuss the strengths and weaknesses of using ASSRs to determine hearing sensitivity, the starting point for any hearing aid or cochlear implant prescriptive procedure. Dimitrijevic and Cone-Wesson (Chapter 12) describe ways in which ASSR may be used to infer suprathreshold hearing abilities, some of which may be relevant to determining the benefits of amplification or cochlear implantation. In the first segment of this chapter (“The Role of Auditory Steady-State Responses in Fitting Hearing Aids”), Zenker and Barajas propose a procedure for using (acoustically evoked) ASSR testing for fitting hearing aids. This method uses ASSR threshold, amplitude, and amplitude growth data to estimate the hearing aid

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gain, maximum power output (MPO), and compression as a function of frequency.

Would there be any benefit to using electrically evoked ASSRs for the purpose of fitting (“mapping”) a cochlear implant? Have there been any attempts to do so? The answer to the first question is best considered in light of the experience with other electrically evoked auditory potentials, such as eighth nerve action potentials (Abbas, Brown, Hughes, & Gantz, 1999; Brown, 2003; Hughes, Brown, Abbas, Wolaver, & Gervais, 2000a), ABR (Hughes, Brown, Abbas, Wolaver, & Gervais, 2000b; Kileny & Zwolan, 2004), middle latency and cortical evoked potentials (Kileny, 1991; Kileny, Zwolan, Boerst, & Telian, 1997; Sharma, Dorman, & Spahr, 2002). The answer to the second question is a qualified yes, and the one published attempt is described subsequently.

The need for electrophysiological methods for the purposes of verification of implant function and to determine electrical stimulation levels (i.e., threshold, comfortable and uncomfortable loudness settings) has been addressed by cochlear implant manufacturers. The manufacturers have developed the hardware and software for recording eighth nerve action potentials through the implant itself in conjunction with a computer interface used for the implant device programming. That is, the surgically implanted cochlear implant electrode to provide an electrical stimulus to the auditory nerve (spiral ganglion cells) and also as the recording electrode for the compound nerve action potential resulting from the electrical stimulus. The computer interface provides a method for display of the potentials and some rudimentary analyses. It is possible to determine the (electrical stimulus)

threshold of the auditory nerve action potential for a given electrode location. Such electrically-evoked auditory nerve-action potential (E-AP) thresholds can be used to guide initial stimulation settings. It is the case that the E-AP threshold does not approximate perceptual (electrical stimulation) threshold, but the presence of the E-AP indicates that the eighth nerve is stimuable at that electrode site. Furthermore, the E-AP threshold and the perceptual threshold have a fairly constant relationship across the electrode array, so determination of the amount of “offset” of the electrical from the perceptual threshold at one electrode site can be used to estimate the perceptual threshold across the array (i.e., set the “T” level). This is the major clinical application of the E-AP: to estimate perceptual threshold for the purpose of mapping. In research, the E-AP also can be used to investigate refractoriness/adaptation for electrical stimulation, and to estimate the spread of current in the cochlea which has implications for designing more (spectrally) precise stimulation schemes.

Given the fact that E-AP technologies are well developed, is there a need for electrically evoked ASSR (E-ASSR)? The ASSR for modulation rates of 70 Hz and above is primarily a brainstem response. But is there any advantage for E-ABR over E-AP? The experience with E-ABR (Brown, 2002; Hughes et al., 2000b) is similar to that with E-AP: The threshold of the E-ABR falls within the dynamic range of the listener but may well be closer to the comfortable listening levels than it is to perceptual threshold. Yet the presence of an E-ABR indicates that the stimulus provided by the implant is stimulating brainstem auditory nuclei and pathways. So the presence of an E-ASSR (for high

modulation rates), in conjunction with an E-AP, would give additional information with regard to brainstem auditory function but probably would not provide any additional information that could be used for device programming. Furthermore, unlike acoustically evoked ABRs or action potentials, in comparison with the ASSR, the issue of frequency specificity is not relevant, because the cochlear place of stimulation is determined by the choice of implant electrode stimulated.

Firzst and associates have completed seminal work on electrically evoked ABR (E-ABR), middle latency response (E-MLR), and E-CAEP in deaf adults who use implants to hear (Firzst, Chambers, & Kraus, 2002; Firzst, Chambers, Kraus, & Reeder, 2002). The work of these investigators shows that the latency and amplitude of the E-MLR had the highest correlations (among the E-AEPs) with speech perception abilities. This finding suggests that the E-ASSR for carriers modulated at 40 Hz, for which the generators are thought to be the same as those for MLR, may provide some information relevant to speech perception abilities. This is speculative at best.

The overriding technical issue in recording an evoked potential using electrical stimulation is that of electrical artifact created by the implant. Single or several cycles of biphasic electrical pulses are used to evoke transient potentials such as the E-AP or E-ABR. Even then, the stimulus artifact often swamps the electrophysiological response. To overcome artifact issues, the E-AP is derived from a forward masking paradigm in which the electrical artifacts of both the masker and the probe are subtracted from the average containing a response to the probe stimulus. In

the case of E-ABR, the stimulus artifact obscures early components of the response, and wave V may be the only component visualized. E-ABRs and later responses such as E-MLR and E-CAEP also require the use of radiofrequency filters and electrophysiological preamplifiers that are not overloaded by the electrical artifact created by the implant. For E-ASSR, the biphasic pulses must be presented in a steady-state (continuous) manner. This means that the stimulus artifact is present during the entire recording epoch and is orders of magnitude greater than the response.

Menard and colleagues (2004) tackled the technical problem of recording ASSRs for electrical stimulation provided by an implant. The Lyon group recorded ASSRs to amplitude-modulated steady-state biphasic pulses electrical pulses provided by an MXM Digisonic cochlear implant. The modulation rates were between 70 and 85 Hz. These investigators varied both pulse width (duration) and intensity to derive thresholds and response amplitude input-output functions. They reasoned that stimulus artifact would show a linear growth with intensity and duration, whereas physiological responses would show nonlinear growth functions. In this way, they estimated the portion of the response that was physiological versus the artifactual portion. This method took advantage of the charge density basis of neural excitability. Of note, this method of estimating the portion of response due to stimulus artifact and that portion due to neural excitation was used only for higher-intensity and longer-duration stimuli. The investigators made an assumption that the responses obtained for short pulse width durations were free of artifact. They compared the

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E-ASSR thresholds (in this case, the minimum pulse width duration that resulted in an ASSR) with perceptual thresholds obtained as a function of pulse width duration. A linear relationship was observed between perceptual threshold and ASSR threshold, and the differences between the two measures were on the same order of magnitude as those seen for E-ABR. This is not surprising: With the modulation rates used, the E-ASSR was generated primarily at the brainstem. The investigators further suggest that the E-ASSR input-output functions show saturation, and this saturation point may prove useful in the estimation of comfortable listening levels but is yet untested. (This concept is similar to that underlying the Zenker and Barajas method [see Chapter 13] of using ASSR amplitude to estimate MPO for hearing aids.)

The question remains: Is there an advantage to using E-ASSR over other evoked potential methods? One advantage may be that several electrode sites could be tested simultaneously by using different modulation rates for each. Given that programming cochlear implants requires the setting of threshold and comfortable listening levels for up to 22 electrodes, this could be advantageous in comparison with the single electrode test techniques (such as E-AP or E-ABR). This advantage may be outweighed by the technical demands of separating the steady-state stimulus artifact from the steady-state response.

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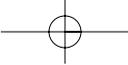
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