0196/0202/90/1101-0031$02.00/0
EAR AND HEARING

Vol. 11, No. 1
Printed in U.S.A.

AMPLIFICATION AND AURAL REHABILITATION

Evaluation of an In-Situ Output Probe-Microphone Method for Hearing Aid Fitting Verification*

Robyn M. Cox and Genevieve C. Alexander

Department of Audiology and Speech Pathology, Memphis State University [R. M. C.] and Veteran's Administration Medical Center [R. M. C., G. C. A.], Memphis, Tennessee

ABSTRACT

Several hearing aid prescription procedures specify frequency gain functions in terms of the desired levels of amplified speech. The most direct method for verifying hearing aid fittings based on these procedures requires a measurement of amplified speech in the ear canal. This paper describes the evaluation of a probe microphone measurement procedure designed to measure amplified speech to verify hearing aid fittings using the MSUv3 prescription procedure. With minor modifications, the same protocol could be used with other prescriptive procedures. Hearing aids were fitted to a group of subjects with sensorineural hearing impairment. Data derived from the fittings were analyzed to assess several issues relating to the procedure itself and to prescriptive fittings in general. The main results were: (1) with over the ear hearing aids, most prescriptions can be matched with an RMS error of 5 dB or less through the frequency range from 500 to 2500 Hz; (2) even though actual fittings usually do not perfectly correspond with their prescriptions, differences among frequency gain prescriptions are preserved in hearing aid fittings if the RMS error of the fitting is 5 dB or less; (3) if sound field stimuli presented to the hearing aid are precisely controlled, an in situ output verification method produces valid results; (4) when hearing aids are fitted with gain similar to that prescribed by the MSUv3 procedure, the maximum comfortable speech input level is typically about 72 dB; (5) the SSPL90 prescription generated by the MSUv3 procedure overestimates desired SSPL90 by 7 dB on average.

Verification of hearing aid fitting through real ear performance measurement is an important stage in many hearing aid prescription procedures. One approach to real ear verification uses probe microphone measurements in the ear canal. Measurement of insertion gain is the method of choice to verify hearing aid fittings for prescriptions that specify hearing aid frequency gain function in terms of desired real ear gain (Byrne & Dillon, 1986; McCandless & Lyregard, 1983). To date, most of the investigations and clinical work with probe microphone techniques have focused on the measurement of insertion gain and much useful information has resulted.

Some hearing aid prescription approaches specify frequency gain function in terms of the desired level of amplified speech rather than in desired real ear gain values (Cox, 1983; Seewald & Spiro, 1985; Skinner, Pascoe, Miller, & Popelka, 1982). Measurement of insertion gain does not lead to a straightforward verification of fittings based on these approaches. A more direct verification procedure involves a measurement of amplified speech in the ear canal: a measure of in situ output. There are several problems to be solved in developing an accurate in situ output measure of amplified speech in the ear canal. For example, the prescription rationale must be translated into appropriate ear canal target levels, the spectrum of the speech or speech-like signal used to drive the hearing aid must replicate the spectrum assumed by the prescription method, and the effects of an earmold vent on expected ear canal levels must be accommodated.

This paper reports the evaluation of a method for verifying a hearing aid fitting using ear canal measurement of amplified speech. The method was developed to implement the MSU (version 3) hearing aid prescription (Cox, 1988). However, with minor modifications, it could be adapted to other prescriptive approaches. In addition to describing the in situ procedure itself, we will present data derived from its application that have been used to assess the accuracy and validity of the measurements. Further, the adequacy of assumptions made in the MSUv3 procedure for SSPL90 prescription has been evaluated using in situ output measurements.

The Probe Microphone Measurement Procedure

Instrumentation The subject is seated in a double-walled, sound-treated room, 1 m from, and at a 0° azimuth

*This work was supported by VA Rehabilitation Research and Development funds.

Software to control administration of the probe microphone verification procedure was written by Robert M. Joyce.

to, a small, wall-mounted loudspeaker (Radioshack Minimus-7). Frequency response irregularities of the loudspeaker-transduced signal at the listener's position are equalized using a narrowband equalizer (White, model 4002). The loudspeaker is driven by stimuli from an audiometer (Fonix 3100). White noise and warble tone stimuli are produced by the audiometer. Speech noise and spondee words are digitized and routed through the audiometer. The audiometer is controlled by a microcomputer (IBM-AT clone) which also performs the necessary calculations.

The probe microphone (Etymotic ER-7) output is coupled directly to the input of a sound level meter with 1/3 octave band filters (Larson.Davis 800B). The probe tube is routed through a curved section of No. 13 hearing aid tubing, placed in the intertragal notch. The probe tube is extended into the ear canal to a minimum of 5 mm beyond the earmold tip. The assembly is taped to the face (using paper tape) to prevent movement of the tube.

Developing Ear Canal Frequency Gain Target Levels The MSUv3 prescription procedure computes desired gain at eight test frequencies: 250, 500, 800, 1000, 1600, 2500, 4000, and 6300 Hz. The long-term listening range at each frequency is defined as the range from threshold in SPL (SPHL) to the highest comfortable loudness level (HCL)†. Appropriate gain at each test frequency is determined according to one of two rules: (1) if the long-term listening range is $> =30$ dB, the 1/3 octave band spectrum of average speech at an overall level of 70 dB SPL‡ in the sound field should be amplified to the midpoint of this range, or (2) if the long-term listening range is <30 dB, the 1/3 octave band spectrum of average speech at 70 dB SPL should be amplified to a point 15 dB lower than the HCL.

The average speech spectrum assumed in the MSUv3 prescription is taken from Cox and Moore (1988). For a broadband level of 70 dB SPL the 1/3 octave band levels at the eight test frequencies are 60, 62, 56.5, 55, 52, 48, 46, and 45.5 dB, respectively. To produce an appropriate speech spectrum noise for this verification procedure, a broadband noise was filtered to correspond to the average spectrum and digitized with 7 kHz audio bandwidth. Figure 1 illustrates the long-term average spectrum of the speech noise.

The ear canal target levels for amplified speech could be obtained by measuring SPHL and HCL with the probe microphone in place, allowing direct measurement of the ear canal levels for these variables. The main drawback of this procedure is that at frequencies where hearing is close to normal, SPHL measurement may be compromised by the internal noise of the probe microphone (microphone 1/3 octave band internal noise levels are 21–25 dB SPL at, and below, 1000 Hz). Although there are several solutions to this problem, our approach is to measure the SPHLs and HCLs using supraaural earphones (these measurements may be made in advance of the hearing aid fitting session, if desired). The resulting levels are calibrated, as usual, in a standard 6 cm^3 coupler. These levels are later transformed to ear canal levels using values derived for each individual during the hearing aid fitting session.

The 6 cm^3 coupler-ear canal differences are determined for each subject and for the particular probe microphone location used. The procedure is as follows: (1) the supraaural earphone is placed on the 6 cm^3 coupler and 1/3 octave band levels are measured at each test frequency for white noise presented via the earphone at 70 to 80 dB SPL overall (this is done in advance and need not be repeated for each new subject); (2) after the probe microphone has been positioned in the ear canal, the same supraaural earphone is placed over the subject's ear, the same white noise is presented, and the 1/3 octave band levels at the test frequencies are measured in the ear canal; (3) the differences are derived by subtracting the coupler level from the ear canal level at each frequency (because the probe microphone has essentially the same frequency response and sensitivity as the 1 inch condensor microphone used in the 6 cm^3 coupler, measurements from the two microphones can be directly compared).

If the hearing-impaired individual has relatively good low-frequency hearing, the prescription calls for minimal (or negative) low-frequency gain. Also, the hearing aid is typically fitted with a large vent in the earmold or shell.

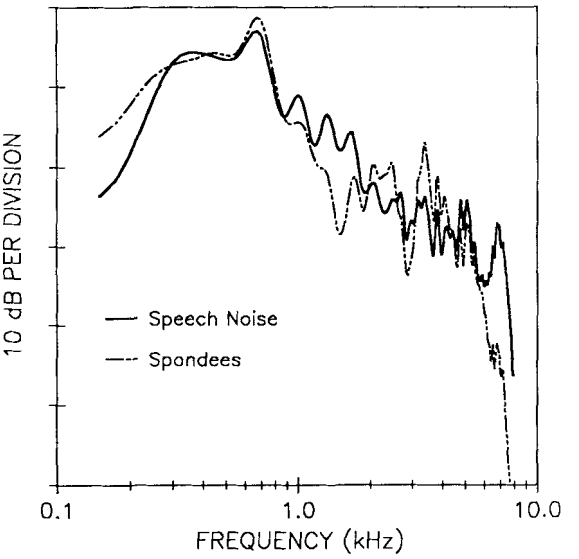

Figure 1. Long-term RMS spectrum of speech noise and spondee words used in the in situ output probe microphone verification procedure. These spectra were measured in the sound field at the subject's position with the subject absent.

†In previous work the highest comfortable loudness level has been referred to as the upper limit of the comfortable loudness range and abbreviated using the pneumonic ULCL. However, the term ULCL is often confused with UCL (uncomfortable loudess). To avoid this, the terminology has been changed. The psychophysical procedure for ULCL/HCL measurement is described in Cox, 1985, Appendix A.

‡Speech level is calibrated with reference to the level of the frequent peaks on a sound level meter (RMS, fast).

The desired result is that the low-frequency energy reaching the ear canal is predominantly natural, unamplified sound. To verify this type of fitting, it is necessary to modify the ear canal target levels to reflect the expected unamplified, low-frequency levels. To predict unamplified 1/3 octave speech-band levels in the ear canal, the free field to eardrum transfer function (Shaw, 1974) was added to the average speech spectrum values.§ The resultant levels (61, 64, 59, and 57 dB at 250, 500, 800, and 1000 Hz, respectively) are the lowest acceptable target levels at these frequencies. If any ear canal target level calculated using the prescriptive procedure is lower, it is replaced with the unamplified level.

In summary, SPHLs and HCLs are determined using the supraaural earphone. Second, the ear canal-coupler differences are measured. Third, SPHLs and HCLs are transformed to ear canal levels. Finally, target ear canal levels for amplified speech are established using the rules given above.

Verifying the Frequency Gain Function After the amplified speech target levels have been established, a preselected hearing aid, configured to approximate the prescription and coupled to the subject's earmold, is positioned in the ear canal without moving the probe tube. A speech-spectrum noise is presented via the loudspeaker at 70 dB SPL overall. Ear canal 1/3 octave band levels for each test frequency are measured (via the probe microphone and sound level meter) and compared to the target levels. Adjustments are made to maximize the correspondence between the target levels and the measured levels.

Developing Ear Canal Maximum Output (SSPL90) Target Levels The MSUv3 prescription procedure computes desired SSPL90 levels at each test frequency based on the premise that, if possible, speech quality should be preserved under all likely listening conditions. To achieve this, SSPL90 levels are prescribed so that only 1% of the instantaneous speech peaks will be limited when speech is amplified to the highest level acceptable to the hearing aid wearer. It is assumed that amplified speech can comfortably be increased in level until its overall level is equal to the narrowband HCL in the frequency region where the amplified speech spectrum and the HCL contour are closest together. Operationally, this is defined by the assumption that when speech is amplified to its maximum acceptable level, the closest approximation between the 1/3 octave band speech spectrum and the narrowband HCL contour is 5 dB. The rationale for these assumptions is presented in Appendix A.

The in situ output verification procedure discussed in this paper can be used to directly test the highest amplified speech level likely to be chosen by the hearing aid wearer (thus, obviating the need to assume that the closest 1/3 octave band level of amplified speech will be 5 dB lower than the individual's HCL contour). This is accomplished by: (1) measuring the sound field speech level that, when amplified, produces the HCL for aided speech, and (2) determining the ear canal 1/3 octave band levels for a speech spectrum sound equal to the spondee HCL, presented in the sound field and amplified by the hearing aid.

The procedure is performed after the frequency gain function has been adjusted as described above. Spondee words (Auditec recording) are presented in the sound field (Fig. 1) and the subject is instructed to respond whenever their loudness would be comfortable for long-term listening. The level of the spondee words is varied according to the psychophysical procedure for HCL measurement (Cox, 1985) until the level required to produce the HCL for aided speech is determined. Next, a speech spectrum sound equal to this level is presented in the sound field and the RMS 1/3 octave band levels in the ear canal are measured at the test frequencies with the probe microphone. These levels are the estimate of the highest amplified 1/3 octave band speech levels likely to be accepted by this listener and, thus, the highest levels for which significant speech distortion should be avoided. SSPL90 ear canal target levels can then be computed for each 1/3 octave band by adding the speech peak to long-term RMS levels (Dunn & White, 1940) to the ear canal highest amplified speech levels.

Verifying the SSPL90 Levels It is possible to test the hearing aid's SSPL90 levels by presenting narrowband, high level stimuli in the sound field, measuring amplified output levels in the ear canal, and comparing these with the target levels. Alternatively, adjustments may be made to the coupler SSPL90 prescription derived from the MSUv3 procedure. For example, if the maximum acceptable speech level is determined to be 10 dB lower than the nearest point on the HCL contour rather than the assumed 5 dB, the prescribed SSPL90 coupler values can be decreased by 5 dB. If desired, the coupler SSPL90 prescription can be further refined by adjustment for large errors in the amplified speech levels measured in the ear canal. For example, if the amplified speech level at 4000 Hz is 15 dB less than its target value, the coupler SSPL90 prescription can be reduced by 15 dB at 4000 Hz.

An Illustration Table 1 gives data obtained in an in situ output probe microphone verification session. No data were obtained at 6300 Hz. Each column is numbered to facilitate description of the Table. Columns 2 and 3 give the threshold and HCL data obtained using warble tones via TDH-39 earphones. Column 4 gives the ear canal-coupler differences for the probe tube location used in this subject: these values are used to transform the coupler data to ear canal levels. Column 5 gives the target amplified speech values: these are the levels to which speech at 70 dB SPL should be amplified in the ear canal. They are computed from data in columns 2, 3, and 4 using the rules given earlier. Note that the targets for the four lowest frequencies consist of two values with the second in parentheses. The first number (e.g., 80 at 250 Hz) is the target derived using the MSUv3 prescription rules. The value in parentheses (e.g., 61 at 250 Hz) is the

§Individual measurements of free field to eardrum transfer functions are not performed because both transfer function values and intersubject variability are small in the low frequencies.

Table 1. Data obtained from an in-situ output probe microphone verification session. 6 cm^3 SPHL = hearing threshold (dB SPL re 6 cm^3 coupler); 6 cm^3 HCL = highest comfortable loudness level (dB SPL re 6 cm^3 coupler); RE-6 cm^3 Diffs = ear canal-6 cm^3 coupler differences; NB = narrow band.

(1) Test Hz	(2) 6 cm^3 SPHL	(3) 6 cm^3 HCL	(4) RE-6 cm^3 Diffs	(5) Target Output	(6) Aided Output	(7) Unaided NB HCL	(8) Aided Spch HCL
250	75	115	−15	80 (61)	66	100	76
500	65	105	−6	79 (64)	82	99	91
800	66	111	0	88 (59)	88	111	97
1000	71	106	0	88 (57)	90	106	99
1600	79	116	−1	96	91	115	101
2500	79	109	1	95	94	110	104
4000	74	119	−2	94	81	117	92

expected value of unamplified speech in the ear canal. The higher of these two values is the appropriate target. If the subject's prescription calls for minimal, low-frequency gain and a vented earmold, the parenthetical value will probably be higher. In this illustration, the MSUv3 prescription value is higher and is, therefore, the appropriate target. Column 6 shows the actual levels to which the 70 dB speech noise was amplified in the ear canal. Comparison of columns 5 and 6 reveals the extent to which the frequency gain function of the fitted hearing aid matches the prescription. In this case, the RMS (root mean square) error in the frequency range from 500 to 2500 Hz is 2.8 dB.

Column 7 shows ear canal HCL levels for warble tones. These are derived by adding columns 3 and 4. Column 8 gives the highest acceptable $^1/_3$ octave band levels of amplified speech. Columns 7 and 8 are compared to determine whether adjustments are necessary to the SSPL90 prescription. In this case, the smallest difference between narrow band HCLs and the $^1/_3$ octave band levels of amplified broadband speech at HCL is 6 dB at 2500 Hz. This implies that, with this hearing aid, the minimum separation between the highest acceptable speech spectrum and the HCL contour is 6 dB rather than the assumed 5 dB. Thus, a 1 dB adjustment could be made in the SSPL90 prescription (probably too trivial a change to matter). In addition, because amplified speech at 4000 Hz is 13 dB less than its target value, the SSPL90 prescription could be reduced 13 dB at this frequency. A similar adjustment could be made at 250 Hz.

METHOD

This hearing aid fitting verification procedure was used to fit hearing aids for a group of hearing-impaired subjects. Data from the hearing aid fittings were analyzed to evaluate reliability, accuracy, and validity issues.

Subjects Data were collected on 19 to 24 hearing-impaired subjects (different numbers of subjects served for different aspects of the procedure). All were adults with sensorineural impairment bilaterally. A wide range of audiometric configurations was encompassed. Speech reception thresholds ranged from 10 dB HL to 67.5 dB HL (re: ANSI, 1969). Audiogram slopes (500–4000 Hz) ranged from −2.0 to 22.0 dB/octave.

Procedure Each subject was fitted with three hearing aids, each attempting to match a different frequency gain prescription. The three prescriptions were derived from: (1) the MSUv3 procedure; (2) the MSUv3 procedure plus 4 dB/octave, giving greater high-frequency emphasis; and (3) the MSUv3 procedure minus 4 dB/octave, giving greater low frequency emphasis. The three prescriptions for any particular individual probably covered the range of slopes likely to be called for by most prescriptive procedures.

All hearing aids used were conventional, analog, linear, nondirectional, over the ear devices. Each hearing aid was selected, adjusted, and measured using the procedure described in this paper. Earmolds, earhook dampers, and tone controls were used to maximize correspondence between target and fitted levels. Each hearing aid was configured for maximum SSPL90 so that the measurement of maximum acceptable amplified speech level would not be compromised by saturation effects.

RESULTS AND DISCUSSION

Ear Canal-Coupler Differences Figure 2 illustrates the mean and variability of ear canal-coupler differences measured for 24 subjects. Error bars give 1 SD. Overall, the differences are similar in shape and intersubject variability to the coupler to eardrum transformation given by Cox (1986). The standard deviations are about 2 to 4 dB, implying a range of individual differences of ±6 to ±12 dB, depending on frequency. These intersubject differences are sufficiently large to indicate that derivation of individual ear canal-coupler differences adds considerable precision to the verification procedure.

Measurement of the ear canal-coupler differences was repeated at the end of the testing session for eight subjects. The average test-retest difference (all frequencies included) was 0.95 dB.

Accuracy in Matching Frequency Gain Target Levels Several investigators have demonstrated that different prescriptive strategies often produce substantially different prescriptions for the same individual (Byrne, 1987). However, anyone experienced in probe microphone verification of hearing aid fittings is familiar with the fact that compromises often seem to be necessary when real ear performance is matched to desired performance determined from a prescription. Indeed, one may form the impression that the accuracy of prescriptive fittings with currently available hearing aids is rather poor. If this is the case, theoretical approaches that produce

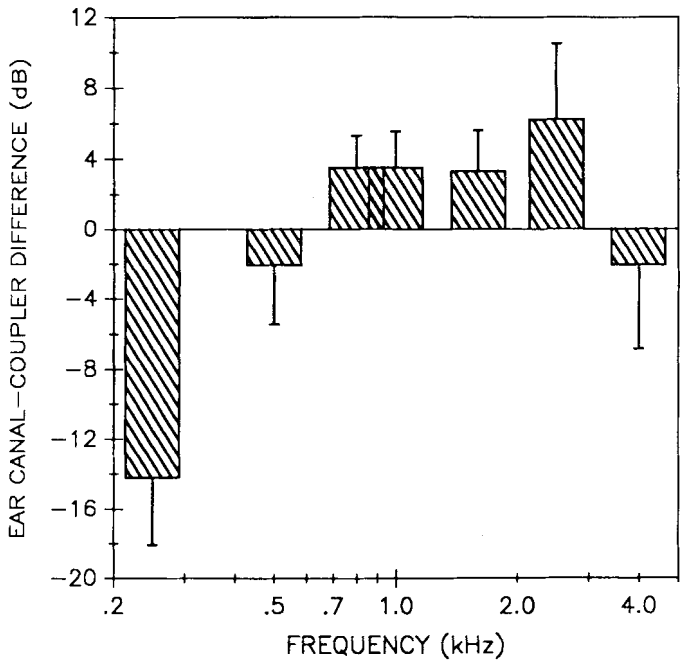

Figure 2. Typical difference between ear canal and 6 cm^3 coupler levels at test frequencies. Data are mean of 24 subjects. Error bars show 1 SD. Negative differences imply that coupler levels were higher than corresponding ear canal levels.

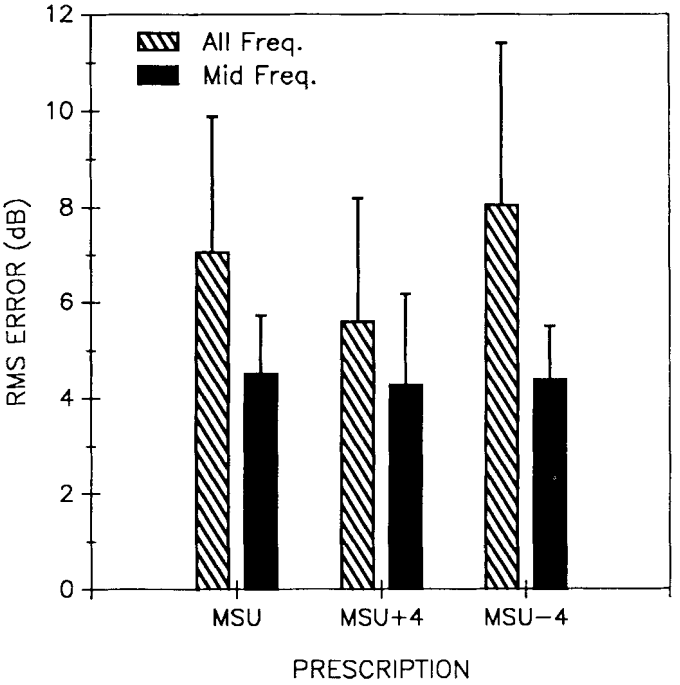

Figure 3. Mean RMS difference between real ear target output levels and corresponding measured aided levels. Data are shown for three prescriptions and for two test frequency ranges. All freq. = RMS differences computed across all test frequencies, 250–4000 Hz. Mid freq. = RMS differences computed across the middle test frequencies. $N = 20$.

different prescriptions may not actually result in different hearing aid fittings.

To evaluate this possibility, data from the hearing aid fittings were analyzed to address the following questions:

1. After a hearing aid has been adjusted to match a prescription, what is the typical RMS difference between the target and fitted frequency gain functions (real ear measurements)?

2. Are differences between prescriptions preserved through the fitting process in spite of the compromises necessary in matching real hearing aids to theoretical prescriptive requirements?

RMS differences were computed between real ear target levels and actual real ear levels for each subject and each prescription. In addition to computing the RMS difference across all tested frequencies from 250 through 4000 Hz, a midrange RMS difference was computed across the frequencies 500 through 2500 Hz. Figure 3 illustrates the means and variabilities of the resulting data. This figure reveals that, when the RMS difference was computed across the entire range of tested frequencies, the correspondence between target and fitted amplified speech levels was best (i.e., error was smallest) for the positive slope prescription and worst for the negative slope prescription. This difference was statistically significant ($p < 0.01$). However, when the RMS difference was computed across the middle range of frequencies, the correspondence between target and fitted amplified speech levels was about the same for all three prescriptions. In addition, the mean overall difference between target and fitted levels was 4.5 dB or less for the midrange frequencies but 5.5 to 8.0 dB for the full range of frequencies. Because the frequency response slopes fitted for each individual probably encompassed most of the plausible range of slopes, these data suggest that most prescriptions can be matched with fair precision throughout the midfrequency range.

How closely must real ear performance correspond to the real ear prescription if a hearing aid fitting is to exemplify the essential aspects of the prescriptive approach? No clear guidelines exist on this important clinical issue. Byrne and Dillon (1986) reported relevant data suggesting that an RMS difference of 3 dB or more between required and obtained frequency response was sufficient to produce a decrement in speech intelligibility or pleasantness. The implication is that fitted and prescribed performance should not differ by more than 3 dB (RMS). The results of the present study suggest that an accuracy criterion of 3 dB or less would not be a reasonable goal with conventional (nonprogrammable) hearing aids for many prescriptive fittings. However, a midrange RMS difference of 5 dB or less between target and fitted levels is an achievable goal for almost all routine fittings employing over the ear hearing aids. With currently available conventional hearing aids, fitting accuracy may be considerably poorer for the lowest and highest frequencies.

The data were further analyzed to determine whether the slope differences among the three prescriptions were preserved in the aided performance of the fitted hearing

aids. The fact that fitting accuracy was relatively poor for the lowest and highest frequencies, as shown above, suggested that slope differences among prescriptions may not be very well preserved in actual fittings. This issue was addressed by examining the relationship among the coupler frequency gain prescriptions and comparing this with the relationship among coupler frequency gain functions of fitted hearing aids. Figure 4 illustrates the mean frequency gain prescriptions (*upper panel*) and the mean frequency gain fittings (*lower panel*). Comparison of the two panels reveals that, although the overall shape of the frequency gain functions of the fitted hearing aids is somewhat different from the shape of the prescriptions, the relationship among the three lines is very similar in both panels.

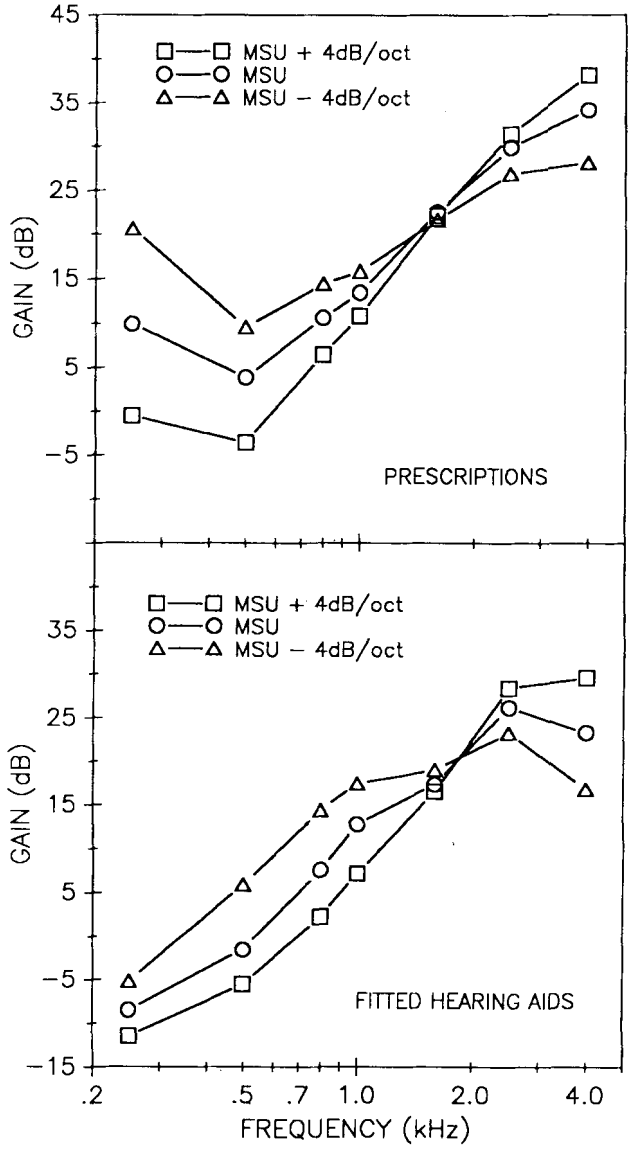

Figure 4. Mean 2 cm^3 coupler frequency gain prescriptions for the group of 20 hearing-impaired subjects (*upper panel*), and mean 2 cm^3 coupler frequency gain functions of hearing aids fitted to match those prescriptions (*lower panel*).

This outcome indicates that, overall, differences of 4 dB/octave among prescriptions are maintained in hearing aid fittings when individual fittings have a midrange RMS error (between target and fitted levels) of 5 dB or less, as in this study. However, closer examination of the hearing aid fitting data revealed that the extent to which prescription slope differences are preserved in fittings is somewhat dependent on audiometric configuration. This is illustrated in Figure 5, which summarizes data for eight subjects with relatively flat hearing losses in the *upper panel* (slope < 11 dB/octave) and for nine subjects with relatively sloping hearing losses in the *lower panel* (slope > 16 dB/octave). Clearly, the distinctions among the three prescriptions were less well maintained for the subjects with the more sloping hearing losses. Nevertheless, marked differences are observable between fittings for the three prescriptions even for this difficult to fit group.

Validity of Approach for Frequency Gain Verification

If the in situ output approach used here is valid (i.e., if all the relevant variables have been controlled), the coupler gain of fitted hearing aids, averaged across a group of subjects, should be equal to the mean coupler prescription for that group. To investigate validity, mean 2 cm^3 coupler frequency gain prescriptions were compared to mean 2 cm^3 coupler frequency gain functions of hearing aids fitted using the in situ output method. A separate comparison was made for each of the three prescriptions. Results were essentially the same for all prescriptions. Figure 6 illustrates the outcome for the MSUv3 prescription. This Figure has two noteworthy features. First, fitted gain at 250 and 4000 Hz was substantially less than prescribed gain for those frequencies.‖ This is consistent with the accuracy data described above and is attributed to the design of the hearing aids. That is, in order to get a close match to target levels in the midfrequencies, it was necessary to accept less gain at the two extreme frequencies.

The second noteworthy aspect of Figure 6 concerns the match between prescribed and fitted gain through the 500 to 2500 Hz range. Although real ear fitted levels corresponded rather closely to real ear target levels through this frequency range, the coupler gain of fitted hearing aids was consistently 2 to 5 dB less than the prescribed coupler gain. This discrepancy suggested that the in situ output measurement procedure was not producing a completely valid verification of the hearing aid fitting. In a valid procedure, the mean fitted coupler gain should accurately

‖Because the MSUv3 procedure prescribes more gain at 250 Hz than most other prescription methods, the discrepancy observed between prescribed and fitted gain at 250 Hz might not be observed if a different prescriptive approach were used. However, the result at 4000 Hz would probably be found with almost all prescriptive procedures because adequate gain at this frequency is difficult to achieve with most conventional hearing aids.

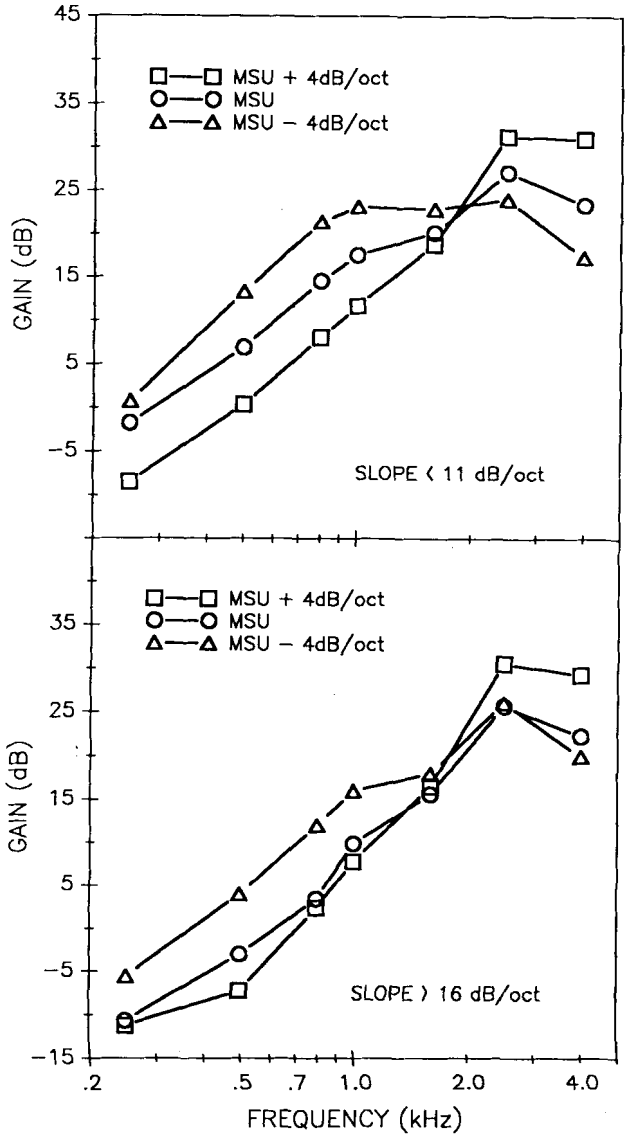

Figure 5. Mean 2 cm^3 coupler frequency gain functions of hearing aids fitted to match three prescriptions for eight subjects with relatively flat audiometric configurations (*upper panel*), and for nine subjects with more sloping audiometric configurations (*lower panel*).

match the mean prescribed coupler gain when real ear output levels closely approximate real ear target levels.¶

Further investigation revealed a problem with the speech spectrum noise used to simulate average speech.

¶It should be noted that such a match would not necessarily be expected in single subject data because of individual differences in real ear acoustic effects. However, across a group of subjects, real ear acoustic effects should combine to produce an average result similar to the assumptions made in the prescription procedure. Thus, prescribed and fitted frequency gain functions should match when data are combined across a relatively large group of subjects.

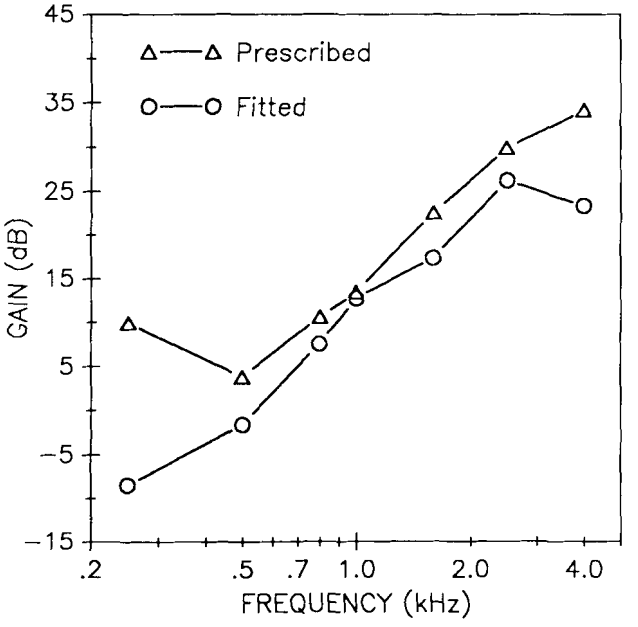

Figure 6. Mean 2 cm^3 coupler frequency gain prescription derived from the MSUv3 procedure (*triangles*), and mean 2 cm^3 coupler frequency gain function of corresponding fitted hearing aids (*circles*). $N = 20$.

This noise was produced in the sound field at 70 dB SPL overall and was amplified by the hearing aid to generate the ear canal levels that were compared to the target levels. Measurements of actual $^1/_3$ octave band levels of speech noise at the subject's position in the sound field indicated that the combined effects of room acoustics and low-frequency roll-off had resulted in delivery of the $^1/_3$ octave band test frequencies at levels 1 to 4 dB higher than the levels assumed in the prescription procedure. As a result, the gain required at these frequencies to produce the target ear canal levels was less than predicted by the prescription. This problem was readily corrected by varying the overall level of the speech noise until the differences between $^1/_3$ octave band levels assumed and observed at the subject's position were minimized (even though this did not correspond with an overall speech noise level of 70 dB SPL).

This observation of unexpectedly high speech-band levels at the tested frequencies underscores a substantial limitation of the in situ output verification procedure: because ear canal output is a combination of sound field signal and hearing aid gain, it is essential that good control is maintained over both variables. Systematic deviations of the simulated speech signal from the speech signal assumed in the prescriptive procedure will result in systematic inaccuracies in hearing aid fittings verified using the in situ output approach.

Measurement of Highest Acceptable Amplified Speech Level One of the noteworthy advantages of an in situ output approach to hearing aid fitting verification is the capability of observing hearing aid performance under conditions close to those of everyday use. This advantage

is clearly illustrated when SSPL90 requirements are determined according to the MSUv3 procedure. In the conventional application of this procedure (without in situ output measurements), it is necessary to make an assumption about the highest acceptable level of amplified speech for the individual listener. However, with in situ output measurements, this level can be determined directly by measuring the 1/3 octave band contour of speech that is amplified to the HCL. The results of these measurements, in addition to providing direct determination of SSPL90 requirements, provided data that are useful in evaluating the assumptions about HCL for amplified speech embodied in the conventional MSUv3 procedure.

After each hearing aid was adjusted to produce an output corresponding to the target levels, the highest comfortable loudness level was measured for amplified spondee words. Two measures were obtained: (1) the sound field level of the spondee words at HCL and (2) the 1/3 octave band ear canal levels of speech noise presented at the HCL for spondee words and amplified by the hearing aid. These data were used: (1) to determine the highest acceptable speech input level for hearing aid fittings using gain similar to that prescribed in the MSUv3 procedure, and (2) to assess the relationship of amplified broadband speech at HCL to the the HCL contour previously established for narrow band stimuli.

Results indicated that the mean spondee word HCL was 73.5 to 75.5 dB SPL (SD = 6.5–7.5 dB) in the sound field at the subject's position (differing slightly but not significantly across the three prescriptions). These values should be reduced by 2 dB to account for the inaccuracy, described above, of the 1/3 octave band speech noise levels used in the frequency gain verification stage. The implication of this observation is that, when the gain of a linear hearing aid is adjusted according to the MSUv3 prescription, overall speech input levels up to about 72 dB are comfortable for the typical listener. Speech inputs above this level are not comfortably loud.

This outcome is interesting in light of the relatively low gain prescribed by the MSUv3 procedure compared to many other prescriptive methods (Skinner, 1988, p 186). These data suggest that the result of using higher gain would be an even lower limit on comfortable speech input levels.

The MSUv3 procedure assumes that amplified speech at HCL (measured in 1/3 octave bands) falls 5 dB below the closest point of the narrowband HCL contour. To evaluate this assumption, the contour of mean 1/3 octave band levels for amplified speech at HCL was determined for each prescription. These three contours and the mean narrow band contour are shown in Figure 7. All data are plotted in ear canal sound pressure levels. When the amplified speech spectra are compared to the narrowband HCL contour, it is evident that the assumption made for SSPL90 prescription in the MSUv3 procedure was not verified: none of the mean amplified spectra approaches within 5 dB of the mean HCL contour. Instead, closest approximation of the amplified speech spectrum to the

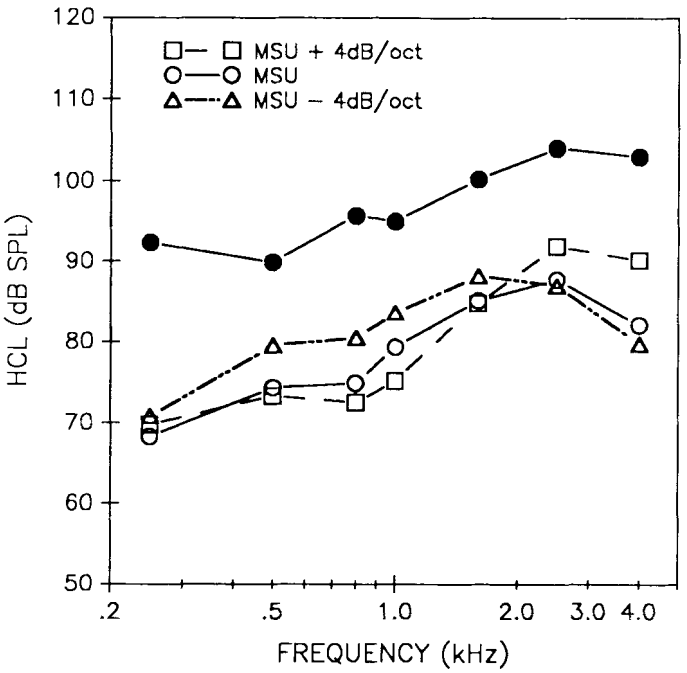

Figure 7. Mean unaided narrow band HCL contour (*filled circles*); mean 1/3 octave band spectra of amplified speech for three prescriptions when amplified speech was set to the maximum acceptable level (*open symbols*). All data are in ear canal sound pressure levels. *N*= 19.

narrowband HCLs averages about 12 dB across prescriptions.#

The implication of this outcome is that, in the typical case, the SSPL90 prescriptions generated by the MSUv3 procedure overestimate desired maximum output by 7 dB (the difference between observed and assumed values). It also should be noted that there was considerable variability across individuals in the minimum separation between the HCL contour and the maximum acceptable amplified speech level (SD about 10 dB). Thus, individual testing of this factor seems advisable for each hearing aid fitting.

As noted earlier, MSUv3 SSPL90 levels are also derived on the assumption that amplified speech can comfortably be increased in level until its overall level is equal to the narrowband HCL in the frequency region where the amplified speech spectrum and the HCL contour are closest together. The data were explored to determine whether this assumption was confirmed. The average overall level of amplified speech in the ear canal was estimated for each frequency response slope using the mean 1/3 octave band levels. Again, the outcome was not as assumed. For the MSU-4 slope, the mean overall level was estimated at 95 dB whereas the mean level of the closest narrowband HCL was 90 dB (500 Hz). For the MSU slope, the mean overall level was estimated at 93 dB and the mean level of the

#It is possible that the amplified speech spectrum approached more closely to the HCL contour in some frequency region(s) that were not measured. However, because hearing aids seldom have narrow, high peaks when measured in real ears, this is not likely to be a major factor.

closest narrowband HCL was 100 dB (1600 Hz). Finally, the mean overall level for the MSU + 4 slope was estimated at 97 dB and the mean level of the closest narrowband HCL was 104 dB (2500 Hz).

Overall, the assumptions made in the MSU procedure about the maximum acceptable level of amplified speech were not confirmed when this level was actually measured in hearing aid wearers. Further investigations to explore the determinants of maximum acceptable amplified speech level are needed. In the meantime, it seems prudent to avoid assumptions about this issue, if possible. Thus, the use of in situ output measurements seems especially beneficial for determination of SSPL90 requirements.

FINAL COMMENT

This evaluation of an in situ output procedure for probe microphone verification of hearing aid fitting revealed both positive and negative aspects of this approach. The major advantage of in situ output measurements is their ability to portray the performance of the hearing aid fitting under conditions close to those of actual use. Certain assumptions necessary with an insertion gain approach do not have to be made with an in situ output verification method. This offers the promise of more accurate hearing aid fittings. The major disadvantage seems to be the requirement for rather precise control of the simulated speech stimulus and any other stimuli used as an input to the hearing aid.

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Address reprint requests to Robyn M. Cox, Ph.D., Memphis Speech & Hearing Center, 807 Jefferson Ave., Memphis, TN 38105.

Received May 19, 1989; accepted September 15, 1989.

APPENDIX A Rationale for MSUv3 procedure for SSPL90 prescription

The relationship between the HCL for amplified speech and the contour of HCLs for narrowband stimuli is not obvious and has not been empirically established. There is considerable evidence that loudness comfort for speech depends primarily on overall sound pressure level and is essentially independent of the bandwidth of the stimulus (Christen & Byrne, 1980; Hawkins, 1980; Ventry, 1976). However, there is not general agreement on this point because the role of loudness summation in speech perception is not clear, particularly for hearing-impaired listeners.

It is known that loudness summation contributes to the loudness of continuous sounds with a bandwidth wider than a critical band (Zwicker, Flottorp, & Stevens, 1957). If loudness summation occurs for speech, a speech stimulus would reach a specific comfort level (e.g., the HCL) at a lower overall level than a narrowband stimulus. Dirks and Kamm (1976) have produced evidence supporting this hypothesis for normal hearers. On the other hand, other studies have reported little or no loudness summation (as a function of bandwidth) in sensorineurally hearing-impaired listeners (Scharf & Florentine, 1982). This is consistent with the finding of widened critical bands in sensorineural impairment that has been reported by several investigators.

Taken together, the results of these investigations suggest that it is reasonable to assume that, in hearing-impaired listeners, loudness comfort level for speech is largely determined by overall sound pressure level (i.e., loudness summation is a minor factor).

Assuming: (1) that loudness comfort depends mainly on overall sound pressure level and (2) that the narrowband HCL contour is at a constant level across frequencies, it follows that the overall level of speech at HCL will be equal to the narrowband HCL. Furthermore, the closest approximation between the $^1/_3$ octave band speech spectrum and the narrowband HCL contour will be 8 dB at 500 Hz.** If the bandwidth of speech is restricted by a hearing aid, the difference between the narrowband HCL contour and the level of the 500 Hz $^1/_3$ octave band of speech could be less than 8 dB—perhaps 5 dB.

The application of this logic to hearing aid fittings is complicated by the fact that the shape of the amplified speech spectrum is usually different from the shape of the unamplified spectrum (e.g., high frequencies may be emphasized). In addition, the narrowband HCL contour for hearing-impaired listeners is often not flat as a function of frequency. The combined effects of these factors is difficult to predict.

Based on these considerations, the MSUv3 procedure assumes that when speech is amplified to its HCL, the closest approximation of the $^1/_3$ octave band speech spectrum to the narrowband HCL contour will be 5 dB.

**In typical speech the 500 Hz $^1/_3$ octave band has the highest long-term level and is 8 dB below the overall level (Cox & Moore, 1988).