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Speech discrimination with an 8-channel compression hearing aid and conventional aids in background of speech-band noise*

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Abstract—The design for a multichannel compression hearing aid was developed from previous experimental and theoretical work in our laboratory concerning pitch perception in normal-hearing subjects. The new hearing aid, implemented with off-line digital signal processing, was tested on twenty subjects with sensorineural hearing loss using speech sounds in a background of speech-spectrum noise. Five signal-to-noise ratios (+15 to −5dB) were used at two noise levels (60 and 70 dB SPL). Hearing-loss subjects listened to these stimuli under three different conditions: a) processed by the new multichannel compression hearing aid; b) processed by a conventional hearing aid; and c) unprocessed. The performance of normal-hearing subjects with the unprocessed stimuli provided another condition against which the performance in the two hearing aid conditions could be evaluated. Both aided conditions provided improved performance over the unprocessed condition and the multichannel compression aid produced better performance than the conventional aid. In the case of 4 of the 20 subjects, with less severe gradually sloping hearing losses, the new multichannel compression aid produced near-normal performance even at low signal-to-noise ratios. Some aspects of the results also suggested that learning to use the aid was more important in the case of the multichannel compression aid than in the case of the conventional aid. These results indicate that a multichannel compression hearing aid can be very effective in some individuals with sensorineural hearing loss and is superior to a conventional hearing aid in most subjects.

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INTRODUCTION

A multichannel compression hearing aid is one in which the amplification in each frequency band, or channel, decreases as the amplitude of the signal in that channel increases. The dependency of amplification upon amplitude also is usually different for the various channels. Using such an aid, an individual with sensorineural hearing loss (SNHL) could exhibit both normal thresholds and normal discomfort levels for narrow band stimuli and, in that very limited sense, may be said to exhibit "normal hearing." It is this obviously oversimplified view of multichannel compression that originally may have made it seem so promising and perhaps that still makes us continue to try to demonstrate its presumed superiority over simple linear amplification.

In one sense, however, such a view of multichannel compression cannot be dismissed as completely naive. Consider the pattern of neural signals, for some particular stimulus (either simple or complex to any degree), traveling up the auditory nerve in a normal-hearing subject. If it were possible to modify this stimulus in such a way that the modified stimulus produced the normal pattern of neural signals in the auditory nerve of a patient with cochlear hearing loss, then the patient should exhibit "normal hearing" for that stimulus, however complex it might be. A device that could "normalize" the auditory nerve signals for all possible stimuli for a particular SNHL patient, would be the "perfect" hearing aid for that patient. Although it may never be possible to make this "perfect" hearing aid, our current

understanding of SNHL indicates that mulitchannel compression would be an important component process in such a device. In SNHL, low intensity stimuli generate reduced responses or no response while high intensity stimuli generate essentially normal-magnitude responses and the threshold as well as the rate of response growth may vary with frequency. This phenomenon, called recruitment in the context of loudness perception, seems to be exactly the problem for which multichannel compression is the solution.

Demonstrating the efficacy of multichannel compression has proven much more difficult than recognizing its potential. In their review, Braida et al. (3) conclude that previous studies have not unambiguously shown an advantage for mulitchannel compression over a well-fit linear aid. Braida et al. attribute the lack of positive results primarily to the technical and methodological complexity of the problem rather than to an essential flaw in the idea of multichannel compression. Among the specific difficulties discussed by Braida et al. are: a) lack of subject training with compressed signals, b) use of speech stimuli that do not cover a wide enough range of intensities, c) lack of testing in noise, especially at lower signal-to-noise ratios (S/N), d) lack of a suitably chosen linear aid control condition against which the multichannel aid was compared, e) improper subject selection, f) distortion produced in the compression system that degrades speech recognition, and g) failure of the compression system to compress the speech signals into the intended intensity range.

Villchur (21) agrees that the advantage of multichannel compression has not been thoroughly demonstrated, although his emphasis differs somewhat from that of Braida et al. On one point, the need for a larger intensity range in the speech test material, Villchur and Braida et al. seem to be in complete agreement. Villchur stresses the need for a well-fit linear control condition, but also a realistic one that would be acceptable to the user in the range of conditions encountered outside the laboratory. In terms of subject selection, Villchur (again in agreement with Braida et al.) suggests concentrating efforts upon those subjects with severely reduced residual dynamic ranges.

Three studies published more recently have included at least some multichannel compression: Laurence, Moore, and Glasberg (11); Nábělek (16); and Walker, Byrne, and Dillon (22). Walker, Byrne, and Dillon, using a mixed (compression/expansion) system, reported no general advantage of that system over a linear control, although they did note improvement in one subject for stimuli loud enough to be predominantly in the compression phase of the device. Nábělek reported an advantage for a single channel (wideband) compression system but much less, if any, for multichannel compression. Laurence, Moore, and Glasberg reported better performance for their two-channel compression aid than for their linear system, both in quiet and in noise. However, their compression aid actually contains two separate compression stages, a single channel stage (not unlike Nábělek's wide band system) followed by a two-channel stage. Since Laurence, Moore, and Glasberg did not test either compression stage separately, it remains to be determined whether the success of their aid resulted from the action of the initial single channel compression or the later two-channel stage. Thus, including these recent studies, there remains no convincing evidence for the advantage of multichannel compression in hearing aid design.

The purpose of the current study was to test a particular multichannel compression hearing aid developed from previous experimental and theoretical work in our laboratory in normal-hearing subjects. The critical parts of the previous work as it relates to our mulitchannel compression system are a) the theoretical model formulated to explain certain dichotic pitch effects (25) and b) the development of a method to measure one parameter of the model—the intensity-response (I-R) transduction function of the cochlea at any frequency (6). From one perspective, the current study tests how effectively the model can represent an individual subject's SNHL, when its I-R parameters are measured for that subject. The individualized "SNHL" model is used to determine the type of signal processing (i.e., the hearing aid) that must be applied to its input in order to make its output, as much as possible, like the output of the model with a "normal-hearing" I-R mechanism. (For further details on the model see Yund and Efron (25), and on the I-R measurements see Efron, Yund, and Divenyi (6). Some typical I-R typical functions for one SNHL subject and how to derive amplification parameters from these are illustrated in the Method Section below.)

METHOD

Subjects

Twenty-nine volunteers were recruited to participate in this study from local speech and hearing centers and from the community. Seventeen men and three women (S2, S7, S11) ages 45–77 years (mean 64 years), with varying degrees of cochlear pathology, completed the study. All volunteers were selected on the basis of peripheral hearing function using pure-tone air conduction, speech-reception threshold testing, tympanometry, reflexometry, and performance-intensity functions for phonetically balanced words (9) using recorded CID W-22 word lists (8). Several audiometric examinations were administered to each over the course of the research; where minor threshold changes occurred, only the most recent values will be presented.

Audiometric test results for all subjects were consistent with cochlear pathology. Etiologies were unknown for some of the subjects; however, other subjects had familial histories of hearing loss with aging and 16 reported exposure to excessive noise levels. Eleven of the subjects were hearing aid users at the time of these experiments (S4, S6, S10, S11, S12, S13, S15, S16, S18, S19, S20). Of these, seven owned and wore the hearing aid that was used as the control (conventional) aid (S6, S11, S12, S13, S15, S16, S18). In one of these, S15, the subject began using the aid in the month before his speech testing started: he decided to purchase the recommended aid and began using it immediately.

All subjects were native speakers of English and none had a history of middle ear pathology, neurological problems, or alcohol/drug abuse. All subjects were paid for their participation.

Test Materials

The test materials consisted of the closed-set Nonsense Syllable Test (NST) developed by Resnick et al. (17). In the original version of this test, the subject was required to select the consonant heard from among a limited number of response foils in each of eleven subsets. These subsets differed in consonant voicing (voiced or unvoiced), position of consonants (initial or final), and vowel context (/a/, /i/ or /u/). The syllables had been originally recorded by a male speaker within the carrier phrase "You will mark ____ please." However, for our experiments, the syllables were presented in isolation rather than in this carrier phrase.

The NST was chosen because of its high reliability when administered to hearing-impaired listeners with mild-to-moderate losses (5), its sensitivity to changes in performance with alteration of hearing aid parameters and its ability to provide information about the consonant error patterns made by the listener (13). Walker, Byrne, and Dillon, (23) demonstrated learning effects with their version of the NST (using a female Australian speaker). Such learning effects should not confound the results in the critical comparisons in this study because of the multiple, randomized presentations of the test conditions and the practice sessions that were not included in the data analysis. In the methods section of a later paper, Walker, Byrne, and Dillon (22) present an excellent analysis of the reasons for using the NST in experiments like ours and we need not repeat that analysis here.

The stimuli were digitized at a rate of 20 kHz with 12-bit precision from the prerecorded tape and all further processing was done by digital means. After any of the remaining carrier phrase was trimmed away, the stimuli were filtered into eight bands using the ILS (Interactive Laboratory Systems, Signal Technology, Inc.) signal processing package. We used six half-octave bands from 750 to 4000 Hz in addition to one above 4000 Hz and one below 750 Hz. With the exception of the low-pass band, the digital Butterworth filters had slopes outside their bandpass of 50 to 70 dB per octave; the slope of the low-pass band was 20 to 40 dB per octave. The bands crossed at −3 dB at 595, 841, 1189, 1682, 2378, 3364, and 4757 Hz. For the unprocessed condition, these bands were recombined after filtering while for the processed conditions the bands were selectively amplified before recombination. Amplification within bands could be a simple linear constant—for a conventional aid—or a function of the signal level determined individually for the subject—for our model aid. Peak clipping or compression (based on input or output levels) was done after recombination to complete the conventional aid. Twelve of the conventional aids were linear with peak clipping; two (S9, S16) had output compression; six (S4, S7, S8, S10, S19, S20) had input compression. Speech spectrum noise (SSN), shaped to the long-term average speech spectrum (7) was added to the syllables prior to the processing that corresponded to the aids. The duration of the noise was 1 second.

Procedure

Speech stimuli were generated from the computer through a 16-bit digital-to-analog converter system, low-pass filtered at 8 kHz (Frequency Devices model 901F), attenuated (Hewlett-Packard model 350D), and delivered to the subject either through a Danavox SM-W wide band receiver coupled to the subject's earmold or, in five cases (S1, S3, S4, S11, S17), through a TDH-39 earphone and circumaural cushion. Earphones were routinely calibrated with a Bruel and Kjar sound level meter (Model 2203) with an octave filter set (Model 1613). The receivers were calibrated to the earphones using a loudness balance procedure with normal hearing listeners. All test items were presently monaurally to the ear judged to be the best ear for amplification.

Testing was done in a single-walled IAC booth in 2-hour sessions with rest periods between each run of the 91 syllables. Within each run, the order of presentation of the 91 syllables of the NST was independently randomized. Each trial of a run consisted of the following parts: a) A numbered list of the syllables in one subset of the NST was displayed on a computer terminal in front of the subject; b) One of the stimuli from that subset was presented; and c) The subject entered the number of the syllable (s)he had heard. This response entry was the computer's signal to begin the next trial. Subjects were told that a response was required to go on to the next trial and that they should just guess if they had no idea what sound they had heard. They also were told that they should use the sound they heard to reduce the size of the set from which they had to guess, whenever possible. Prior to testing with stimuli in noise, each of the first 10 subjects (S1 to S10) ran at least 20 2-hour sessions without noise, including unprocessed, conventional aid and model aid stimuli. These quiet sessions served primarily as training since all aids tested worked well in quiet. For this reason, we reduced the number of 2-hour sessions without noise to at least 10 for S11 to S20.

There were three experimental aid conditions at each S/N: a) "unprocessed," b) "conventional" aid, and c) "model" aid. Two noise levels (60 and 70 dB SPL) each including the five S/N values (+15, +10, +5, 0, and −5 dB) means that speech signals were presented at intensity levels corresponding to soft (55 dB SPL) and to loud speech (85 dB SPL) and were centered around the typical input level for a hearing aid wearer of 70 dB SPL according to Walden, Schuchman, and Sedge (24). Eight of the subjects (S1, S2, S4, S5, S6, S7, S8, S11) ran the 60 dB noise condition before the 70 dB noise condition. Within noise conditions, the five S/N were always run in descending order, beginning with +15 and ending at −5 or earlier if the subject was unable to perform above chance. Within each S/N condition, runs for each aid condition were done in random order; in this way learning effects, if any, were distributed equally across aid conditions.

As indicated above, all stimulus processing was done off-line in a digital computer. In the unprocessed aid condition, the stimuli were filtered into the eight bands and then the bands were merely summed back together again. We were unable to hear any differences between these unprocessed stimuli and the original digitized stimuli delivered through the same experimental system; normal subjects also performed the task equally well with either set of stimuli. In the unprocessed noise conditions, the noise was similarly filtered into eight bands and recombined.

In the conventional aid condition, each of the eight bands of the speech sound or speech sound plus noise was amplified according to the frequency and gain characteristics of the subject's own or independently recommended hearing aid. For those subjects using their own aids for the conventional aid condition, we did not always know the rationale by which the aid was prescribed. For the others, the independent hearing aid recommendations were made by a practicing audiologist in the community using the Carhart procedure (24). We chose the audiologist by professional recommendations and a personal interview and then placed essentially no limitations on her normal hearing aid evaluation procedure. In this way, we should have obtained a conventional aid control condition that represents a good fit for the subject in terms of frequency-gain characteristics and also a realistic hearing aid to use in the variety of conditions encountered outside the laboratory. The measured gain (Phonic Ear HC 2000 Hearing Aid Acoustic Computer) for each aid used is included in the data figures (**Figures 4–7**). In terms of the experiment, these gain curves represent the "real ear gain" of the conventional aid condition over the unprocessed condition; since both sets of stimuli were delivered to the subjects in exactly the same way, the gain curves in the figures illustrate the entire difference between unprocessed and the conventional aid stimuli. The curves were measured with the volume setting for most comfortable loud-

ness (MCL) by the subject with a 65 dB SPL input.

The model aid is an 8-channel compression aid where the amplification characteristic in each channel is determined with the model of Yund and Efron (25) using the method of Efron, Yund, and Divenyi (6) to measure the I-R functions. Since the model and the methods have been described elsewhere, only their application to the present problem will be presented here. I-R functions for a young normal-hearing subject at the frequencies 500, 750, 1000, 1500, 2000, 3000 and 4000 Hz are illustrated in **Figure 1**. Each panel has the right and left ear I-R function for one frequency. On the ordinate in each panel is the response of the model's intensity-response transduction stage; intensity is plotted on the abscissa. Note that the normal-hearing subject's I-R functions are essentially straight lines, except for 500 and 750 Hz. **Figure 2** illustrates I-R functions measured on our SNHL subject S4. This subject's I-R functions are typical of our SNHL population: not only do they begin at higher intensities (representing his elevated thresholds) but also the functions show much more curvature than those of the normal-hearing subject at the same frequency. (A more detailed examination of I-R functions in SNHL subjects is beyond the scope of this report and will be prepared for a future publication.)

Determining the amplification characteristic at a frequency using the normal and the SNHL I-R function at that frequency is quite simple. For

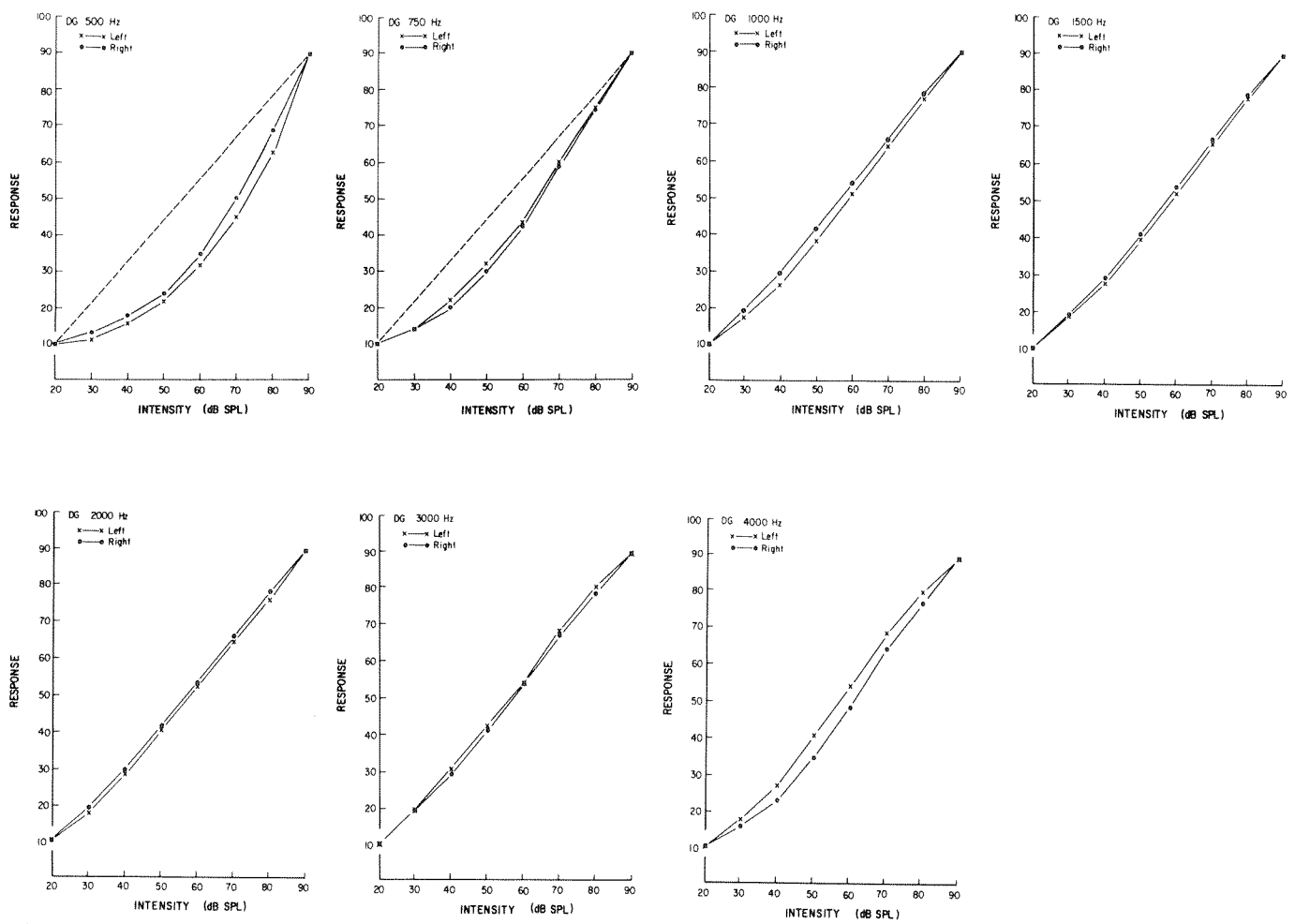


Figure 1.
I-R functions for normal-hearing subject, DG whose thresholds were 10 dB HL [ANSI (1)] or better between 250 Hz and 8 kHz in both ears. Each panel illustrates left and right ear I-R functions at the frequency given in the upper left of the panel. The response in arbitrary units is plotted as a function of intensity in dB SPL. The straight dashed line in the 500 and 750 Hz panels indicates the approximate position of the I-R functions for the higher frequencies for purposes of comparison.

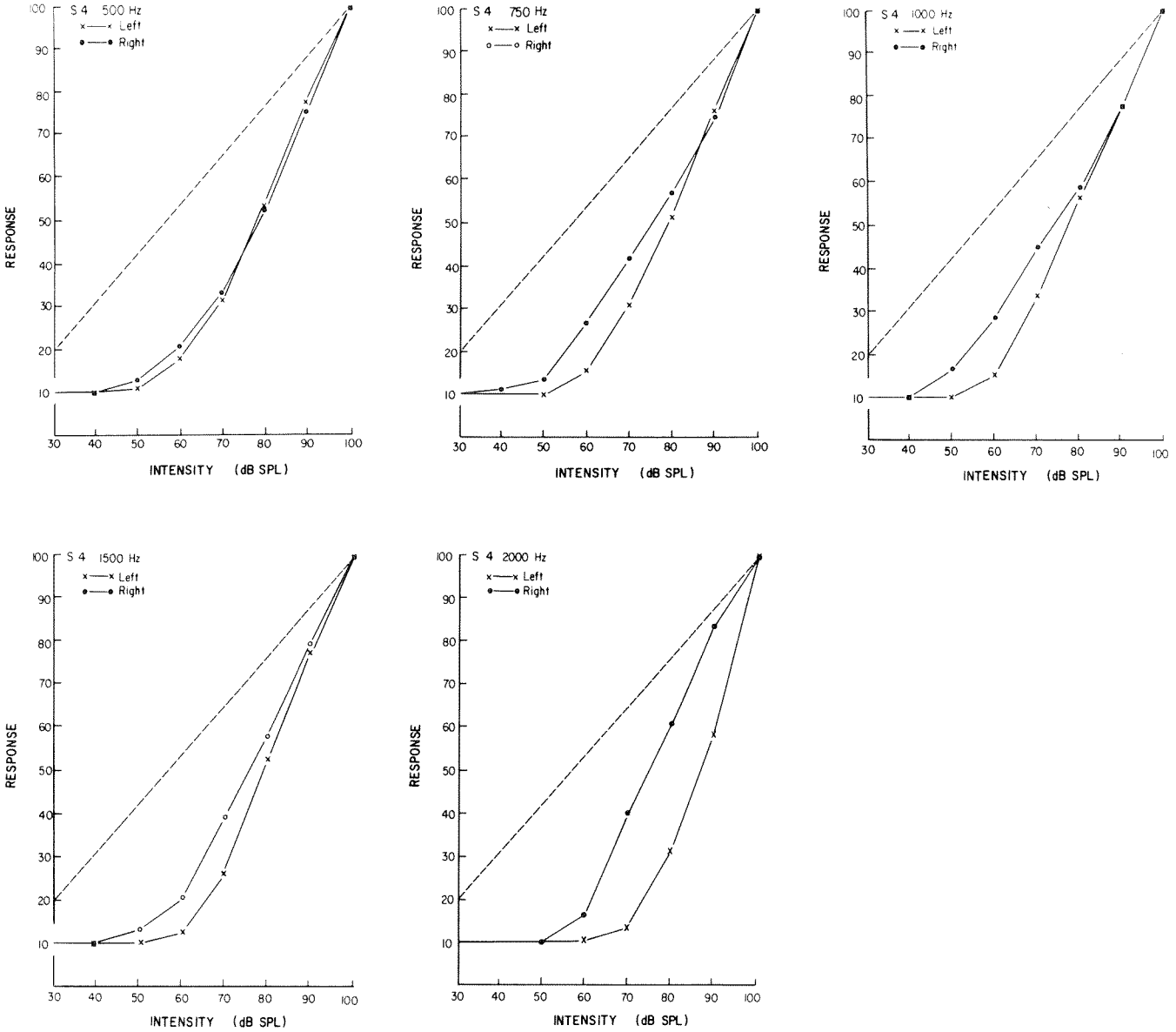


Figure 2.
I-R functions for SNHL subject S4. Each panel illustrates left and right ear I-R functions at the frequency given in the upper left of the panel. The response in arbitrary units is plotted as a function of intensity in dB SPL. The straight dashed line is in the same position as in the 500 and 750 Hz panel of the normal-hearing subject of **Figure 1** and thus at 1000, 1500 and 2000 Hz indicates the position of the normal-hearing I-R functions.

example, if we find that the normal curve yields a response of 20 for a 30 dB SPL input, then we look at the I-R curve of the SNHL subject to determine what input would be required to generate a 20 response—let it be 65 dB SPL. That would mean that the aid should amplify a 30 dB input at that frequency by 35 dB and thereby cause the impaired ear to generate the "proper" response, i.e., a response of 20. **Figure 3** illustrates such amplification values, plotted as a function of frequency and intensity, derived from the curves shown in **Figures 1** and **2**. **Figure 3** is thus a graphical representation of the amplification characteristics of our multichannel compression hearing aid for subject S4. Scaling assumptions in the I-R function calculations cause the amplification to fall to 0 dB at or below

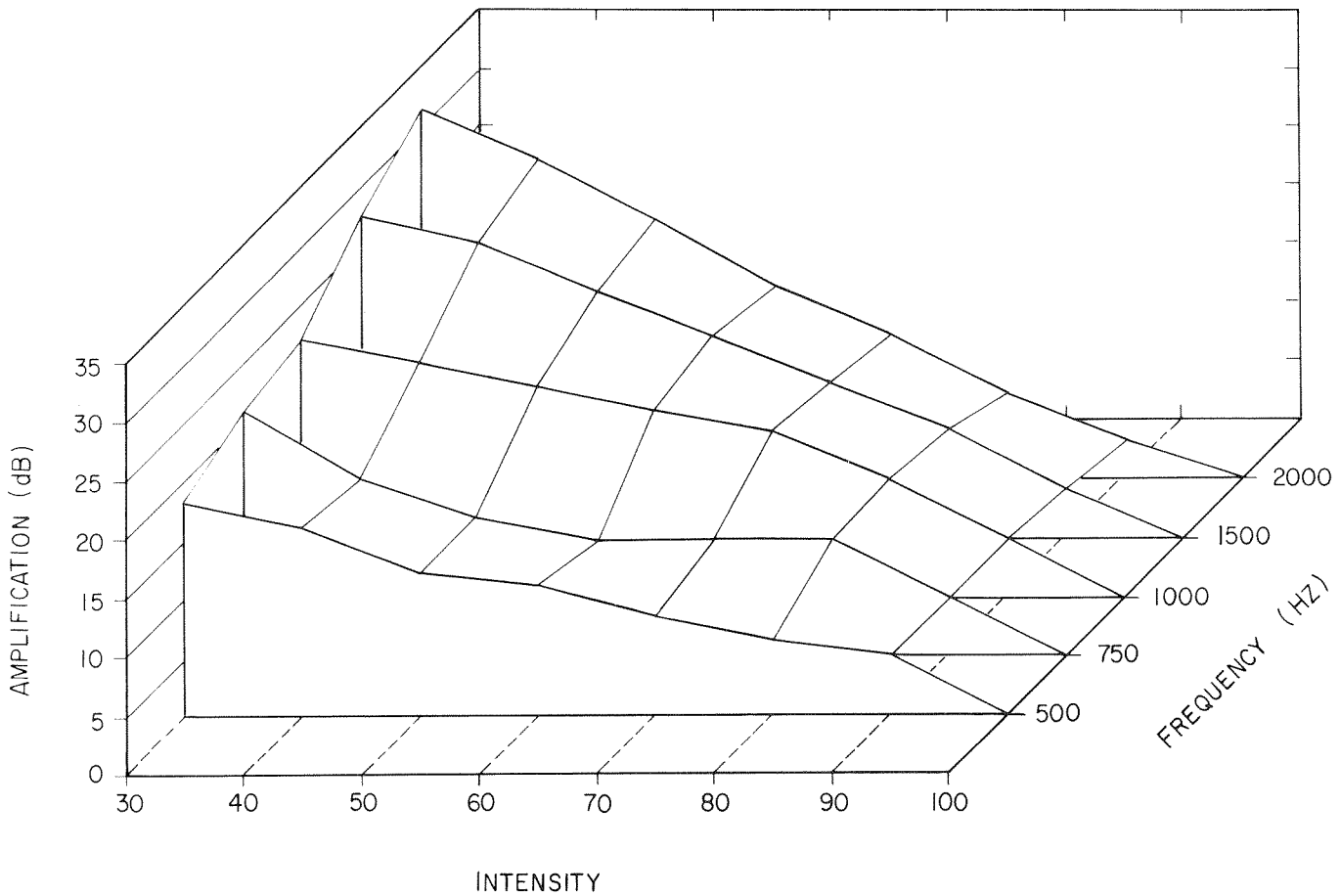


Figure 3.
The amplification characteristic for subject S4 from 30 to 100 dB SPL and 500 to 2000 Hz. See text for details.

100 dB SPL in all frequency bands and also prevent the model aid from ever producing any output level over 100 dB SPL. It should be emphasized that, although we are doing an intensity correction to effectively normalize the I-R function in each band, the methods used to measure the I-R functions all involve only pitch judgements. No loudness judgements are required for these corrections.

Several additional aspects of the processing system must be determined before the amplification characteristic of **Figure 3** could be applied to a stimulus. The parameters of the filters and the frequency bands used have been described above. We used the compression method suggested by Robinson and Huntington (18) with a 10 ms symmetrical averaging window; this yields an attack and release time, defined in the usual way, of approximately 4 ms. We have no reason to believe that these or other parameters are optimal since we made no attempt to optimize them.

RESULTS

Individual data for each of the 20 subjects are presented in **Figures 4–7**. The three panels in one row of each figure are for one subject. The left and middle panels in each row illustrate the data from the 60 dB and the 70 dB SPL SSN conditions, respectively. In these panels, percent correct syllable identification is plotted as a function of the signal-to-noise ratio (S/N). The three curves in each left and middle panel represent the subject's performance with unprocessed, or unaided, stimuli (filled circles and dotted line), with the individually recommended conventional aid (X symbols and dashed line), and with the 8-channel compression model aid (open circles and solid line). The unconnected diamonds represent the performance of normal-hearing subjects with the unprocessed stimuli. Each data point is a mean of five repeats of the 91 stimulus NST set. The panel on the right in each

row illustrates the subject's pure tone audiogram (filled circles and solid line) and the conventional aid gain (X symbol and dashed line). In this panel, dB HL or dB gain, respectively, is plotted as a function of frequency in Hz.

The subjects fell reasonably well into three groups: a) The ten subjects of **Figures 4** and **5** have the more gradually sloping audiograms. b) The five subjects of **Figure 6** have the more steeply sloping audiograms. c) The five subjects of **Figure 7** have losses that are different and/or more complex than those of the subjects in **Figures 4**, **5** and **6**. First we will point out some of the interesting aspects of the data in these figures and then we will present tables to summarize the statistical analyses that were done on each subject's data. In one sense, the statistical tables are almost unnecessary. On the scale at which the figures are drawn, points that are not touching each other always represent values that are significantly different from each other.

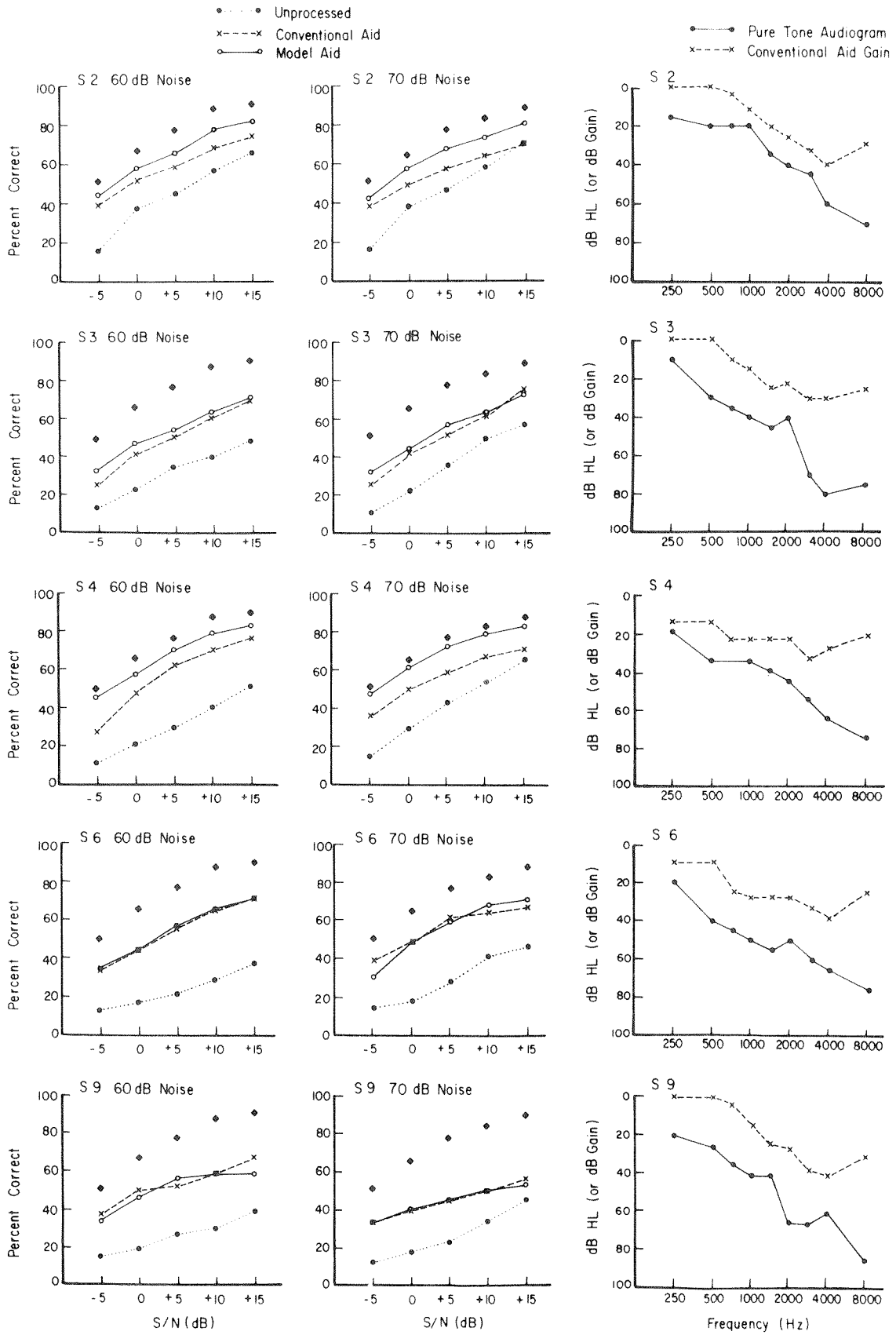
Consider, first, the data of the subjects with the more gradually sloping losses presented in **Figures 4** and **5**. Perhaps the most obvious aspect of these results is that the absolute and relative positions of the curves representing the unprocessed, conventional aid and model aid conditions varies considerably across subjects (different rows) and somewhat less across noise levels within subjects (left versus middle panel in the same row). Comparing the unprocessed results with the normal-hearing results indicates that all ten of these subjects have a clear deficit in recognizing speech sounds in noise. Of these subjects, S14 is least impaired with the unprocessed sounds while S19 is most impaired. Indeed, only S14 was able to perform much above the chance level of 12 percent at −5 dB S/N, a condition at which normal-hearing performance (diamond symbol) is about 50 percent.

Both conventional and model aids produce significant improvement over unprocessed performance for all but a few points and these all occurred at higher S/N where the unprocessed performance is at its maximum. For S14 there were even two conditions (+15 and +10 dB S/N, 70 dB SPL SSN) at which the aids produced a performance decrement. In four of these ten subjects, S2 and S4 in **Figure 4** and S15 and S17 in **Figure 5**, the model aid results are quite close to those of the normal-hearing subjects. It was S4 who produced the most consistently high scores with the model aid across all conditions and S17 who produced the only score equal to normal-hearing under any condition (0 dB S/N, 70 dB SPL SSN). In further comparing the model and conventional aid conditions, we see that six of these ten subjects (S2, S3, S4, S15, S17, S18) were able to obtain consistently higher scores with the model aid at 60 dB SPL SSN than with the conventional aid, while in no case in these subjects did the conventional aid outperform the model aid. At 70 dB SPL SSN, only S2 and S4 maintain the entire model aid curve above the conventional aid curve, although for S3, S17 and S18 the model aid curve is still predominantly above that of the conventional aid. For S19, however, the conventional aid performance was consistently superior to the model aid performance at 70 dB SPL SSN.

The difference in the relative performance of the aids between the 60 and 70 dB SPL SSN conditions, may represent an effect of learning to use the model aid rather than a true effect of noise level. Earlier subjects (S1-S10) ran a longer set of stimulus intensities without noise, a set which was reduced for the later subjects when it became clear that only the noise experiments were able to differentiate between aids. The later subjects in this group all ran the 70 dB SPL SSN condition before the 60 dB SPL SSN condition. Since the clear difference between noise conditions (model aid relatively better at 60 dB SPL SSN) occurred only in the later subjects (**Figure 5**), it seems more likely to be due to learning to use the model aid than to the difference in noise levels.

Figure 6 shows the results for subjects with steeply sloping audiograms. Although, as a group, these subjects benefited less from either aid, they demonstrated a very similar pattern of relative performances with respect to the model and conventional aids: The model aid performance was more often superior to that of the conventional aid. The model aid curve is consistently below the conventional aid curve at 70 dB SPL SSN for only one subject, S20 (possibly due to a learning effect, see above).

Figure 4. (*opposite page*)
The data for subjects in the first half of the gradually sloping audiogram group. Each row contains data from one subject. The left and middle panels illustrate results of the NST for the 60 and 70 dB noise conditions, respectively. Percent correct is plotted as a function of S/N for three conditions for the SNHL subject (three lines, see key above panels) and for normal-hearing subjects (unconnected diamond symbols). The right panel illustrates the SNHL subject's pure-tone audiogram with dB HL [ANSI (1)] plotted against frequency and the conventional aid gain with dB gain plotted against frequency.


Unprocessed
Conventional Aid
Model Aid
Pure Tone Audiogram
Conventional Aid Gain
S 2 60 dB Noise
S 2 70 dB Noise
S 2
S 3 60 dB Noise
S 3 70 dB Noise
S 3
S 4 60 dB Noise
S 4 70 dB Noise
S 4
S 6 60 dB Noise
S 6 70 dB Noise
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S 9 60 dB Noise
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S 9
Percent Correct
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Frequency (Hz)

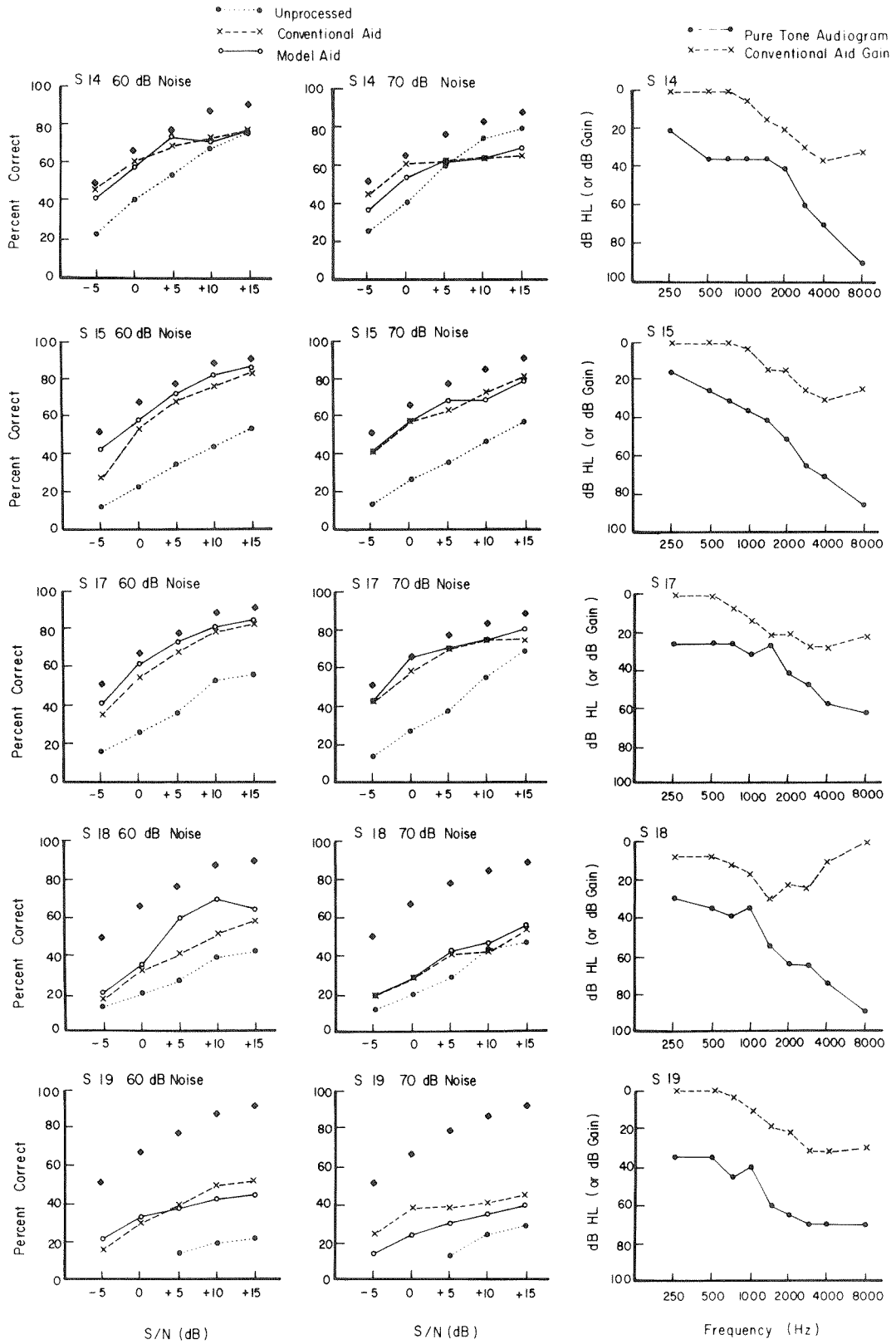

Figure 5.
The data for subjects in the second half of the gradually sloping audiogram group. Otherwise the same as **Figure 4.**

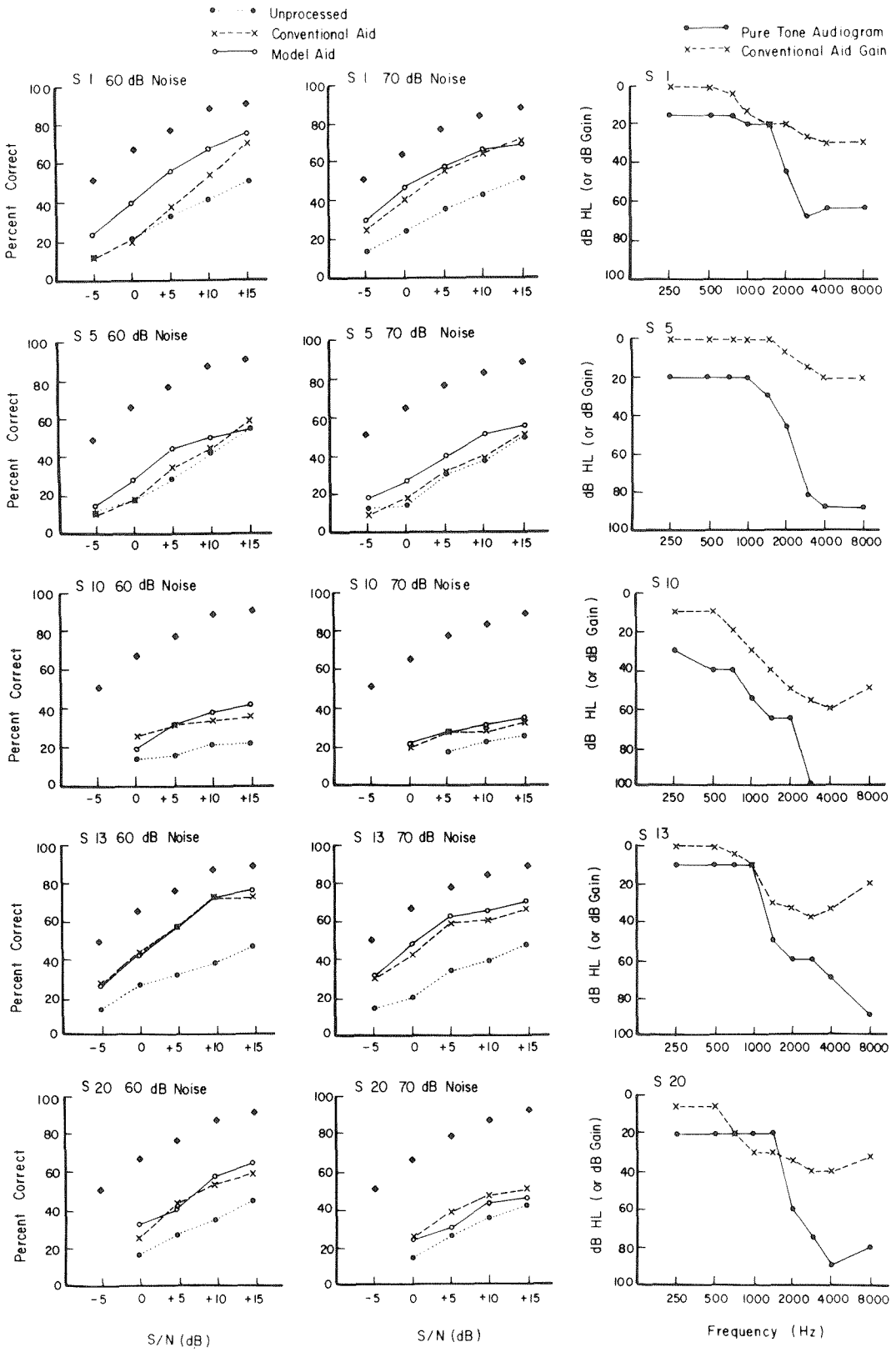

Figure 6.
The data for subjects in the steeply sloping audiogram group. Otherwise the same as **Figure 4.**

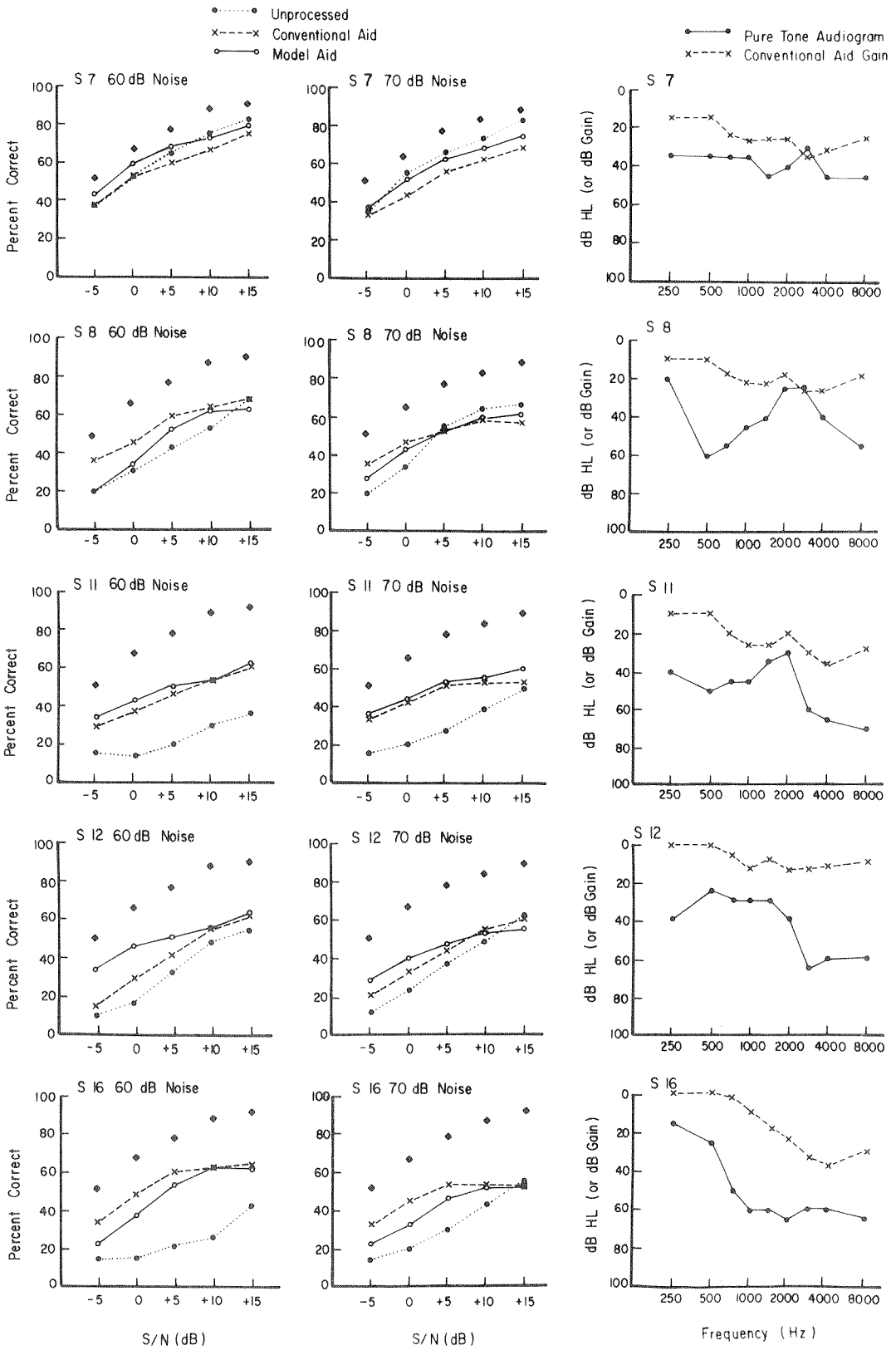

Figure 7.
The data for subjects in the miscellaneous audiogram group. Otherwise the same as **Figure 4.**

The results of the subjects in the miscellaneous category are illustrated in **Figure 7**. Of these, subject S7, with a relatively mild, flat hearing loss, produced higher scores with no aid (unprocessed) under six of the ten signal and noise combinations. Only at low S/N and at the lower noise level did the model aid prove to be of noticeable benefit, but it was always superior to the conventional aid for this subject. S8 and S11 both have reverse slope losses, but the conventional aid gave better performance than the model aid for S8 at most conditions and the reverse is true for S11. S12 almost belongs in one of the sloping-loss-subject groups, both in terms of his hearing loss configuration and his performance on the NST. S16 performed better with the conventional aid, but his results on the various pitch tests involved in the I-R function phase of the experiments indicated that he has a much more complex hearing loss than any of the other 19 subjects in the experiment. Consideration of the complexities of this hearing loss will be deferred until the discussion.

Now that the presentation of the data for all 20 subjects is completed, we will present the summary tables of the statistical analyses. An analysis of variance was done, individually for each subject and noise level, on the raw data used to compute the percent scores of **Figures 4–7**. These analyses are summarized in **Table 1** where the significance levels for the main effects of aids (unprocessed, model aid, conventional aid), and the interactions of aids and S/N, are given. The main effects of S/N are not included in the table because they were significant at the $p<0.001$ for all subjects at both noise levels. As expected, and as was seen in **Figures 4–7**, every subject performed better at higher S/N ratios. The main effects of aids were significant at $p<0.001$ for all subjects at 60 dB SPL SSN and also at 70 dB SPL SSN except for S8 (n.s.) and S14 ($p<0.05$). The interaction of aids and S/N was significant for 19 subjects at 60 dB SPL SSN (excluding S17) and for 17 subjects at 70 dB SPL SSN (excluding S5, S10 and S20).

Based upon these analyses of variance, Tukey WSD (10) post hoc analyses were performed for both the main effect of aids and the aids × S/N interaction. **Tables 2, 3,** and **4** illustrate, respectively, the comparisons of the model aid and the unprocessed condition (M vs U), the conventional aid and the unprocessed condition (C vs U), and the conventional and the model aid (C vs M). Within the body of these tables, both the superior condition and the significance level are given: a) The superior condition is identified by the letter in the table entry, U for unprocessed, C for conventional aid, and M for model aid. A hyphen (-) appears if neither was superior. b) The significance level is indicated by

Table 1.
Summary of Analyses of Variance.

	60 dB SPL SSN		70 dB SPL SSN	
Subject	**Aids**	**Aids X S/N**	**Aids**	**Aids X S/N**
S1	<0.001	<0.001	<0.001	<0.01
S2	<0.001	<0.001	<0.001	<0.001
S3	<0.001	<0.05	<0.001	<0.001
S4	<0.001	<0.001	<0.001	<0.001
S5	<0.001	<0.001	<0.001	n.s.
S6	<0.001	<0.001	<0.001	<0.001
S7	<0.001	<0.005	<0.001	<0.005
S8	<0.001	<0.001	n.s.	<0.001
S9	<0.001	<0.001	<0.001	<0.001
S10	<0.001	<0.001	<0.001	n.s.
S11	<0.001	<0.001	<0.001	<0.001
S12	<0.001	<0.001	<0.001	<0.001
S13	<0.001	<0.001	<0.001	<0.05
S14	<0.001	<0.001	<0.05	<0.001
S15	<0.001	<0.001	<0.001	<0.01
S16	<0.001	<0.001	<0.001	<0.001
S17	<0.001	n.s.	<0.001	<0.001
S18	<0.001	<0.001	<0.001	<0.05
S19	<0.001	<0.001	<0.001	<0.001
S20	<0.001	<0.01	<0.001	n.s.

the number of letters, one for <0.05 and two for <0.01. To make these summary tables slightly easier to read, upper case letters are used for the main effect columns and lower case letters for the interaction columns. **Tables 2–4** indicate with statistical significance essentially what we have already seen in the **Figures 4–7**. Either aid is almost always better than no aid and the model aid is more often better than the conventional aid. Looking specifically at the comparison between the model and conventional aids in **Table 4**, we see the following: a) Only in S8 at 60 dB SPL SSN was the conventional aid superior in the main effect. b) Only S16 never performed better with the model aid than the conventional aid, i.e., there is no "M" or "m" in the S16 row of **Table 4**. c) Five subjects produced a significant main effect in favor of the model aid (S1, S2, S4, S12, S18) at 60 dB SPL SSN and one, S4, maintained that at 70 dB SPL SSN. d) Six subjects never performed better with the conventional aid than the model aid (S2, S4, S7, S11, S17, S18).

DISCUSSION

Before beginning the discussion of these particular results and their relationship to other work in the area of multichannel compression hearing aids, two general points of major significance must be considered. The first is the general problem that any multichannel compression hearing aid is a very complex device with a large number of parameters whose values must be determined prior to its implementation. All of the frequency, intensity, and time parameters involved in such a device must be defined. While it is difficult to specify how many parameters such a device might have, it is clear that little is known about determining optimal values for any of them. We do not know which parameters might be critical or even how to set most of them so that their values will not have a negative impact on hearing in SNHL subjects. If, in addition, different parameters have different effects in different SNHL subjects, then the difficulty of developing a multichannel compression hearing aid increases even further.

In the face of such complexity, investigators still wishing to study multichannel compression must at the onset of a research program make many decisions based upon general knowledge and *a priori* theoretical assumptions concerning the nature of normal hearing and the nature of the functional deficit(s) in individuals with SNHL. To the extent that the general knowledge and theory is correctly applied to the problem and, even more importantly, to the extent that the assumptions concerning the

Table 2.
Summary of Tukey WSD Analyses (Model vs Unprocessed).

	60 dB SPL SSN						70 dB SPL SSN					
Subject	ALL	15	10	5	0	−5	ALL	15	10	5	0	−5
S1	MM	mm	mm	mm	mm	mm	MM	mm	mm	mm	mm	mm
S2	MM	mm	mm	mm	mm	mm	MM	mm	mm	mm	mm	mm
S3	MM	mm	mm	mm	mm	mm	MM	mm	mm	mm	mm	mm
S4	MM	mm	mm	mm	mm	mm	MM	mm	mm	mm	mm	mm
S5	M	—	mm	mm	mm	mm	—	mm	mm	mm	mm	mm
S6	MM	mm	mm	mm	mm	mm	MM	mm	mm	mm	mm	mm
S7	—	uu	uu	mm	mm	mm	—	uu	uu	uu	uu	m
S8	—	uu	mm	mm	mm	—	—	uu	uu	uu	mm	mm
S9	MM	mm	mm	mm	mm	mm	MM	mm	mm	mm	mm	mm
S10	MM	mm	mm	mm	mm	—	M	mm	mm	mm	mm	—
S11	MM	mm	mm	mm	mm	mm	MM	mm	mm	mm	mm	mm
S12	MM	mm	mm	mm	mm	mm	—	uu	mm	mm	mm	mm
S13	MM	mm	mm	mm	mm	mm	MM	mm	mm	mm	mm	mm
S14	M	—	mm	mm	mm	mm	—	uu	uu	—	mm	mm
S15	MM	mm	mm	mm	mm	mm	MM	mm	mm	mm	mm	mm
S16	MM	mm	mm	mm	mm	mm	M	uu	mm	mm	mm	mm
S17	MM	mm	mm	mm	mm	mm	MM	mm	mm	mm	mm	mm
S18	MM	mm	mm	mm	mm	mm	—	mm	mm	mm	mm	mm
S19	MM	mm	mm	mm	mm	mm	M	mm	mm	mm	mm	mm
S20	M	mm	mm	mm	mm	—	M	mm	mm	mm	mm	—

functional deficits in SNHL are correct, investigators studying multichannel compression should expect to find a significant degree of success, at least in some subjects. Given the complexity of the problem, a significant degree of failure should be expected in some subjects as well. In interpreting the results in such a context, a high degree of success in a small number of subjects offers critically important support for the basic idea. When such high performance is obtained in some subjects, relatively poor performance in other subjects can be viewed more as an opportunity to learn about the optimal parameters than as a general failure of the basic idea. Furthermore, statistical tests for an entire group of subjects should not be applied in this context and thus, there must be enough data over a large enough range of conditions within one subject to evaluate the performance of an aid on each subject individually.

A second general problem that has plagued research in multichannel compression hearing aids has been the question of how to evaluate any performance benefit obtained with such a device. The obvious answer to this question, "Use the appropriate other hearing aid, i.e., the best one not in the mulitchannel compression category," only further complicates the problem, since the optimal choice of parameters for such "other" hearing aids is not much better understood than it is for multichannel compression aids. But how can a multichannel hearing aid be evaluated without comparing its performance to that of the "other" aid? One critical point of comparison is the performance of the hearing-loss subject with no aid at all; clearly, any "good" hearing aid must provide a significant improvement over no aid at all. Another critical point of comparison is the performance of normal hearing subjects; clearly, any hearing aid that produced normal performance for a subject with SNHL, would be a very good hearing aid for that subject. Of course, any hearing aid (multichannel compression or other) would be expected to fall somewhere between these two extremes, but its position in that range is surely a critical piece of information about its performance. We would also argue, however, that some other hearing aid must be included in a multichannel compression study. The other aid is necessary if only to establish that the particular methods chosen do not produce virtually the same level of performance with any reasonable hearing aid. In this sense, the other aid becomes as much a reference condition to evaluate the experimental procedures as it is a control condition against which to evaluate the multichannel compression aid.

In our present experiments we chose as the other aid condition a commercially available hearing aid

Table 3.
Summary of Tukey WSD Analyses (Conventional vs Unprocessed).

	60 dB SPL SSN						70 dB SPL SSN					
Subject	**ALL**	**15**	**10**	**5**	**0**	**−5**	**ALL**	**15**	**10**	**5**	**0**	**−5**
S1	—	cc	cc	cc	—	—	CC	cc	cc	cc	cc	cc
S2	CC	cc	cc	cc	cc	cc	C	—	cc	cc	cc	cc
S3	CC	cc	cc	cc	cc	cc	CC	cc	cc	cc	cc	cc
S4	CC	cc	cc	cc	cc	cc	CC	cc	cc	cc	cc	cc
S5	—	cc	c	cc	—	—	—	—	cc	cc	cc	—
S6	CC	cc	cc	cc	cc	cc	CC	cc	cc	cc	cc	cc
S7	—	uu	uu	uu	—	cc	—	uu	uu	uu	uu	uu
S8	CC	—	cc	cc	cc	cc	—	uu	uu	—	cc	cc
S9	CC	cc	cc	cc	cc	cc	CC	cc	cc	cc	cc	cc
S10	CC	cc	cc	cc	cc	—	C	cc	cc	cc	cc	—
S11	CC	cc	cc	cc	cc	cc	CC	cc	cc	cc	cc	cc
S12	C	cc	cc	cc	cc	cc	—	uu	cc	cc	cc	cc
S13	CC	cc	cc	cc	cc	cc	CC	cc	cc	cc	cc	cc
S14	C	—	cc	cc	cc	cc	—	uu	uu	c	cc	cc
S15	CC	cc	cc	cc	cc	cc	CC	cc	cc	cc	cc	cc
S16	CC	cc	cc	cc	cc	cc	CC	uu	cc	cc	cc	cc
S17	CC	cc	cc	cc	cc	cc	CC	cc	cc	cc	cc	cc
S18	CC	cc	cc	cc	cc	cc	—	cc	—	cc	cc	cc
S19	CC	cc	cc	cc	cc	cc	CC	cc	cc	cc	cc	cc
S20	C	cc	cc	cc	cc	—	CC	cc	cc	cc	cc	—

fit by an independent practicing audiologist because that method at least provides a realistic control aid that could be used outside the laboratory. It can be argued that this is not the best possible other aid, but the actual frequency-gain characteristics of the conventional aids used (**Figures 4–7**, right panels) puts them well within the range of accepted values: our conventional aid frequency-gain slopes generally have more high frequency emphasis than the LDL characteristics (12) and less than the most extreme L4 curve used by Lippmann, Braida, and Durlach (14). Furthermore, our conventional aid was individually fit to the subject and it has the same fidelity and low distortion as the multichannel compression and unaided conditions because, except for the digital processing that represents the action of the other aid, all other parts of the system are identical. The above considerations suggest that only very minor differences (in an unpredictable direction) would exist between our conventional aid condition and linear control conditions used in other studies.

The results presented here (**Figures 4–7**) demonstrate that the model aid can be very effective for some subjects and is usually more effective than a well-fit conventional aid. In terms of the general evaluation criterion presented above, the most important results occur in 4 of the 20 subjects (S2, S4, S15, S17) where the performance with the model aid approaches that of the normal-hearing subjects on this same task. The two earlier subjects, S2 and S4 in **Figure 4**, showed this high degree of performance on both noise levels while the other two, S15 and S17 in **Figure 5**, showed such consistently high performance only on the 60 dB noise condition. These differences may be a result of a practice effect since both S15 and S17 ran the 70 dB noise condition before the 60 dB noise condition and the later subjects (S11 to S20) had only half the number of "quiet" sessions prior to "noise" sessions as did the earlier subjects (S1 to S10). Laurence, Moore, and Glasberg (11) report a practice effect with both their compression aid and their high fidelity linear aid and Braida et al. (3) discuss practice effects in experiments with a multichannel compression hearing aid.

Whether or not the results for S15 and S17 also include a practice effect, it is clear that in the case of 4 of the 20 subjects, the model aid was able to produce near-normal performance on this difficult task of speech recognition in noise. It should also be stressed that the near-normal performance occurred not only under the relatively easy, high S/N conditions, but also under the most difficult, low S/N conditions. This low S/N performance is especially important because it represents those conditions outside the laboratory where almost every

Table 4.
Summary of Tukey WSD Analyses (Model vs Conventional).

	60 dB SPL SSN						70 dB SPL SSN					
Subject	**ALL**	**15**	**10**	**5**	**0**	**−5**	**ALL**	**15**	**10**	**5**	**0**	**−5**
S1	M	mm	mm	mm	mm	mm	—	cc	—	—	mm	mm
S2	M	mm	mm	mm	mm	mm	—	mm	mm	mm	mm	mm
S3	—	—	mm	mm	mm	mm	—	cc	m	mm	—	mm
S4	MM	mm	mm	mm	mm	mm	MM	mm	mm	mm	mm	mm
S5	—	cc	mm	mm	mm	mm	—	mm	mm	mm	mm	mm
S6	—	—	—	—	—	—	—	mm	mm	cc	—	cc
S7	—	mm	mm	mm	mm	mm	—	mm	mm	mm	mm	mm
S8	CC	cc	c	cc	cc	cc	—	mm	—	—	cc	cc
S9	—	cc	—	mm	cc	cc	—	cc	—	—	—	—
S10	—	mm	mm	—	cc	—	—	mm	mm	—	—	—
S11	—	mm	—	mm	mm	mm	—	mm	mm	mm	—	mm
S12	MM	mm	—	mm	mm	mm	—	cc	cc	mm	mm	mm
S13	—	mm	—	—	cc	c	—	mm	mm	mm	mm	—
S14	—	—	—	mm	cc	cc	—	mm	—	—	cc	cc
S15	—	mm	mm	mm	mm	mm	—	cc	cc	mm	—	—
S16	—	cc	—	cc	cc	cc	—	—	c	cc	cc	cc
S17	—	—	—	mm	mm	mm	—	mm	—	—	mm	—
S18	MM	mm	mm	mm	mm	mm	—	mm	mm	—	—	—
S19	—	cc	cc	—	mm	mm	—	cc	cc	cc	cc	cc
S20	—	mm	mm	cc	mm	—	—	cc	cc	cc	—	—

normally available speech cue would be critical for maintaining communication.

It is interesting that all four of the subjects who achieved near-normal performance come from the group of ten subjects (**Figures 4** and **5**) characterized as having more gradually sloping audiograms. Furthermore, these four high performance subjects also have relatively mild hearing losses compared to the other subjects in that group. Mean thresholds for each of the ten subjects for 500-4000 Hz (see **Figures 4** and **5**) are, in order of increasing threshold: S2 (34.3 dB HL), S17 (35.0), S4 (44.3), S14 (44.3), S15 (45.0), S9 (47.1), S3 (48.6), S6 (52.1), S18 (52.8), S19 (55.0). Thus, according to mean threshold value, the four high performance subjects have the first, second, third, and fifth least severe hearing losses in this group of the subjects. Whether the severity of the hearing loss places some direct limitation on the individual's ability to extract information from the compressed signal or whether the effect is a less direct one (e.g., where learning to use the system is merely more difficult) cannot be determined at this time. Perhaps even subjects who are not severely impaired need training to effectively use the information in the compressed signals. It also would be valuable in future research to determine how well normal-hearing subjects perform these discriminations with compressed signals of the type used here.

None of the subjects in the group characterized as having more steeply sloping audiograms (**Figure 6**) achieved near-normal performance with the model aid. However, the two subjects in the steep-slope group who achieved the highest level of performance with the model aid, S1 and S13, were also the two who have the least severe losses. S1 and S13 are the only subjects in this group with thresholds below 80 dB HL for 2, 3, and 4 kHz. Overall, subjects in this group demonstrate a less consistent advantage for the model aid over the conventional aid, especially in the more severe cases (S10, S20). Perhaps there are special problems in implementing a compression mechanism in a frequency region where the thresholds, and thus also the predicted amplification functions, are changing very rapidly. Another possibility (surely true, at least, for S10 above 2 kHz) is that the residual sensitivity at higher frequencies is insufficient to be useful in extracting information from the signals in the higher frequency channels. A third possiblity, already suggested above, is that more learning is required for the subject to extract this information in bands where the hearing loss is greater.

One of the subjects in the miscellaneous group, S7 of **Figure 7**, showed a high level of performance with the model aid, especially at −5 to +5 dB S/N with 60 dB noise. However, her unaided scores were so high under all conditions that, outside the laboratory, she would probably not find any hearing aid worthwhile at this time. Subject S8 has a reverse slope hearing loss between 500 Hz and 2 kHz with very little threshold elevation at 2 and 3 kHz and an increasing loss beyond 3 kHz. He performed the speech task better with the conventional aid than with the model aid, perhaps because the model aid amplified signal components in the lower frequencies more, and those in the 2 to 3 kHz range less, than the conventional aid. Lutman and Clark (15) reported that 2 kHz was a critical frequency in determining speech performance in their experiments. Clearly, the model aid does not work well in the case of a hearing loss like that of S8. S11 has a less extreme reverse-slope audiogram and performed better with the model aid than the conventional aid. She also showed much more of an unaided deficit than did S8. Future studies on S8 and S11 as well as other such subjects might yield a better understanding of such subjects' abilities to use multichannel compression systems.

S16 is a subject who demonstrated several unique aspects in his hearing loss: a) The sloping portion of his audiogram is all in the region around 500 Hz. b) From 1 to 8 kHz, his audiogram is virtually flat. c) A 90 dB SPL speech signal with conventional aid processing was too loud (so uncomfortable that he listened to only a few such stimuli, although his conventional aid did include output compression). d) Loudness discomfort levels were less than 100 dB SPL above 750 Hz. e) Above 1 kHz, he displayed an abnormally large amount of binaural diplacusis with the right ear needing a frequency about 10 percent higher than the left to yield a matching pitch up to 3 kHz but then approaching 4 kHz the match was about 7 percent in the opposite direction. [According to van den Brink (20), normal differences are less than 4 percent; also see Shambaugh (19) for a discussion of binaural diplacusis in patients with hearing loss.] In S16, the conventional aid was generally superior to the model aid and it is tempting to speculate that the success of the conventional aid is due to the fact that it amplifies less in the region from 750 Hz to 3 kHz than one might want

in looking at the audiogram shape. Further study of this subject could be very interesting from both a clinical and a theoretical perspective.

To the extent that there is any pattern in the model aid results across subjects, the model aid was least effective in comparison to the unaided condition at the high S/N and that effect was more apparent at the 70 dB noise condition than at 60 dB noise. In this case, we cannot attribute the 70 dB versus 60 dB noise difference to learning to use the model aid because S7 and S8, who most clearly show the effect, are among those subjects who ran the 60 dB noise condition prior to the 70 dB noise condition. Nevertheless, the overall effect as a function of S/N is not very surprising, since at the higher S/N the signal components are at higher intensities than noise components. Under these circumstances, the model aid tends to amplify signal components less than noise components and, in effect, reduce the S/N. The difference in effects at 60 dB and 70 dB noise may be a result of a greater effective S/N reduction because of where stimulus and noise components fall on the amplification functions at the different intensities or a result of increased masking at the higher intensity. At low S/N the effective reduction of S/N no longer occurs (since signal and noise are at almost the same level) and other, beneficial effects may occur: a) Low level signal components in channels where relatively little noise occurs may be amplified—this is essentially the explanation that accounts for the positive effect of the "frequency equalized" linear aid at low S/N. b) Signal components in channels where there is a relatively high average noise level may occur at times when the noise is, by chance, at a low level. If the system can respond quickly enough to the chance reduction of the noise and increase its amplification to bring that signal component to a level that the subject can use, it could produce an effective increase in the overall S/N relative to either the unprocessed or the conventional aid condition. This explanation for the compression aid's success at low S/N is related to what presumably is happening in the normal-hearing subjects without any external amplification. Any signal components such as those described above would be helpful to the normal-hearing subject because they are suprathreshold and thus can be used whenever a negative fluctuation in the noise permits them to be detected. It should be noted that point (b) above would not apply to all or even most multichannel compression systems, but only to those with narrow channels and short release times and where at least the noise components were in the compression phase of the channel involved. The narrow channels and short release times give the system an opportunity to apply its compressive function to enhance the lower level signal relative to the noise.

Points (a) and (b) in the previous paragraph may explain why Walker, Byrne, and Dillon (22), using the same type of stimuli, found primarily negative results for a multichannel compression/expansion system. Low level components either in a different channel or in the same channel at a different time are relatively reduced in the expansion range of the device. They found positive results only at the high intensity condition where stimuli were mainly in the compression range of the device. Indeed, according to the rationale of the model aid, restricting the compression phase of such a device to higher intensities (with either linearity or expansion at lower levels) would increase the difference between the neural information coming from the cochlea in the SNHL subject as compared to the normal-hearing subject.

Barfod (2) used a similar rationale and equal loudness contours to determine "hearing loss" as a function of frequency and intensity. Those hearing loss functions (his Figure 3) are not dissimilar to the amplification function plotted in our **Figure 3**. He demonstrated that a 4-channel compression system based on the hearing loss functions was as good as the optimized linear control for speech at 65 dB SPL with noise at various levels (S/N from +15 dB to −5 dB). It is probably true, as Barfod asserts, that the 4-channel compression system would work better than the linear control at other speech intensities, but that should be demonstrated. Barfod's data would be much easier to interpret if normal-hearing results and these subjects' unaided results had been included. To apply the second general point from earlier in this discussion, consider how differently one might interpret Barfod's data if the performance of the 4-channel and the linear systems were near-normal as opposed to near-unaided. Consider also, how easy it would be to compare Barfod's data to the data presented here if it were possible to normalize the performance of any subject in either study to a numerical scale where 0 represented unaided and 100 represented normal.

In summary, the 8-channel compression hearing aid studied here was designed to precisely compensate for deficits in the subject's cochlear I-R transduction mechanism, deficits that vary with frequency and intensity. The particular I-R functions used here were developed from a model formulated to explain dichotic pitch interactions. The model aid is a compression device because of the shape of the I-R functions and a multichannel compression device because that shape varies with frequency. Although no other specifications of the aid are defined by the model, many narrow channels with fast acting, nondistorting compression mechanisms seem desirable. Using a broad range of test conditions (55 to 85 dB SPL speech sounds and 60 and 70 dB SPL speech spectrum noise), we have shown that a particular 8-channel compression system developed around model I-R functions can be a very effective hearing aid. The aid is more effective for more gradually sloping audiograms and for less severe hearing losses. Four subjects with less severe, gradually sloping losses produced near-normal performance in the model-aid condition. Some aspects of the present results suggest that learning to use the model aid may be important. Future work will include detailed analyses of the error patterns in the speech results presented here as well as other psychophysical studies on these subjects in an attempt to understand the differences among subjects and their differential success with our 8-channel compression system.

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