HEARING AIDS - PAST, PRESENT AND FUTURE: SNR LOSS COMES OF AGE

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I. THE PAST

In high-level noise, hearing aids used to make it hard to hear. This was shown clearly in a series of studies reported by Tillman, Carhart and Olsen (1970), where the hearing aid they used caused those with normal hearing to require 9 dB greater signal-to-noise ratio (SNR) listening through that hearing aid compared to listening with their own ears. As shown in Figure 1, their subjects with presbyacusis typically required 14 dB greater SNR for 50% correct scores under the aided condition, on top of their 13 dB deficit unaided. In all cases, the task was NU-6 words against competing sentences (single talker). The study was designed so that audibility was not a problem; each test was performed at 30 dB sensation level. (Thus the unaided comparison in Figure 1 might better be called "loudspeaker aided," since the sound-field presentation level was often quite high.)

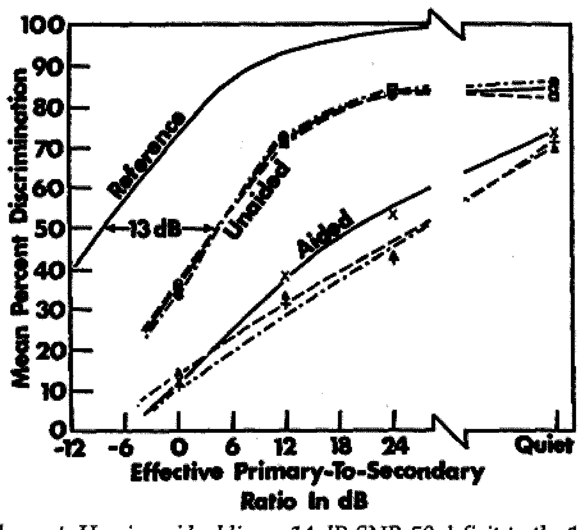

Figure 1 The past: Hearing aid adding a 14 dB SNR-50 deficit to the 13 dB deficit of 13 subjects with sensorinueral loss (Tillman et al, 1970, Figure 11).

The Tillman et al study was important for another reason, since it clearly established that a wideband sound-field audiometer could provide better amplification than the 1960's body-worn hearing aids they tested. Interestingly enough, Bentler and Duve (1997) recently confirmed this result: their tests included a comparison between an early body aid and high-level, unaided presentation of speech in noise.

In his Ph.D. research, Pascoe (1975) showed that the narrow-band frequency responses typical of the hearing aids of that time seriously degraded intelligibility in noise for his hearing-impaired subjects, subjects who had been chosen as typical of the hearing aid wearers he was regularly seeing at Central Institute for the Deaf. From Pascoe's data, we can estimate that the wideband aids with appropriate frequency response improved by 10 dB the SNR for 50% correct (we will denote this as SNR-50 in much of what follows.) Hal Davis -- principal investigator on the 1947 Harvard Report -- was on Pascoe's Ph.D. committee. Since Davis had previously summarized those prestigious 1947 findings as recommending "a frequency response...from 300 to 4000 Hz" and that "The idea of individual selective amplification is fallacious," you can imagine the care with which Pascoe conducted his study. Pascoe's study was a landmark study that helped change our thinking. Until then, the common understanding went something like this: "By limiting the bandwidth of the hearing aid to the most important speech frequencies (500 to 4000 Hz), we filter out bothersome noise at high and low frequencies where there is little speech information." I can remember Hugh Knowles explaining that to me in the early 1960's when I entered the hearing aid field, and wondered why we didn't make wider-bandwidth transducers. I once ran across an old-fashioned glass lantern slide of Hugh's whose data illustrated there was little speech information contained between 4000 and 7000 Hz.

Subsequently, Skinner, Karsted, and Miller (1982) and Skinner and Miller (1983) demonstrated that -- at least for the nine moderately-severely hearing-impaired subjects they tested -- a frequency response extending from 250 to 7000 Hz provided significantly better intelligibility than the narrower band-widths typical of hearing aids of that time.

To the question "Why some hearing aids don't work well?" Killion (1994) suggested that the narrow bandwidth, peaky response, and nonlinear distortion of some hearing aids substantially reduced intelligibility in high-level noise compared to the unaided condition.

II. THE PRESENT

A. CIRCUITS

Many hearing aids of more recent design, both analog and digital, provide just as good intelligibility for high-level sounds (sounds heard clearly without a hearing aid) as can be obtained unaided.

Before proceeding, it is important to remember that all hearing aids, past and present, have improved the intelligibility of low-level speech in noise. In many cases, the wearer would have been unable to hear either the speech or the noise unaided. As Pascoe (1980) once said, "Although it is true that mere detection of a sound does not ensure its recognition, it is even more true that without detection the probabilities of correct identification are greatly diminished."

For a while, however, we hoped for more help than audibility assistance from the hearing aid. We hoped that some sort of analog or digital speech processing would help us hear the speech we wanted to hear and ignore the interfering speech we didn't want to hear. In the most general case, it was obvious that improving on the brain would be difficult. We all use what Broadbent (1958) called "selective listening" to separate speech from noise and to choose which speaker we wish to follow at a social gathering. As Pascoe showed, in low- and moderate-level noise, the hearing aid can be an enormous help. It does this by making redundant speech cues audible so that more cues are left after noise has taken its toll (Villchur, 1992).

But there is no way (at present) that the hearing aid can know which talker you wish to follow, especially since a few moments later you may be tired of listening to the person in front of you and prefer to turn your attention to the more interesting conversation off to the side.

> Note:
> At some time in the future, perhaps, we will have a hearing aid circuit that can be switched into a "search" mode, whereupon it will lock onto one talker after another until, as with your car radio, you push the stop button when it comes to the speaker you wish to follow. At the moment, however, the closest we can come to selective listening by a hearing aid is to use the techniques described below: Use of directional microphones and close-talking microphones.

There is now a fair body of evidence that says that hearing aid circuits improve intelligibility in (speech) noise only by making things more audible. The most recent check on this conclusion may have been the study of "hearing aid benefit over the ages" reported by Bentler and Duve (1997). They measured their subjects intelligibility (and SNR) at sound-field presentation levels of 53, 63, 83, and 93 dB SPL, for a variety of hearing aids: four modern high-tech circuits (two analog and two digital), a not-so-modern peak-clipping narrowband hearing aid design, a 1934 body aid (fresh out of the original box and working properly), and an 1800's speaking tube held at the side of the head to form a low-gain ear trumpet.

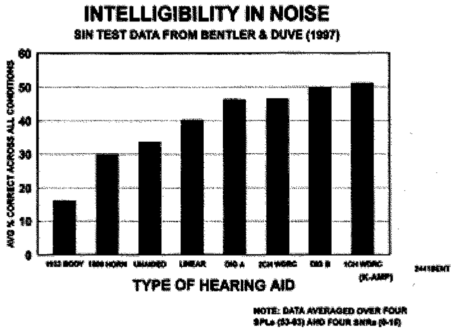

Figure 2 Aided intelligibility in noise for seven hearing aids averaged across four SPLs (53-93) and four SNRs (0-15) (Bentler and Duve 1997).

I was pleased to see that across all the levels and SNRs, the hearing aid circuit that Etymotic Research designed in 1989 came out best -- by a small margin -- as shown in Figure 2. The point for our present argument is shown in Figure 3, however. None of the high-tech hearing aids did significantly better at high input levels where everything was <u>already audible</u>, than unaided hearing or even an 1800's speaking tube. (Although it had a moderately wide bandwidth, no one expected that the speaking tube would do so well. We all realized afterwards that if it had been tested as normally used, with the mouthpiece at the lips of the talker, it would have won hands down over all the electronic hearing aids, which use headworn microphones!)

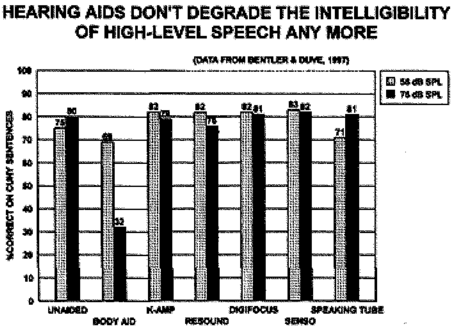

Figure 3 High level performance of four modern hearing aids, an 1800's speaking tube and unaided hearing: once everything is audible further improvement requires improving the SNR at the hearing aid input (Bentler and Duve 1997).

B. DIRECTIONAL MICROPHONES

To date, the only way to improve the SNR at which listeners can understand speech has been to increase the SNR itself at the input to the hearing aid circuit. Close talking and directional microphones can do this.

1. Close Talking Microphones.

The SNR-50 scores of hearing-impaired subjects can be improved by 15-20 dB or more if they use 1800's speaking tubes properly. Those wishing to avoid the mechanical link between talker and listener can obtain a similar benefit with an FM system. We are most familiar with these as used in the classroom with hearing-impaired children, but today's smaller FM systems make them attractive for a wider audience. Phonak has an integrated circuit FM receiver that can fit into an ITE or smaller hearing aid, for example. If the listener is willing to hand the transmitter to the talker of interest, even someone with severe-profound loss can carry on a conversation comfortably without wires or speaking tube.

These FM systems are capable of a 15 to 20 dB improvement in SNR, because they move the pick-up microphone from the listener's head to the talker's mouth, where the talker's level will be 15 to 20 dB higher. (The SPL right at the mouth can run 110 to 120 dB or more.)

2. Directional Microphones.
While moving the microphone closer to the talker's mouth gives the greatest known SNR improvement, a significant improvement can be achieved in most situations with head-worn directional microphones. First-order (hypercardioid, for example) head-worn directional microphones can improve the SNR by nearly 5 dB; head-worn array microphones can improve it by nearly 10 dB.

Using real-world binaural recordings of head-worn directional and omni-directional hearing aid microphones and subjects with flat and sloping sensorineural loss, we have confirmed that:

a. **Two ears are better than one.** When binaural presentation was compared to the presumed-better-ear monaural presentation, the average SNR for 50% correct score (SNR-50) was 3 dB better. This held true when either omni-directional or directional microphones were used (Chung, 1999).

b. On recordings made in three locations -- two restaurants and one cocktail party -- **the average indoor improvement in SNR was approximately 4 dB** (Killion et al, 1998), similar to the 4.4 dB we predicted for such "diffuse-field" conditions from our lab measurements on KEMAR (Roberts and Schulein, 1997).

c. On the recordings made on Bourbon Street in New Orleans, where each of the bars on the four corners had a band playing and the street was crowded with people drinking, talking, and laughing, **the average outdoor improvement in SNR was 8 to 10 dB.** This better-than-expected improvement came about because of the outdoor environment, where few sounds were reflected back to the listener and the major contributors to the 95 dBA overall level were the two loud bands located over the shoulders of the live "recording dummy."

III. THE FUTURE

A. NEW DIRECTIONAL MICROPHONES
Since only close-talking and directional microphones have improved the ability of hearing aid wearers to hear in high-level noise (recall that almost any hearing aid can help in low-level noise by increasing audibility), these microphones are likely to be the most important continuing developments.

For those with severe-profound SNR loss, we have been working on a multiple-talker wireless system. At a restaurant, for example, a hearing-impaired listener could pass a miniature microphone-transmitter to each person at the table.

This approach can provide 20 dB improvement if the microphones are close enough to the talker's mouth, even with three microphones live.

For those with somewhat less SNR loss, Wim Soede (1990) showed that suitable binaural head-worn array microphones could provide approximately 8 dB SNR improvement over the open ear. We have been working with Soede to develop an improved version. We hope to shortly introduce this improved array microphone, which provides an SNR improvement of 10 dB over the open ear.

Present headworn directional microphones can provide some 2 to 3 dB of improvement (BTE) and 3 to 5 dB improvement (ITE) in SNR compared to omni microphones. Directional microphones with reduced size, improved stability, and reduced noise are in the works. We have recently been concentrating on the limitations we once thought would stop us from making effective canal-aid and CIC directional microphones. It now appears that it may be possible to obtain close to a 5 dB SNR improvement even with such microphone locations. We have also seen indications that BTE directional microphones with improved directivity may be possible.

In all of this, there are two popular approaches to directivity: a) subtract the output of two omni microphones after treating one to an electrical time-delay network, or b) use the two sides of the diaphragm in an omni microphone to subtract the two acoustic signals after delaying one of them acoustically. Either approach can yield good directivity, although the single-cartridge approach gives lower self-noise and is less subject to loss of directivity from transducer drift.

B. NEW CIRCUITS

Needless to say, circuit and transducer improvements can be expected to continue. Although the predicted demise of analog circuits (for all applications including hearing aids) has been postponed, there is no question that digital technology can do nearly as well as analog technology even today. The analog-to-digital converters in today's hearing aids (presently 13- or 14-bit equivalent before analog assistance) are approaching the 16 bits used in CD recordings. In the future, continued improvements in digital technology should bring us to the day when high-quality signal processors for hearing aids will be best made digitally. In my view, continued improvements in analog technology have been pushing that date into the future.

Although most experiments indicate that the beneficial effects of compression are restricted to keeping signals audible, reducing distortion, and providing comfortable adjustment-free listening, that is not so bad. The trend toward the use of wide dynamic range compression will surely continue, despite the large U.S. NIDCD/VA experiment showing compression limiting was preferred more often than wide dynamic range compression (Haskell 1999). That experiment used variable-recovery-time compression for compression limiting but fixed-recovery-time compression for the wide-dynamic-range compression. A possible explanation can be found in the experiments of Fikret-Pasa (1993) who showed a several-dB improvement in SNR-50 with variable-recovery-time compression over fixed-recovery-time compression.

C. NEW TRANSDUCERS

Today's transducers are greatly improved over those I helped design 15-35 years ago. At that time, receivers often failed at 2000 to 4000 g of shock; some now survive 20,000 g of shock. Microphones have long survived 20,000 g of shock, but their noise level for a given volume has been dropping steadily. Today's smallest hearing aid microphones are nearly as quiet as the ceramic and electret microphones of the late 60s. Indeed, we have recently assembled a 7.6 mm (0.300") diameter microphone -- from seven 2.5 mm (0.100") diameter microphones -- with an A-weighted noise level of 18 dB equivalent SPL. A microphone 7.6 mm in diameter and 2.5 mm high would have been considered very small when I first joined Knowles Electronics.

The following statement was already true 20 years ago: Available transducers provide no limitations to hearing aid fidelity, permitting 16 kHz bandwidth and low distortion. For example, Etymotic Research's 16 kHz bandwidth ER-4 high-fidelity insert earphones have received the highest reviews in high fidelity magazines; they use modified Knowles ED-model receivers. Reduced vibration output from the receiver is a long-standing goal of receiver manufacturers, since it allows increased gain before feedback. A dramatic reduction in vibration (and increase in gain before feedback) can be obtained by mounting two matched receivers belly to belly and driving them electrically in phase, following Harada (1989).

IV. SNR LOSS

The irony is that even though "trouble hearing in noise" is the biggest complaint of hearing aid wearers, almost no one knows how much SNR loss a given patient has. It appears that the two available tests for measuring SNR loss -- the HINT test (Nilsson, Soli and Sullivan, 1994) and the SIN test (Killion and Villchur, 1993; Fikret-Pasa, 1993) take too much time to be readily adopted for clinical use.

If everyone with hearing loss had the same audiogram, we would not need to measure thresholds. Similarly, if everyone with a given hearing loss had the same SNR loss, we could predict SNR loss from the audiogram and not have to measure it. Unfortunately, the correlation between threshold loss and SNR loss is not strong.

A. THE DATA

Many studies of speech in noise have been reported, especially by Plomp and his colleagues (Plomp et al, 1978 and subsequently). Dirks, Morgan, and Dubno (1982) studied the reception of NU-6 words in noise, reporting a range of 25 dB in SNR-50 across their hearing-impaired subjects.

Lyregaard (1982) studied the reception of CVCV logotomes in speech-spectrum-shaped Gaussian noise, and reported the *SNR-50 results as a function* of pure-tone averages. His data for sensorineural subjects are shown in Figure 4.

More recently, studies using the HINT test have produced data for subjects with normal hearing (Hanks and Johnson 1998) and mild and moderate hearing loss (Bentler and Duve 1997 and Bentler 1999) have become available. Those data are shown in Figure 5.

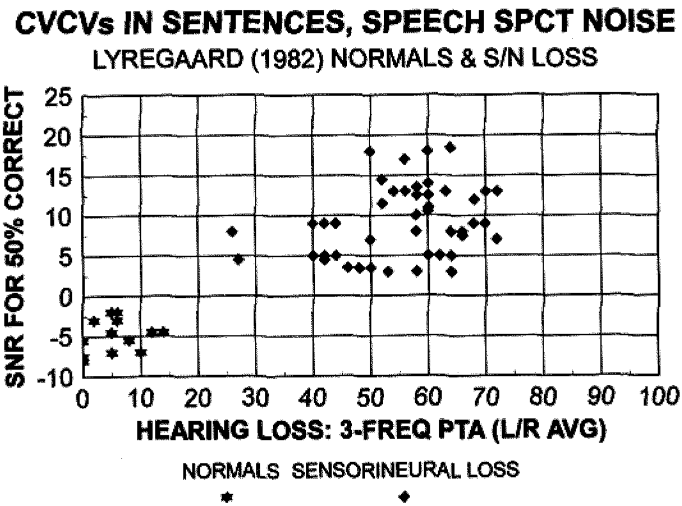

Figure 4 SNR-50 vs. hearing loss. CVCVs in sentences against speech spectrum noise (Lyregaard 1982).

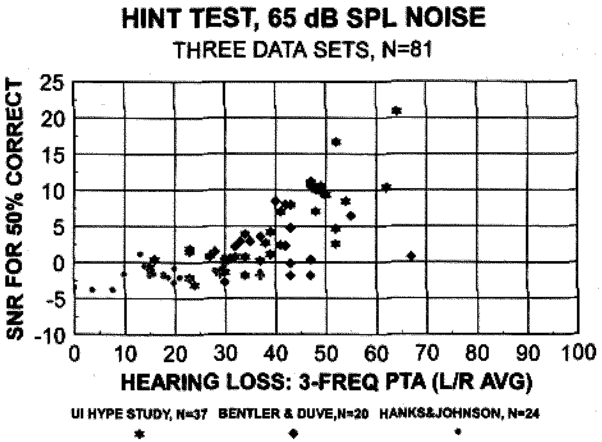

Figure 5 SNR-50 vs. hearing loss. HINT test against 65 dB SPL noise.

Most recently, data on hearing-impaired subjects has been obtained using the SIN test, where five key words in each sentence are scored using half-word scoring (a response containing one or more correct or similar-sounding phonemes is given half credit). The IEEE sentences are recorded by a female talker and tested against a babble of four talkers (three female, one male), each reading a different passage. This configuration was chosen by Fikret-Pasa (1993)

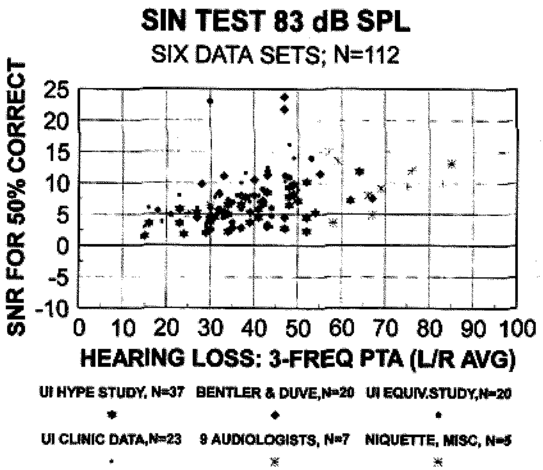

Figure 6 SNR-50 vs. hearing loss. SIN test at 83 dB SPL.

after a search for a realistic test simulating normal social conditions. Figure 6 shows the SNR-50 vs. pure-tone averages for the available SIN test data. Most of these subjects were hearing aid users.

Statistics note: The reliability of an SNR-50 SIN-test score obtained from a single block of sentences (five each at 0, 5, 10, and 15 dB SNR) has a standard deviation of 0.7 dB. For word scoring, the statistics of each five-sentence, 25-word score with the recommended half-word scoring is equivalent to that of normal 25-word lists with whole-word scoring: 10% for scores near 50%. In other words, the use of half-word scoring almost exactly cancels the sentence effect (which tends to give an all-or-none score to words in a sentence). While the words in IEEE sentences such as "Her purse was full of useless trash" (my favorite) are not completely independent, neither are they entirely predictable from each other.

The SNR-50 standard deviation can be predicted from the word-score standard deviation by remembering that typically two or more sets of 25-word sentences go into any SNR-50 determination, and that scores for normal subjects increase about 10% per dB SNR.

In any case, some of the points in Figure 6 have a standard deviation of 0.4 dB, being the average of four SIN-test blocks. None of the points has a standard deviation of more than 0.7 dB, so the large spread in data shown in these figures indicates real differences in ability to hear in noise.

In two separate experiments, Bentler and Duve (1997) and Bentler (1999) obtained both HINT test data and SIN test data on a total of 57 subjects with hearing loss. The comparison between the two tests is shown in Figure 7. The correlation is reasonable, but emphasizes the fact that any test only tests what it tests, and can't be expected to exactly predict the results of a different test.

Figure 8 shows the distribution of SNR loss based on earlier SIN test results for patients who purchased hearing aids. Those data indicate that roughly half of hearing-impaired subjects have more than a 5 dB SNR loss, meaning they require 5 dB greater SNR than normal to obtain 50% correct word-in-sentence scores[1]. A few have a 20 dB SNR loss. Just as important, it is not possible to determine from the audiogram anything more than the likely range of the SNR loss. For almost any hearing loss between 30 dB and 60 dB, we find some patients with nearly normal SNR-50 scores, and others with up to 20 dB SNR loss.

1 Limited SIN test results on 55-80-year-old subjects with normal hearing indicate that their SNR-50 may average 2-2.5 dB higher than that of young normal graduate students on which most of the SIN test norms for normal-hearing subjects were base. This would be consistent with the findings of Dubno and Dirks (1984).

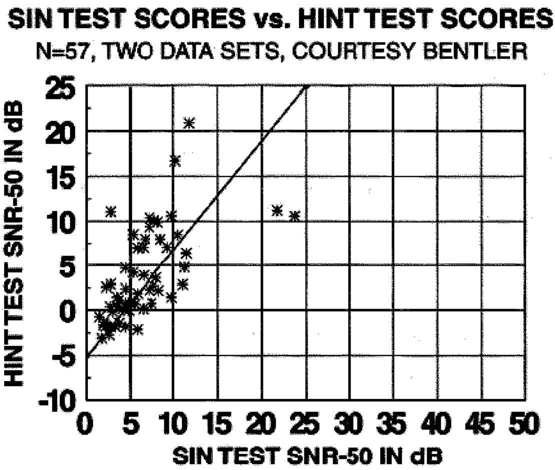

Figure 7 SIN test scores vs. HINT test scores.

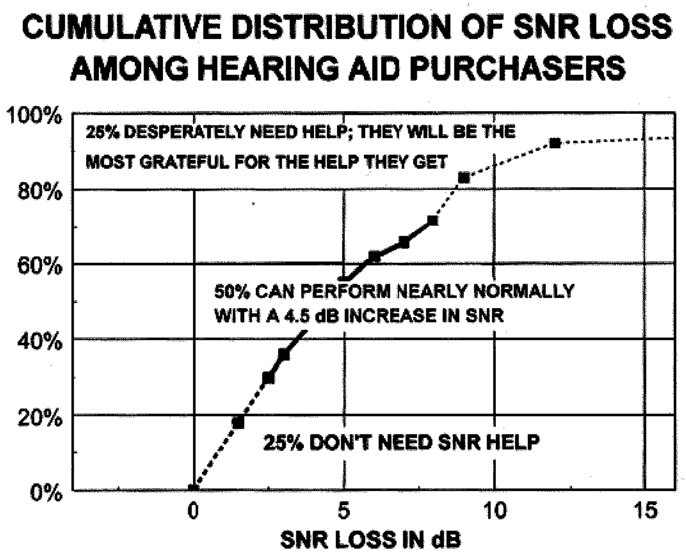

Figure 8 Cumulative distribution of SNR loss among hearing aid purchasers.

B. THE NEED

Since "trouble hearing in noise" is the most common complaint of hearing aid wearers, and "I can't hear quiet sounds" is a relatively infrequent complaint, one might think that we would have abandoned the normal threshold audiogram in favor of a simple SNR-50 test. Abandoning the threshold audiogram is perhaps not so good an idea, since the degree of hearing loss tells us a good deal about how much of the hearing-in-noise complaint may be caused simply by lack of audibility. Moreover, the audiogram is a good indicator of the amount of gain needed to make quiet sounds audible, and the amount of frequency-response shaping appropriate for that loss.

Given the wide range in SNR loss for a given hearing loss, however, the need for a test so simple that everyone can use it becomes clear. Although Lyregaard argued that the simplest test might be the determination of the critical ratio (pure tone threshold above masking-noise spectrum level), that idea has not caught on as of yet. Thus we are developing a "Quick SIN" test as another attempt to supply this need.

Instead of five sentences at each SNR, the Quick SIN uses only one sentence per SNR, and covers the range of 0 to 25 dB in 5-dB steps. The result is a six-sentence block that can be administered in about one minute. The total words correct (including half-word scores) are added and subtracted from 27.5. This is a modified version of the Tillman-Olsen (1973) recommendation for spondee threshold determination. As a quick illustration, if a subject obtained all five words correct at 25 dB SNR and 2.5 words at 20 dB SNR, the SNR-50 would be 20 dB (50% correct at 20 dB). Numerically, the total score would be 7.5, and the calculated SNR-50 would be 27.5-7.5 = 20 dB.

The predicted 95% confidence limit for a single Quick SIN block is +/- 3.1 dB. While this is adequate for many purposes, the average of two blocks (taking two-three minutes) will be within +/-2.2 dB of the correct value 95% of the time. How accurate is 2.2 dB? To put this number in perspective, consider that this is about 9% of the likely range (0-25 dB). consider also that the audiometric 95% certainty is +/-7 dB or 6% of the 110 dB range. (A clinical threshold has a +/- 3.5 dB standard deviation.) From this standpoint 2.2 dB appears reasonable.

V. SUMMARY

The technology exists to give nearly every hearing-impaired individual the ability to carry on conversations in noisy environments such as restaurants and parties. With mild SNR loss, simply increasing audibility with amplification will be enough. With greater SNR loss, some form of directional or close-talking microphones will be required. In all cases, however, we need to know the extent of the SNR loss so we can:

a) choose the appropriate technology (it makes no sense to give everyone with a given hearing loss an FM system just because some individuals with that audiogram need it),
b) adequately counsel our patients about the degree of their difficulty and the problems they will encounter after various levels of intervention have been adopted.

We should be able to provide significantly better benefit and satisfaction to our hearing aid patients in the future. The rapid measurement of SNR loss should help us choose among the increasingly sophisticated circuits and microphones that will be available.

ACKNOWLEDGMENTS
The digital recordings required for the Quick-SIN development were made by Larry Revit. Shilpi Banerjee has been obtaining the normal and hearing-impaired data, on the basis of which the final Quick-SIN recordings will be normalized. Wendy Hanks and Ruth Bentler gave us permission to use their unpublished data in Figures 5 and 6.

REFERENCES

Bentler RA and Duve M (1997). Progression of hearing aid benefit over the 20th century. Poster presented at American Academy of Audiology Convention, Fort Lauderdale, FL.

Bentler RA (1999). Personal communication.

Broadbent DE (1958) Perception and Communication New York: Pergamon Press.

Chung K (1999). Binaural/monaural recordings from experiments with "Real World" experiments (Killion et al 1998).

Davis H, Hudgins CV, Peterson GE and Ross DA (1946). Desirable Frequency Characteristics for Hearing Aids. J Acoust Soc Am 18:247.

Davis H, et al (1947). Hearing Aids: An Experimental Study of Design Objectives Harvard University Press, Cambridge, MA, page 197.

Dirks DD, Morgan DE, and Dubno JR (1982). A procedure for quantifying the effects of noise on speech recognition. J Spe Heari Dis 47:114-123.

Dubno JR, Dirks DD and Morgan DE (1984). Effects of age and mild hearing loss on speech recognition in noise. J Acous Soc Amer 76:87-96.

Fikret-Pasa S (1993). The effect of compression ratio on speech intelligibility and quality. Ph.D. Thesis, Northwestern University. University Microfilms, Ann Arbor, MI.

Hanks WD and Johnson GD (1998). HINT listeners equivalency using older listeners. J Spe Lang Hear Res 41:12, pages 1335-1340.

Harada M (1989). A single port coupler for resonance splitting in a hearing aid. U.S. Patent D300,660.

Haskell G (1999). Special Session presentation entitled The VA-NIDCD Hearing Aid Clinical Trial. International Hearing Aid Conference V, Iowa City, IA.

Killion MC & Villchur E: Kessler Was Right--Partly But SIN Test Shows Some Aids Improve Hearing In Noise. The Hearing Journal, September 1993, Vol. 46, No. 9, pages 31-35.

Killion M (1994). Why some hearing aids don't work well!! The Hearing Review 1:1, pages 40,42,43.

Killion M, et al (1998). Real-world performance of an ITE directional microphone. The Hearing Journal 51:4, pages 24-28.

Lyregaard PE (1982). Frequency selectivity and speech intelligibility in noise.Binaural Effects in Normal and Impaired Hearing (eds, Pedersen OJ and Poulsen T), Scand Audiol Suppl. 15. From the 10th Danavox Symposium, June 8-11, 1982, Denmark.

Nilsson M, Soli SD and Sullivan J (1994). Development of the Hearing In Noise Test for the measurement of speech reception thresholds in quiet and in noise. Journal of the Acoustical Society of America 95:1085-1099.

Pascoe DP (1975). Frequency responses of hearing aids and their effects on the speech perception of hearing-impaired subjects. Annals of Otology, Rhinology and Laryngology Supp. 23, 84:Sept-Oct, No. 5, Part 2 (See especially Table XI).

Pascoe D (1980). Clinical implications of nonverbal methods of hearing aid selection and fitting. Seminars in Hearing 1:217-229.

Plomp R (1978). Auditory handicap of hearing impairment and the limited benefit of hearing aids. Journal of the Acoustical Society of America 63:533-549.

Roberts M and Schulein RA (1997). Measurement and intelligibility optimization of directional microphones for use in hearing aid devices. Preprint 4515 (B-3) from 103rd Meeting of the Audio Engineering Society, New York, NY.

Skinner MW, Karstaedt MM and Miller JD (1982). Amplification bandwidth and speech intelligibility for two listeners with sensorineural hearing loss. Audiology 21:251-268.

Skinner MW and Miller JD (1983). Amplification bandwidth and intelligibility of speech in quiet and noise for listeners with sensorineural hearing loss. Audiology 22:253-279.

Soede W (1990). Improvement of speech intelligibility in noise. PhD thesis, Delft University of Technology, Delft, The Netherlands.

Tillman TW, Carhart R, Olsen WO (1970). Hearing aid efficiency in a competing speech situation. Journal Speech Hearing Research 13:789-811.

Tillman TW and Olsen WO (1973). Speech audiometry. In Jerger J (ed) Modern Development in Audiology 2nd ed. New York: Academic Press, pages 37-74.

Villchur EV (1992). A different approach to the noise problem in hearing aids.Presentation given at the Mayo Audiology Video Conference, October 10.