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THE EFFECTS OF MULTICHANNEL COMPRESSION AMPLIFICATION AND FREQUENCY SHAPING ON SPEECH INTELLIGIBILITY FOR HEARING-IMPAIRED SUBJECTS

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# THE EFFECTS OF MULTICHANNEL COMPRESSION AMPLIFICATION AND FREQUENCY SHAPING ON SPEECH INTELLIGIBILITY FOR HEARING-IMPAIRED SUBJECTS

by

RACHELLE S. ABRAMOVITZ

A dissertation submitted to the Graduate Faculty in Speech and Hearing Sciences in partial fulfillment of the requirements for the degree of Doctor of Philosophy, The City University of New York.

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# CHAPTER I

## INTRODUCTION

Among the major problems that exist for the hearing-impaired individual with a sensorineural hearing loss are difficulty in discriminating among speech sounds, and an intolerance to intense sound because of recruitment. Of interest to those concerned with designing and prescribing hearing aids is the development of instruments that can help overcome these problems.

The majority of early hearing aid research was aimed at attempting to optimize speech discrimination through the use of <u>non-compression</u> hearing aids. A non-compression instrument is one in which a change in the level of the input signal yields an equal change in the system's output. Results of these studies varied from recommending <u>fixed frequency-gain characteristics</u>, which is amplification utilizing uniform frequency characteristics for most cases of hearing loss, to <u>individualized selective amplification</u>, which is amplification varying in amount at different frequencies, and fitted to the individual.

Early findings with non-compression hearing aids did not optimally resolve the discrimination problems of the hearing impaired. In addition, the method used to resolve the tolerance problem was that of <u>peak clipping</u>, which raised an additional problem. In peak clipping, when the amplified signals exceed a predetermined maximum output level, their amplitude is reduced at the peaks. This mode of amplitude limiting introduces distortion, which may have the undesirable effect of reducing speech discrimination.

Much research therefore turned to the use of <u>compression</u> amplification to resolve the twofold problem of reduced discrimination and intolerance to intense sound. In compression amplification the gain of the aid is automatically reduced when the input exceeds a predetermined level, known as the compression threshold. Compression is theoretically advantageous because low inputs are amplified more than high inputs, protecting the user who has a reduced Loudness Discomfort Level, while avoiding the distortion that results from peak clipping. Studies in this area have resulted in much conflicting data as to whether or not compression amplification improves discrimination over non-compression amplification. This led to the use of compression in a multi-channel form. In this form, the frequency range is split, with each channel or band being processed separately. This was in an attempt to compensate for the differing characteristics of individuals' hearing loss at various frequencies. However, the advantage of multi-channel compression amplification has still not been clearly established.

The present study was designed as follows: Non-compression amplification, with and without peak clipping, was compared to both single- and two-channel compression amplification. Conditions included one fixed frequency-gain response, that of 0 dB/octave slope, and one response that was selectively amplified for each subject. This latter frequency response involved amplification of each one-third octave band of speech to just below each subject's Loudness Discomfort Level (LDL) for that band, and was known as the "LDL frequency response."

The data were analyzed in an attempt to answer the following two basic questions:

1. What is the optimum frequency-selective amplification for sensorineurally impaired listeners with characteristics such as those chosen for the present study?

2. Does frequency shaping that places as much of the speech spectrum as possible into the residual hearing area optimize speech discrimination?

Analysis of the data involved its comparison in many ways, including peak clipping v. no peak clipping in a non-compression system; non-compression v. compression; single-channel compression v. dual-channel compression; and the effect of individualized frequency shaping on each of these modes of amplification.

# CHAPTER II

# REVIEW OF RELATED LITERATURE

The area of hearing aid research has been the subject of intensive investigation for many years. The first section of this chapter contains a discussion of the nature of the auditory problems encountered by sensorineural hearing-impaired persons, and the resultant theoretical background of hearing aid selection strategy. It is followed by a section on practical approaches to hearing aid design, including both early and recent studies of interest to the present study. Included in this section is a discussion of both non-compression and compression amplification systems. A summary of these approaches is given in the final section, as background to the present study.

## Hearing Aid Design

Despite all the advances that have been made in recent years in the design of hearing aids, many users are still not satisfied with their aids. Tillman, et al. (1970) found that the very fact of wearing a hearing aid created problems that added to the problems inherent to the loss itself. On the most basic level, the purpose of the hearing aid is to amplify sound for the person with a hearing loss. However, unless it also makes speech more intelligible, or, at least, does not degrade the listener's unaided speech intelligibility, its usefulness may be questionable. Achieving good, or improved, speech intelligibility has been one of the major goals in hearing aid research. In order to better understand this issue, let us first examine some of the problems the hearing-impaired individual faces.

Persons with a conductive hearing loss can often be treated medically, and, in any case, face far fewer of the perceptual problems than those with a sensorineural loss. We will therefore concern ourselves with those problems encountered by persons with a primarily sensorineural loss. Among some of the major problems that may be present for those with the latter type of loss are the following:

1. An inability to hear many elements of the speech signal because of a loss in sensitivity.

2. A reduction in discomfort level, leading to a reduction in auditory area between threshold and discomfort, without any increase in ability to extract information from speech signals within this area. An effect of this problem may be that the listener's speech intelligibility decreases as intensity increases beyond the level at which maximum intelligibility is achieved.

3. Degradation of signals by such factors as spread of masking in frequency, in which low frequency sounds, whether they are speech or background noise,

mask the high frequency components of speech. This problem may be the cause of one of the most common complaints of the hearing aid user, which is difficulty in comprehending speech in the presence of noise or competing speech. This abnormal vulnerability to acoustical interference may affect the listener to the extent that he often cannot understand speech, even though all the signals may be amplified, and competing noise is so slight that the person with normal hearing might ignore the noise very easily.

4. Further degradation of signals by an abnormally faster rate of growth of loudness, or recruitment. There is increasing evidence in recent research (e.g., Villchur, 1974) that, whether or not there are other perceptive aberrations present, recruitment can sufficiently impair speech perception to cause a loss of intelligibility. Because of this evidence, the problem of recruitment will be covered in greater detail below.

As a result of the problems mentioned, it is not uncommon to judge a hearing aid to be suitable for a patient during clinical evaluation, but subsequently to be unsatisfactory for everyday use. Hearing aid research has addressed this apparent problem.

As discussed by Braida, et al. (1976), a common goal in hearing aid design is the selection of signal processing units that match acoustical signals to residual auditory function. Since speech is the most important single class of such signals, it is of primary importance that such matching results in good, or improved, speech perception for the listener. The signal processing unit can function in such a way that lost cues are restored, or that new cues are introduced to the listener. In the latter case, these new cues must be able to serve similar functions as the old ones, and it must be possible for the listener to be trained to use them. In addition, the signal processor must not degrade those cues that have not been adversely affected by the hearing loss, such as the rhythm of speech.

Because of the wide variation of incoming signals in the environment, it is extremely difficult to determine the optimum frequency-gain characteristic for the hearing-impaired person. These variations show up as sex of talker, clarity of speaker, speech material, distance and position of speaker from the listener, reverberation, etc. (Braida, et al., 1976). In addition, rules that may be devised for hearing aid fittings change for different degrees of loss, and especially for different degrees of high-frequency loss.

Following is a summary of how the special design of hearing aids attempts to alleviate some of the effects of hearing loss:

1. Loss in sensitivity can generally be compensated for by appropriate choice of gain. However, the optimum frequency-selective amplification is yet unresolved.

2. The frequency-gain characteristic chosen for an individual should amplify that part of the frequency

spectrum that contributes to his speech perception, but should do so without exceeding the loudness discomfort level for the individual, and without causing parts of the amplified spectrum to mask other parts. The amount of amplification must, of necessity, be limited, because of the threshold of discomfort. Therefore, hearing aids should have a mechanism that limits the intensity of the amplified signals. Peak clipping and compression amplification are two means whereby intensity may be limited. Both these solutions are discussed more fully below.

3. Intelligibility in noise is a special problem for the hearing aid wearer. Because everyday communication entails listening to speech in the presence of other competing signals, tests involving background noise should be an important consideration for accurate assessment of a hearing aid's value in everyday use (Carhart, 1946). In addition, testing in background noise has been found to be more effective in discriminating among hearing aids in the clinical evaluation.

Attempts to solve the problem of the listener's abnormal vulnerability to acoustical interference have not been fully successful. One means, though, of attempting to solve the problem is by appropriate microphone placement, so as to maximize the signal-to-noise ratio (Lippmann, et al., 1976). Other solutions may lie in the appropriate choice of parameters, such as compression threshold in compression hearing aids. The parameters of compression are described in detail later in this review.

4. One reason why amplification may not be fully successful, even when carefully chosen with respect to overall gain and frequency-selectivity, is that it cannot restore the normal relationships between intensity and loudness in the recruting listener's auditory area. Because recruitment has been shown to have such a deleterious effect for the listener's intelligibility (Villchur, 1974), and since a search for a solution to this problem was one of the focal points of the present study, the following sections are devoted to this problem.

Loudness and Recruitment

Recruitment is a disproportionate increase in loudness as a function of intensity. It is one perceptive aberration affecting the hard-of-hearing person with a sensorineural hearing loss of cochlear origin. The relationship between the intensity of a sound and its perceived loudness is known as the loudness function (Huizing, 1952). The steepness of the loudness function depends, in the normal listener, on the frequency of the stimulus, with the curve being steepest in the low frequencies, less steep in the high frequencies, and even less in the middle range (Fletcher and Munson, 1933). For those with a sensorineural hearing loss of cochlear origin, as the intensity of sound increases beyond the elevated threshold of hearing, the loudness of the stimulus increases towards normal. As the amount of hearing loss increases, the corresponding recruitment curve becomes progressively steeper (Hallpike and Hood, 1959).

Various methods have been developed for the measurement of the phenomenon of recruitment. These methods consist of procedures using subjective listener judgments, such as loudness balance measures, and objective procedures, such as impedance measures. The Alternate Binaural Loudness Balance (ABLB) test and the Monaural Loudness Balance (MLB) test are included in the former category, as is the Loudness Discomfort Level (LDL) test. Calculation of the listener's dynamic range, that is, the distance from threshold of audibility to LDL, or his distance from Most Comfortable Loudness (MCL) to LDL, is yet another method of measuring recruitment. Although measurement of the Acoustic Reflex Threshold (ART) has been used by some as an objective means of determining recruitment (e.g., Metz, 1952; McCandless, 1973), this procedure is controversial (e.g., Ritter, et al., 1979) and non-standard.

Experimental Signal Processing to Overcome the Effects of Recruitment

Experimental signal processing has been utilized to exemplify what happens to the loudness relationships of the elements of speech for the recruiting listener. Recruitment, which exaggerates loudness differences among the acoustical elements of speech, was simulated in normal hearing listeners by intensity expansion (Villchur, 1974).

As defined by Villchur (1978), in an amplitude expander, gain increases with input level. The accelerated loudness growth of recruitment can be simulated by this increasing gain. Villchur (1974) achieved simulation of recruitment by using an electronic processor, enabling him to transpose the distorted loudness relationships from the hard-of-hearing subject's range of hearing to the normal range. The same proportionate relationship between speech levels and corresponding hearing ranges was kept, so that those speech elements falling below the hard-of-hearing subject's threshold also fell below the normal listener's threshold.

Villchur's proportionate auditory area fitting is illustrated graphically in Figure 1. The first part of this figure represents values relevant to the hearing-impaired subject, and the second part represents values relevant to the normal hearing subject. In both parts of the figure, the lower solid curve represents threshold, and the upper solid curve represents a 74-phon equal loudness contour. The latter was chosen by Villchur because the peak speech level of conversational speech at 1000 Hz is 74 dB SPL. Speech falling between these two curves is audible to the listener at comfortable levels. Note how much smaller the range is for the hearing-impaired subject than for the normal hearing subject. The dynamic range of the levels of speech covers a range of roughly 30 dB (Dunn and White, 1940). The two dashed curves in each part of the figure show this range. For the hearing-impaired subject, speech

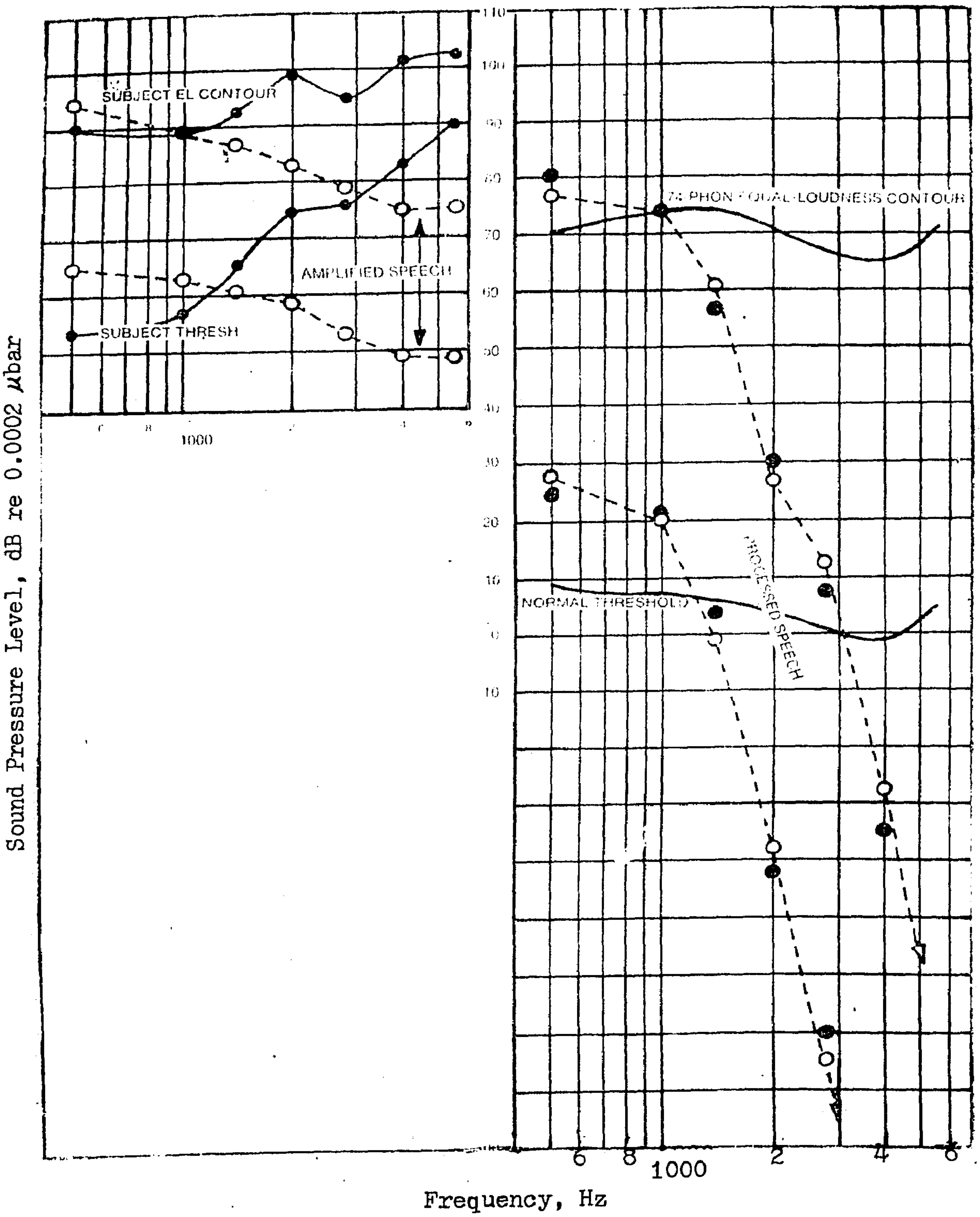


Fig. 1. Projection of the amplified speech band from the deaf-subject span of hearing to the normal span, keeping the same proportionate relationship between speech levels and correponding hearing spans. Solid dots represent the processing actually achieved in Expt. 1 (from Villchur, 1974).

has been amplified so that as much of the speech elements as possible lie within the available auditory range. Note that when the high frequency components of speech are amplified, they still remain below the subject's threshold hearing, and yet the low frequency components are above the subject's equal loudness contour. In the second part of Figure 1, the amplified speech band is projected from the hearing-impaired span of hearing to normal span. The same proportionate relationship between speech levels and available comfortable auditory range is kept, so that, again, the most intense high frequency components still fall below the threshold of hearing, whereas the most intense low frequency components are above the equal loudness contour shown.

To judge the validity of this signal processing, four subjects, each with a unilateral sensorineural hearing loss accompanied by recruitment, were asked to judge the similarity or dissimilarity of speech samples heard in each ear. Unprocessed speech was presented to the impaired ear, while the normal ear received both unprocessed and processed speech, the latter calculated for the subject by the experimenter. The speech was then more finely adjusted by the subject himself to make the sound in his two ears as alike as possible.

Unprocessed speech in the normal ear was always rated as different, in varying degrees, from unprocessed speech in the impaired ear. The final processed speech in the normal ear was always rated as "similar" or "very similar" to unprocessed speech in the impaired ear.

The basis for Villchur's (1974) simulation of recruitment was thus twofold. First, many elements of speech, more often the high frequency elements, fall below the threshold of hearing, even when they are amplified. And second, because of recruitment, loudness relationships among the acoustical elements of speech are distorted. This occurs because recruitment expands the difference in loudness between low frequency high-amplitude vowels, and high frequency low-amplitude consonants. Perceptually, the effect is as if the high frequencies were attenuated.

The fact that intelligibility for normal hearing listeners could be reduced by simulating recruitment implies that, whether or not there are other causes, recruitment in hard of hearing listeners is a sufficient cause for loss of intelligibility (Villchur, 1974). Others have established a relationship between loss of speech discrimination and degree of loudness recruitment present--as the amount of recruitment increases, maximum speech intelligibility has been found to decrease (Hood and Poole, 1971). However, the relationship between these two factors is not necessarily causal, since both could be dependent on a third factor, such as amount of hearing loss.

If Villchur's simulation of recruitment by intensity expansion is valid, amplification and processing of speech will restore even high frequency elements to be within the dynamic range of the listener, counteracting expansion and treble attenuation of recruitment. Villchur reasoned that discrimination for the hearing impaired could thus be improved by making the weaker high frequency acoustical speech elments audible, without making the high amplitude elements uncomfortably loud.

The following section deals with a rationale representing some of the approaches to hearing aid design, including that based on Villchur's simulation of recruitment.

## Approaches to Hearing Aid Design

Hearing aids may be divided into two basic types: non-compression and compression modes of amplification. Considerable research was, and still is, conducted with the former type of system, in an effort to best "fit" this type of amplification to the hearing-impaired individual. Even though complete resolution of the best type of non-compression amplification has not been achieved, new controversies have arisen attempting to relate the best compression system to the hearing-impaired individual. The review of related literature will focus on both the assets and the drawbacks of non-compression hearing aids, with particular attention to the current state of the art regarding compression amplification for the hearing impaired.

### Non-Compression Amplification Systems

In a conventional amplifier, a change in the level of the input signal yields an equal change in the system's output. Within the dynamic range of the amplifier, the ratio of change in input to change in output is a constant. A limit to the maximum power output is built into the hearing aid, that is, a limit of the intensity to which the hearing aid can amplify. When this limit is surpassed, the result is an overload of the system, which results in the peaks of the sound waves being clipped off. This peak clipping introduces non-linear distortion. The effect of peak clipping for the hearing impaired is discussed later in this review.

Hearing aid selection strategies with non-compression hearing aids can be accomplished with several different methods. Three of these methods have been selective amplification, fixed frequency-gain characteristics, and adaptive fitting of a master hearing aid.

Individualized Selective Amplification

There is a reduced and distorted dynamic range between threshold of hearing and threshold of discomfort for the individual with recruitment. A recognized goal in hearing aid fitting has traditionally been to try to bring the sounds of speech into this dynamic range. Individualized selective amplification, which is amplification varying in amount at different frequencies, is one method used to bring all of speech sounds within the listener's small and distorted auditory area. Individualized selective amplification focuses on determining the optimum frequency-gain characteristics on an individual basis. These methods typically require psychoacoustic data such as thresholds of hearing, thresholds of discomfort, and loudness matching (Levitt, 1978b).

Watson and Knudsen (1940) were among the first to attempt to deal quantitatively with the problem of reduced dynamic range for the hard of hearing. This led them to propose a method of selective amplification based on equal loudness contours. Using several frequency-gain variations, they measured discrimination for syllables in 16 impaired ears, of which 7 had sensorineural losses. Their best results were in the case of frequency-gain characteristics derived from the most comfortable equal loudness contour for air conduction, in which the difference (in decibels) between the most comfortable equal loudness curve and the normal auditory threshold was mirrored. This type of amplification proved superior to uniform amplification when first tried, or, in some subjects, after an accommodation effect took place.

While this procedure may have yielded satisfactory results, others, such as Davis, et al. (1947), discounted Watson and Knudsen's results. They did so on the grounds that it was impractical for clinical use with adults, and even more so with children, because of the difficulties involved in obtaining reliable equal loudness contours with untrained subjects. In addition, Davis, et al. concluded that the gain in intelligibility over other fixed frequency responses (discussed later in this review) was minimal.

Another conclusion of Watson and Knudsen's study was the importance of not relying on subjective judgments as a means of choosing amplification. Their reasoning for this was twofold. First, listeners often prefer amplification that stresses those frequencies that they hear best without any amplification, but does not necessarily result in the best improvement in intelligibility. And second, exposure time and training may play an important role in objective test results. Selective amplification will render much of the speech spectrum within the audible range, but, since the various speech sounds are not what the hearing-impaired listener is accustomed to hearing, they may no longer sound "natural" to him. The listener may, therefore, initially not choose this "unnatural" sound, but, after a habituation effect, it may prove to be the best choice for a hearing aid.

Huizing and Reyntjes (1952) have suggested that amplification should emphasize frequency regions where recruitment is less, and suppress regions where recruitment is high, in order not to deteriorate overall intelligibility. This concept departs from the idea of high frequency emphasis, where amplification is greatest for those frequencies where the listener has the poorest thresholds. In a follow-up study, Huizing, et al. (1960) measured the articulation scores for discrete frequency bands by using filtered speech. They concluded that ideal selective amplification would continue to stimulate the entire frequency range of the sensory mechanism, with the concentration on amplifying those frequencies where discrimination is best.

Continuing research with selective amplification, Reddell and Calvert (1966) tested 24 subjects with high frequency hearing losses and no experience with hearing aids. Subjects' responses with an experimental aid that had been custom-adjusted for them by an audiogram-mirroring technique, were compared to responses with two commercially-available hearing aids. The latter were selected by customary audiologic procedures (they did not specify what these procedures were). With the experimental aid, subjects scored slightly better for mean Speech Reception Threshold (SRT), discrimination score, Loudness Discomfort Level (LDL), and their own subjective evaluation of the hearing aid.

The concept of providing gain to mirror the audiogram, that is, amplification where the response curve of the hearing aid compensates exactly for the subject's degree of hearing loss, is thus another means of individualized selective amplification. It has, however, been reported in the literature with conflicting findings as to its value. Audiogram-mirroring may result in a fairly high loudness level at a relatively low sensation level. This may happen since at supra-threshold levels, where listening takes place, the equal loudness contours tend to normalize. As a result, this method has traditionally worked well only for the special case of a primarily conductive impairment, since there is no recruitment present, or for a moderately flat sensorineural loss (as discussed below). Consequently, other means may have to be used to successfully amplify speech for the recruiting listener.

More recently, however, Pascoe (1975) has found that an audiogram-mirroring technique, less a constant gain across all frequencies, was beneficial for eight presbycusics with mild to moderate sensorineural impairments. The shape of their losses was either flat or sloping moderately downward. However, it only worked well provided the real ear response of the hearing aid from sound field to eardrum was used. Pascoe set the gain of the hearing aid equal to the amount of hearing loss at each frequency, less a constant amount. The result was a mirroring of the shape of the hearing loss, as depicted by the audiogram, but with a gain less than the hearing loss itself. His results showed improved discrimination under this condition of amplification, as compared to several other conventional hearing aid fitting procedures, detailed below.

Because some of the methodology of Pascoe's study had a strong influence on the present study, it will be reviewed in greater detail. As compared to previous studies, Pascoe made several changes and/or additions in his research. For example, both male and female speakers were used, and testing was conducted not only in quiet, but also in noisy (signal-to-noise ratio of 6 dB) and reverberant environments. These test conditions were used in order to more effectively evaluate the differences among hearing aids. In addition, the functional gain[1] of the hearing aids tested was specified, as opposed to the traditional method of specifying the gain as measured in a 2cc coupler. The overall gain from sound field to eardrum is the desired ideal. However, since this is a difficult measure to obtain, the functional gain, as defined operationally by Pascoe, gives one the ability to get an approximation of the over-all gain of the hearing aid from sound field to eardrum.

Pascoe obtained intelligibility scores for several frequency-gain characteristics, including: uniform functional gain (UFG), flat gain for one-third octave bands of noise, as measured at the ear; uniform hearing level (UHL), which mirrored the free field audiogram, in terms of functional gain, less a constant; uniform coupler gain (UCG), flat gain as measured with a 2cc coupler; 6 dB rise per octave (RC6), as measured with a 2cc coupler; and commercial aid simulation (AS).

Intelligibility in the sound field in Pascoe's study was measured using a specially designed 50-item word list. The list contained familiar words with many high frequency voiceless consonants. These are the phonemes that tend to be difficult for hard of hearing listeners, but help to

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[1]Functional gain may be defined as the aided versus unaided thresholds for stimuli such as third-octave bands of noise, as measured in a sound field, yielding an estimate of the real ear gain (gain measured at the ear).

differentiate among hearing aids. Gain was always adjusted to Most Comfortable Level (MCL). Averaged over listeners and speakers, in both quiet and in noise, scores were as follows: UHL: 73%; RC%: 66%; UFG: 62%; UCG: 56%; and AS: 54%.

These scores demonstrate the importance of mirroring the audiogram for the type of subjects Pascoe tested. Mirroring of the audiogram was a concept that had been previously rejected by some earlier researchers. Nonetheless, when it was utilized in terms of functional gain, Pascoe found that it yielded the highest scores under his experimental conditions. In addition, the differences in scores that he obtained between UFG and UCG had implications for the conclusions reached in many of the earlier studies regarding frequency-gain characteristics of selected hearing aids. That is, since a hearing aid changes the free field gain of the external ear, the functional gain of a hearing aid more accurately represents the true frequency response from sound field to eardrum.

It should be noted that slope of loss may strongly influence the results of individual selective amplification. For Pascoe's subjects, an audiogram-mirroring technique provided the best speech discrimination scores, his subjects having either flat or moderately sloping audiograms. Skinner (1976) compared several frequency response characteristics in hearing aids similar to Pascoe's but tested subjects with noise-induced hearing losses showing severe high frequency drops in threshold. For these subjects, audiogram-mirroring did not produce good results. Rather, the best speech discrimination scores resulted with a frequency response in which there was no gain below 500 Hz, and 23 dB of gain at and above 1600 Hz. Between these two frequencies, the slope of the functional gain (defined as above, by Pascoe, and also measured with one-third octave bands of noise) reflected the difference between each subject's threshold curve and the normal threshold curve.

Another approach to individual selective amplification was developed by Barfod (1972). He tested the hypothesis that frequency selective amplification emphasizing the high frequencies would improve the subject's speech intelligibility. He tested 10 subjects with "degeneration nervi acustici," with normal hearing in the low frequencies, and a sharp drop in the high frequencies. The test stimuli were (C)CVC(C) words, which were mainly nonsense syllables. All were recorded by a male speaker and embedded within carrier phrases. The tests were administered monaurally under headphones, at a speech input level of 60 dB SPL, with a signal-to-noise ratio of 5 dB. The noise was filtered noise with a spectrum corresponding to that of average male Danish speech.

For subjects with the etiology and audiogram that Barfod selected, the best intelligibility was obtained with a response that transformed the 10% time levels of speech (i.e., those levels of speech exceeded 10% of the time) for normal hearing persons. Although his results may seem to contrast with those of Pascoe (1975), whose best fit was based on audiogram-mirroring less a constant amount, Barfod's subjects had a different shape to their hearing losses than did Pascoe's subjects. It is therefore important to investigate if Barfod's results extend to the hearing impaired with other configurations of losses.

Another variation of frequency selective amplification is that of the bisection method. In this method, the region between the auditory threshold and the threshold of discomfort is bisected. This is achieved by finding the point that lies midway between these two thresholds at each frequency, and then joining these points to form a line of bisection. Wallenfels (1967) suggested a variation of the bisection method, as follows: The line of bisection was calculated between 1000 Hz and 4000 Hz. Above 4000 Hz, the line of bisection was followed only if the listener had an ample dynamic range of about 50 dB or more. If the dynamic range was considerably narrowed, then the recommended curve remained below the line of bisection, closer to threshold of audibility. The recommended hearing level curve below 1000 Hz was dependent on the slope of the line of bisection between 1000 Hz and 4000 Hz. In the case of a steep slope in this latter region, that is, 8-10 dB/octave or greater, the hearing level curve below 1000 Hz continued downwards with the same slope. But if the slope between 1000 Hz and

4000 Hz was less than 8 dB/octave, a downward slope of 8-10 dB/octave was used for the region below 1000 Hz.

Wallenfels' (1967) method took into account the possibility of upward-spread-of-masking, in which the low frequency elements of speech, when amplified, might mask the high frequency elements. This might prevent the listener from receiving important information-bearing cues to speech discrimination. His method corrected for this problem by varying the amount of amplification below 1000 Hz, depending on the severity of hearing loss in the high frequencies. Nonetheless, although the upward-spread-of-masking concept should be considered in hearing aid fitting, Wallenfels' method has not been compared experimentally to other methods of selective amplification.

Another variation of the bisection method has been suggested by Victoreen (1973). He suggested that the amount of gain at each frequency be chosen so that the speech spectrum for the hard-of-hearing listener lies at an intensity between one-third and one-half the distance from threshold of audibility to threshold of discomfort. With this method of selective amplification the speech spectrum has about the same relation to these two thresholds as does (unamplified) speech presented to the normal listener at a comfortable level.

In summary, several methods of frequency shaping have been proposed, such as audiogram-mirroring, bisection, variations of bisection, flat frequency response, and equal loudness methods. Of these various methods, reconstructing the equal-loudness contours appears to be the most promising. But, as Pascoe (1975) has shown, the entire issue is up for review, because the early research did not take the functional gain at the eardrum into account.

Fixed Frequency-Gain Characteristics

A divergence from fitting to the audiogram and the principle of frequency selective amplification, is that of uniform frequency characteristics. This variation has as its goal systematic determination of the optimum frequency-gain characteristic of hearing aids that would provide good performance for most, if not all, cases of hearing loss.

In an early review of research on selective amplification, Watson and Tolan (1949) concluded that approximately 75% of hearing aid users achieved highest scores for PB-50 lists presented in quiet with a flat frequency-gain characteristic, as measured in a standard 2cc coupler. However, under unfavorable conditions, such as non-linear distortion, clipping, or noise, a three to nine dB per octave high frequency emphasis was preferred, especially for listeners with sharply falling audiograms. A supra-threshold equal loudness contour was a more reliable indicator of the amount of high frequency emphasis required than the threshold curve. Some of the studies that were considered in their research are detailed below.

Two of the earliest investigations on fixed frequency-gain characteristics, conducted on a large scale, in the 1940's, were done at the Harvard Psychoacoustic Laboratory (Davis, et al., 1947), and by the Medical Research Council of Britain (1944).

On tests made on 25 ears (18 subjects) with losses of 40 to 90 dB in the speech frequencies, and an average loss of 65 dB, Davis, et al. (1947) compared a variety of frequency-gain characteristics. They were as follows: attenuation of 12 and 6 dB per octave, flat gain, and high frequency emphasis of 6 and 12 dB per octave. PB-50 word lists were administered over a range of 40 to 70 dB. Maximum power output was 124 dB SPL, and frequency range extended from 100 to 7000 Hz. Their conclusion was that a flat or slightly rising characteristic of up to 6 dB per octave over the frequency range of 300 to 4000 Hz, as measured in a 2cc coupler, was the best hearing aid selection for most hearing-impaired listeners.

Several other procedures were evaluated and rejected. These included reliance on the subject's own judgment, a procedure rejected by Watson and Knudsen (1940), too, and selective amplification by mirroring of the audiogram, a still controversial procedure. Amplification restoring the subject's most comfortable equal loudness contour to normal was also rejected. In pilot trials, that pattern received no better results than the rising contour of 6 dB per octave, and the necessary data were significantly more difficult to obtain.

The conclusions reached in the Harvard report (Davis, et al., 1947) were that "...for the usual hard of hearing patient, any detailed 'fitting' is wasteful of time and effort. The differentials between instruments that are indicated by most current tests are largely illusory." According to Davis, et al., only difficult and unusual cases of hearing loss require that they be given special testing. For these latter cases, there may be significant differences based on small changes, and then, more elaborate selective tests may be called for.

Generalizations from the conclusions of Davis, et al. may be limited for the following reasons:

Only six of the twenty-five ears tested had losses that were purely sensorineural in origin. Because of major medical advances since the 1940's, medical and surgical intervention has significantly reduced the number of people with conductive hearing loss. As a result, the proportion of hearing aid users whose hearing impairment can be attributed to a conductive involvement has been reduced. Thus, using conductives as a major part of the study's population, and generalizing to today's population of hearing aid users, is not necessarily valid.

Phonetically balanced monosyllables were used as the test stimuli, although they are not always sufficiently sensitive to detect differences in performance among various hearing aid settings. This was confirmed in later years by Shore, Bilger and Hirsh (1960), who found that this type of test material was not very sensitive in differentiating among hearing aids.

In addition, tests were administered only under conditions of quiet, and differences among hearing aids frequently emerge only under more difficult listening conditions, such as in noise. Tests also employed only male speakers, another condition that may not be sufficiently sensitive in differentiating among hearing aids. It may be that more difficult situations, such as female and children speakers, also help differentiate among hearing aids.

Lastly, because the electrical signal was amplified and then delivered via a supra-aural earphone, the effects of head diffraction and body baffle, which are encountered by the wearer of a conventional hearing aid, were circumvented. The validity of simply applying a standard correction to the frequency response characteristic, to compensate for the absence of these effects, is questionable, since they depend on factors such as room reverberation, location of the microphone, and azimuth of the source (Resnick, 1977).

A study conducted by the Medical Research Council of Britain, in 1944, had, as one of its goals, a task similar to that of the Harvard study (Davis, et al., 1947). They, too, attempted to determine the frequency response characteristic of a hearing aid that would yield the best performance for as large a percentage of hearing-impaired people as possible.

Two hundred and twenty eight hearing-impaired listeners, aged 10 to 70 years, were tested with 50 monosyllabic words, half recorded by a male speaker and half by a female speaker. Sixty-five percent of the hearing impairments of the population were conductive, and all losses ranged in severity from slight to profound, averaged over the frequencies 500 to 4000 Hz. One-third of the patients were hearing aid users. Signal levels extended over a 28 decibel range, corresponding to levels encountered in soft speech, in ordinary conversation, and in loud conversation, in order to represent various grades of social disability.

The testing protocol was unclear, but the Council's general conclusions were as follows: Similar to the Harvard report (Davis, et al., 1947), either a flat gain, or a rising gain at the rate of 5 to 6 dB/octave, and an upper frequency cut-off of 4000 Hz, yielded the best intelligibility for their group as a whole. The suggested low frequency cut-off was a frequency response decrease of 12 dB per octave below 750 Hz. The former specification is similar to that of the Harvard study, whereas the latter differs slightly, since a low frequency cut-off no higher than 500 Hz was suggested in the Harvard study. However, although changing the low frequency cut-off from 100 Hz to 500 Hz did not significantly degrade intelligibility, the best response was obtained with a 200-400 Hz low frequency cut-off (Davis, et al., 1947). There are additional differences between the two studies that relate to the method of calibration, discussed below.

In the MedResCo study, as in the Harvard report, generalization from the conclusions is limited, for the following reasons:

First, sixty-five percent of the population had conductive hearing losses, making the profile of their typical hearing aid user very different from the profile of today's "typical" hearing aid user. Second, experimental procedures were not clearly defined. In addition, there was significant intrasubject variability for some subjects. There was a 20% difference between repeated scores for a given hearing aid setting. Although test-retest reliability is one of the most important factors in hearing aid evaluations, few published studies have effectively dealt with this issue. And last, Pascoe (1975) has suggested that word lists weighted with high frequency voiceless consonants are more effective in differentiating among hearing aid response patterns, than are word lists such as the W-22's, used in the MedResCo study.

Although the Harvard report and the MedResCo study appear to recommend similar frequency-gain characteristics, there were important differences in the method used by each study to specify these characteristics. Taking these differences into account allows one to see that the recommended specifications actually differ significantly. In the Harvard study, the method of measurement eliminated body baffle, head diffraction, and concha and ear canal effects, since the frequency response of the aid was measured by placing the microphone in a free field facing the sound source, and measuring the output of the earphone in a 6cc coupler. On the other hand, although the microphone was similarly placed in a free field in the MedResCo study, a correction was then applied to the free field input. The correction was the amount that an average head and external ear would have increased the input pressure, resulting in the sound pressure that would have existed at the eardrum, had the head been placed in the microphone location. The receiver output was then measured in a 3cc coupler. The larger head-diffraction effect and the smaller earphone-coupler combination effect thus caused differences between the hearing aid's frequency-gain specification that are, superficially, not apparent.

If the correction for head and ear effects made in the MedResCo study is eliminated, significant differences emerge between the recommended frequency responses of the two studies. Where the Harvard study recommended a 0 dB/octave slope, the slope recommedned in the MedResCo study that is often considered equivalent was actually 12 dB/octave up to 750 Hz, and then 5 dB/octave. And where the Harvard study recommended a moderate upward slope of about 6 dB/octave, the MedResCo response is a relatively steep upward slope of 10 or 12 dB/octave (Resnick, 1977).

Knight (1965), using an insert phone, attempted to verify the conclusions of the two previous studies. In the Harvard (Davis, et al., 1947) and MedResCo (1944) studies, a wide range of hearing-impaired subjects were used.

In his study, Knight selected three types of subjects: those with purely conductive hearing loss, those with purely sensorineural hearing loss, and those with mixed hearing loss. Articulation scores were obtained at MCL for speech, and at MCL plus and minus 5 dB, for a total of three listening levels. The three frequency responses used, as measured with a 2cc coupler, were as follows: rising linearly with frequency at 7 dB/octave (similar to the ideal frequency response found by the Harvard study); same as the first response to 1000 Hz, and then increased by an additional 6 dB/octave; and same as the first response to 1000 Hz, and decreased by an additional 6 dB/octave.

Of the three response curves tested, the one judged to be best, based on articulation scores, was the first response, that is, rising by 7 dB/octave. This response was the one closest to that recommended by the Harvard study.

Adaptive Fitting of a Master Hearing Aid

Considerable research has been conducted with fixed frequency-gain characteristics, to establish a protocol as an acceptable tool for hearing aid fittings. However, recent studies have shown that individualized fitting yields improved intelligibility for the hearing impaired over standardized hearing aid characteristics.

One recent study, a variation of the individualized fitting, was the Wearable Master Hearing Aid (WMHA) project (Resnick, Levitt and White, 1977). Its purpose was to develop a practical clinical protocol to be used for the prescriptive fitting of a WMHA. The single hearing aid offered a range of hearing aid characteristics which could be easily and rapidly manipulated. The ultimate goal for its clinical use would enable the audiologist to select the amplification characteristics most suitable for the patient, without the need to maintain a large supply of commercially available hearing aids in stock. A hearing aid with the same electroacoustic characteristics as the prescribed WMHA would then be selected.

Various stages of testing ultimately led to a stage designed to closely approximate the optimum characteristics of the WMHA. This was achieved through successive adjustment of the hearing aid's parameters, according to a prescribed adaptive procedure. This strategy aimed to move continuously closer to those settings that produced high scores. When the best hearing aid setting was achieved, performance with the WMHA was compared to the subject's own clinically-selected hearing aid.

The WMHA study showed that systematic adjustment of the parameters of the Wearable Master Hearing Aid, to best suit the needs of each patient, yielded improved performance. There was a significant difference among subjects of the estimated optimum settings obtained in this way. This finding suggests that methods for fitting hearing aids should take individual differences into account. Another conclusion reached was that use of a common fixed frequency-gain characteristic to help all, or at least the majority of, hearing aid wearers, will not typically produce the best results. For example, results showed a consistent pattern of improvement for the estimated optimum setting of the WMHA, as compared to performance with the subject's own hearing aid. Larger improvements were obtained, on the average, for subjects with initially lower scores. Performance was also compared at the estimated optimum of the WMHA to the performance on the WMHA using the frequency-gain characteristics as recommended by the Harvard study (Davis, et al., 1947). These responses were a flat frequency response condition, and an upward sloping frequency response of 6 dB per octave, as measured in a standard 2cc coupler. Again, consistent improvements were obtained for the WMHA at its estimated optimum setting, the larger improvements being obtained, on the average, for subjects with initially lower scores.

Limitations of Non-Compression Amplification Systems

In sensorineural hearing loss, the dynamic range of the auditory system is usually considerably narrowed. The lower intensity limit shifts by an amount equal to the threshold loss, while the upper limit remains the same, or is reduced. This results in a discrepancy of the loudness values of speech, leading to poor intelligibility. Villchur (1974) has argued that this phenomenon must be taken into account in hearing aid selection strategy. In principle, a non-compression amplifier can be used to reconstruct only one equal loudness contour. This has been shown to yield good results (e.g., Barfod, 1972; Lippman, 1978). However, as pointed out by Villchur, one should attempt to reconstruct several loudness contours for the recruiting listener, in order to counteract the expansion and high frequency attenuation that is a result of recruitment. When high frequency emphasis is used without compression, high frequency environmental sounds may be amplified even beyond the LDL for the hearing-impaired subject.

In order not to exceed the discomfort level for the hearing aid wearer, all hearing aids incorporate some means of output limitation. One technique used to limit the maximum power output is peak clipping. In this form of output limitation, increased input to the hearing aid does not result in an increase in output beyond the maximum power output. This introduces a non-linearity with respect to increase in intensity. The resultant amplitude distortion is called peak clipping, a condition in which the tops and bottoms of the waveform are clipped off. Peak clipping has been the traditional mode of output limiting for non-compression hearing aids, possibly because it is also the easiest form of limiting to implement. X dB of peak clipping is defined as the ratio of the clipped signal to the unclipped signal.

The effect of peak clipping is to produce high frequency distortion. For periodic signals, the distortion takes the form of harmonic distortion (Hodgson and Skinner, 1977). Nonetheless, the normal ear can tolerate as much as 24 dB of peak clipping, and still achieve a discrimination score of more than 95% for monsyllabic words (Goetzinger, 1978). Licklider's (1946) experiment has shown that the effect of this type of amplitude distortion on intelligibility, in both quiet and under certain conditions of noise, is relatively small. In fact, clipped speech was found to be more intelligible than unclipped speech if the waves were equated in terms of peak instantaneous amplitude. Pollack and Pickett (1959) also demonstrated this concept in a study utilizing a wide range of conditions, including peak clipping of 0, 6, 12 and 24 dB. In demonstrating the effect of symmetrical peak clipping on normal hearing listeners, post-clipping gain was introduced to keep the post-clipping speech power constant. Their conclusion was that intelligibility was independent of the level of peak clipping, if the post-clipping level is held constant.

Whereas for the normal listener, distortions are tolerable because of the redundancy inherent in speech, for the hearing-impaired listener it is important to retain as many cues of speech as possible. The effect of peak clipping for the hearing impaired, who are dealing with distortions inherent in their hearing loss, can be very detrimental, and it is therefore desirable to minimize its effects. In a series of experiments, Harris, et al. (1961) found that intelligibility, as measured by speech discrimination tests, can be severely degraded by harmonic distortion generated by peak clipping in a hearing aid. They advanced the hypothesis that this result was related to a masking effect of harmonic distortion products of the fundamental upon the formant structure of the phoneme. When harmonic distortion was less than 20%, they did not find it to have a significantly degrading effect on speech discrimination. Peak clipping was also one of the variables investigated in a Wearable Master Hearing Aid study (Levitt and White, 1978). Their finding was that the wider the range of linear amplification compatible with the subject's comfort, the better the intelligibility score.

It is thus recommended that in hearing aid fitting, a maximum power output be utilized that is at as high a levelas the patient can tolerate comfortably. This will result in the maximum possible dynamic range prior to peak clipping. On the other hand, the output of the hearing aid must be limited to some extent, for reasons of both safety and comfort. Evidence of potential hearing damage, in the form of threshold shift, as a function of excessive amplification, has been provided by researchers such as Macrae and Farrant (1965) and Ross and Lerman (1967). However, since peak clipping may reduce intelligibility for the hearing impaired, another possibility of output limitation would be to use compression amplification, as discussed in the next section.

## Compression Amplification Systems

A common complaint of many hard-of-hearing persons is that when the level of speech is raised enough for the faint sounds to be heard, the more intense sounds become annoyingly loud. In addition, since the sound field may change rapidly, a hearing aid wearer may need to continually re-adjust the gain control in order to listen comfortably, despite the abrupt changes in environmental sound levels. Because the speech signal itself is also constantly fluctuating, and the dynamic range of the hearing-impaired listener is limited, some of the speech signal is lost by being below threshold of audibility, or is intolerably loud by being above LDL.

Additional problems are introduced by the nature of the speech spectrum. The subtle distinctions among consonants, which convey important information, involve cues which lie primarily in the high frequency portion of the speech spectrum. This portion is, however, of low amplitude. /ɔ/, the most powerful speech sound, is about 28 dB more intense than /θ/, the weakest speech sound, exemplifying the great dynamic range that occurs in speech (Fletcher, 1953). The average intensity of consonants is about 12 dB weaker than that of vowels (Kretzinger and Young, 1960). In addition, the dynamic range of conversation, encompassing low levels to high levels, spans roughly 30 dB in any one frequency region. Thus, the range of difference among speech sounds is extended even more.

Problems may arise when using a fixed frequency response (such as the response recommended by Davis, et al., 1947) on pathological ears demonstrating recruitment. If the vowels are brought to a comfortable listening level within the listener's area of audibility, then the unvoiced consonants may be below the threshold of audibility. But the listener cannot use enough hearing aid gain to bring these weak consonants into his useful dynamic range of hearing, because that amount of gain would amplify lower frequency, high amplitude vowels to be intolerably loud (Watson and Knudsen, 1940). Although 60% of the power of speech lies at and below 500 Hz, this low frequency energy controlling the listener's setting for gain, 95% of the intelligibility lies at and above 500 Hz (Gerber, 1974). The end result is poor intelligibility at any tolerable level, since the listener inevitably reduces the amplification, even though the consequence is amplification insufficient for the consonants present in speech.

Shaping the frequency-gain response of the hearing aid to match the listener's residual hearing helps alleviate the above problem to some extent, but does not offer the same flexibility as does compression amplification. A compression amplifier functions so that a change in the level of the input signal modifies the system's gain, i.e., the input-output ratio is not uniform over the entire range of levels. It works as a regulatory system, moving the gain up and down variably, depending on the input level.

The objective of a compression hearing aid has been summarized by Nabelek (1973):

1. As protection to the user from sounds that are too intense and may cause discomfort, or even damage, to the ear. Compression may be preferred over peak clipping, the latter also able to achieve the same objective, since the latter involves more distortion.

2. To keep the average output level within a smaller range, i.e., more constant, than the input level. This would eliminate the need for constant readjustments of the volume control, irrespective of the changes of the average input sound pressure level. This is helpful to the listener in situations where intensity changes occur, such as those that occur as a consequence of change in distance from the source. The listener can then continue to hear faint material, while still retaining comfort in listening to intense materials.

3. To aid persons with a limited dynamic range, due to, for example, recruitment. In this case, the input dynamic range of the sound is matched to the available dynamic range of the hearing-impaired person. That is, the amplitude distribution of the acoustic signal is modified to better match the residual auditory function of the listener. Low intensity sounds are amplified fully, while high intensity sounds are amplified proportionately less. The dynamic range of the hard-of-hearing person is thus utilized more fully, since the varying intensities of

speech and noise are all compressed to within this range. The weaker speech phonemes are amplified fully, while the peak vowel sounds are kept at comfortable levels, through compression.

Parameters of Compression Amplification

The parameters of compression amplification systems include the time constants, which are the attack and release times; the compression ratio; and the compression threshold. These parameters are discussed in detail in Chapter III. However, for the purposes of this discussion, each parameter will be defined briefly before discussing its effect. The potential value a hard-of-hearing person receives from a compression amplification system will be determined, in part, by these parameters. This is in addition to the nature of the user's hearing loss, and the nature of his acoustic environment, in terms of levels and types of stimuli.

1. Time Constants

When there is an abrupt increase in input level, a finite amount of time is required for a compression amplifier to adjust to its new gain. The time taken from the onset of the increase in signal level to that instant when the amplifier gain stabilizes (to within 2 dB of the steady-state value) is defined as the attack time. The release time, or recovery time, is the time elapsed from the decrease in signal level to the instant the amplifier gain stabilizes (to within 2 dB of the steady-state value)

(ANSI, 1976). For convenience of measurement, attack and release times are specified in terms of the change in output signal level. A schematic diagram showing time constants is illustrated in Figure 3.

The attack and release times have also been referred to in the literature as, respectively, overshoot and undershoot. These latter terms are, however, inappropriate, since the system requires a finite amount of time in order to detect that an increase or decrease has occurred in the input signal voltage, and to then make its appropriate adjustments (that is, to compress, or to release from a state of compression, respectively).

Since the goal of compression is to reduce signals that are too intense, and to do so without distorting them, it is important to know what the optimal time constants are, in terms of the least distortion and the best intelligibility.

Addressing themselves to this problem, Lynn and Carhart (1963) constructed a compression hearing aid from hearing aid parts, with which they then evaluated 36 hearing-impaired people with a diversity of hearing losses. They employed 10 testing conditions, one having no compression, and the other nine with various combinations of attack and release times. Testing at one listening level, they found that intelligibility of monosyllabic PB-50's was affected by changes in the relations of time constants. As recovery time increased beyond 150 msec., intelligibility for all groups tended to decrease. Sensorineural impaired listeners scored best when the attack/release time combination was 5/150 msec.

Nabelek (1973) chose seven commercial post-auricular hearing aids that all had similar middle range frequency responses, with attack times ranging from 6 to 130 msec., and release times ranging from 30 to 580 msec. The Modified Rhyme Test was recorded through each of these hearing aids, at an SPL of 85 dB. The signal was then presented monaurally, from the tapes, to 10 normal hearing listeners, by earphone. The tests were presented at post-compression signal-to-noise ratios of -2 dB and 2 dB. Nabelek found that shorter time constants yielded higher intelligibility scores, especially under more difficult listening conditions. The two aids with shortest attack/release time combinations, of 6/30 and 17/50 msec., respectively, were always in the top three in terms of good intelligibility.

The results of Nabelek's study diverge from other studies, mentioned above. Nabelek's conclusions regarding the shorter release time advocated are similar, though, to other early research. Edgardh (1952) and Kretzinger and Young (1960) suggested release times of 20 and 22 msec., respectively.

The two time constants affect the speech in different ways. The attack time may be as short as 1 msec. without creating audible distortion (Villchur, 1973). If attack time is rapid, compression will be quickly activated by intense sounds of short duration. With release time being, relatively, moderately long, consonants will pass through while the system is still in a state of compression. If release time, however, is also short, consonants pass through the system when it is no longer in a state of compression. This results in a limiting of the overall intensity of the speech signal, which is desired, but with a reduction in the intensity differential between vowels and longer consonants. The result is a change of the normal intensity ratio between the vowels and consonants.

Research has attempted to relate time constants to intelligibility. However, as with all studies using commercial hearing aids, a change in one parameter of a hearing aid may result in changes in other parameters. One cannot, therefore, attribute results to one specific parameter rather than to a possible interaction effect. This limits both the experiments' interpretations, and the ability to isolate one parameter across several studies and compare them. In addition, some studies (for example, Nabelek, 1973) were conducted on normal hearing listeners, and extending results on the normal hearing to the hearing impaired must also be tentative. However, physical measurements obtained by Nabelek do demonstrate that hearing aids with relatively shorter time constants have less harmonic distortion, and less distortion during the attack time. Thus, even with the above limitation in mind, the tentative conclusions do favor shorter time constants, to a lower limit, for compression hearing aids used with the hearing impaired.

Although there are conflicting data regarding attack and release times, there is at least one area where there is general agreement, and that is the relative durations of attack and release times. The release time should be longer than the attack time, since, if it is not, severe waveform distortion would be introduced. This would happen as a result of compressor action following the instantaneous amplitude of individual cycles (Villchur, 1973). On the other hand, if the release time were too long, for example, one second, to correspond to the average length of a two-syllable word, sudden decreases in the environmental sounds would create long dead periods.

2. Compression Ratio

The compression ratio is the decibel change in input divided by the decibel change in output, as illustrated in Figure 2. If, for example, for a 20 dB input change only 5 dB of output change results, the compression ratio equals 20/5 or 4:1 (a compression ratio of greater than 1:1). If, on the other hand, a decibel change in input results in a greater decibel change in output, then expansion is taking place. For example, if for a 10 dB input change, 20 dB output results, the compression ratio equals 10/20, or 1:2 (a ratio of less than 1:1). A non-compression hearing aid operates on an input-output ratio of 1:1.

Compression ratio has been the subject of investigation of several studies, which are discussed later in this review.

3. Compression Threshold

Compression threshold is that input level at which compression begins, as illustrated in Figure 2. This parameter has been related to intelligibility. Too low a compression threshold will both increase the relative background noise, and cause excessive noise to be present during periods of no speech input (Lippmann, et al., 1976). It is important to use a compression threshold that is no lower, and a compression ratio no higher, than is needed to achieve good intelligibility (Villchur, 1973). Lippmann, et al. suggested reducing the effects of this degradation by appropriate microphone selection and placement, to provide the best possible input signal-to-noise ratio. Another possibility for reducing its effects, aside from raising the compression threshold, is by providing expansion below the compression threshold (Villchur, 1973). The exact specification of this parameter needed to achieve optimum intelligibility is yet another area for research in compression amplification.

Distortion in Compression Hearing Aids

As with non-compression hearing aids, there are problems that are inherent in compression hearing aids. Some of these problems are reviewed by Lippmann (1975). For example, one problem is a "thump" sound, which is a low frequency transient accompanying gain change, and harmonic distortion caused by the nonlinearity of the gain control. Too long an attack time before compression sets in is another problem that may lead to audible excessive gain. In addition, excessive noise may be introduced by the gain-controlling device, or during periods of no-speech input, caused by too low a compression threshold. The former is internal noise of the device itself, and the latter is unnecessary amplification of the external noise.

Compression Amplification and its Relationship to Speech Intelligibility

There are two basic types of compression amplification: the compressor and the limiter. The compressor amplifier operates at a compression ratio greater than 1:1 throughout its dynamic range. A limiter amplifier, on the other hand, operates as a conventional amplifier over most of its dynamic range, and, once compression is initiated at a very high compression threshold, the compression ratio is high. As is noted below, many of the early studies used the latter type of compression.

The data concerning the effects of compression amplification on the intelligibility of speech for the hard of hearing is both limited and contradictory. One of the first studies in this area was conducted by Davis, et al. (1947). Among the various parameters they evaluated, they studied the effect of replacing the peak clipper of their Master Hearing Aid with a limiter compression amplifier. The attack and release times were 1 msec. and 200 msec., respectively. Compression began at a peak output of

117 dB SPL, with a compression ratio of 10:1. In testing three normal hearing subjects with amplification they introduced compression over a range of 30 dB, or peak clipping over a range of 30 dB. Maximum intelligibility deteriorated less under the condition of compression than under the condition of peak clipping. The same result was found for six hard-of-hearing subjects tested under the conditions of flat gain, and high frequency emphasis of 6 dB/octave.

Hudgins, et al. (1948) continued the earlier work begun at Harvard University (Davis, et al., 1947), and demonstrated the desirability of incorporating compression circuitry into a wearable hearing aid. They compared an experimental hearing aid to the Harvard Master Hearing Aid, and to two commercial instruments. Using monosyllabic PB words, with signal-to-noise ratios of 15 dB or 20 dB, they tested six listeners with average losses of 48 to 63 dB in the speech frequencies. With compression amplification the maximum level of performance was maintained over a wide range of speech-input levels.

Kretzinger and Young (1960) tested the hypothesis that compression is superior to peak clipping in improving intelligibility for PB words in noise. They measured intelligibility for 30 normal hearing listeners, using compression amplification over input ranges of 10 dB and 20 dB, compared to no compression, and to peak clipping as the limiting mechanism. Input level for the words was 70 dB SPL. Scores with no form of limiting were an average of 57%. With peak clipping of 10 dB and 20 dB, scores were 63% and 61%, respectively, and with compression of 10 dB and 20 dB, scores were 85% and 78%, respectively.

This early trend of finding improvement with compression amplification was reversed by several studies performed in the 1960's.

Lynn and Carhart (1963) measured the effect of various attack and release times of a limiter compression hearing aid on speech intelligibility for 30 hard-of-hearing subjects. Attack/release times ranging from 6/30 to 20/500 msec. produced little change in discrimination scores, and little improvement in intelligibility. Averaged over 10 subjects, the maximum increase of compression amplification over non-compression amplification was about 9% at the best attack/release time combination of 5/150 msec. For attack/release times greater than 20/500 msec., there was a negative effect on discrimination. Their conclusion was that proper time constants must be chosen for maximum improvement of intelligibility with compression. They also interpreted their results to indicate that compression is not always an advantage to the hearing impaired.

Some of the problems with Lynn and Carhart's study were as follows:

1. The compression hearing aid was built from hearing aid parts, and, in all likelihood, had many distortion characteristics typical of commercial compression hearing aids.

2. Although there were differences demonstrated among various attack/release time combinations, these differences may have been attributed to differences in presentation level of the monosyllabic word lists used to measure intelligibility. That is, spondee words used to measure Speech Reception Thresholds (SRT) were presented in isolation. When the attack time was long, the first half of the spondee was amplified by the compressor to maximum gain, yielding an artificially better (i.e. lower in intensity) SRT. Lynn and Carhart's data proves this point: independent of release time, SRT's decreased by about 10 dB as attack time increased from 5 to 85 msec. Monosyllabic words were then presented at 25 dB Sensation Level (SL) re SRT. But since intrasubject SRT's differed as a function of the attack time, the words were actually presented at different levels for the various attack times. The differences in scores obtained for various attack times may thus have been influenced by the presentation levels, especially since, in the critical region of 10 dB to 20 dB above SRT, PB function increases by about 4 or 5%/dB.

3. The compression ratio was 5:1, which may have been too high, as indicated by more recent studies.

4. The discrimination scores of the subjects were relatively high to begin with, under the control condition of non-compression amplification. The mean scores were 90.7% for the otosclerotic group, 77.2% for the group with Meniere's disease, and 73% for the presbycusic group.

Thus, there may not have been enough room for improvement for some groups.

5. The type of compression used was the limiter type. A limiter compressor functions as a non-compression amplifier until a critical level, which is very high. Thus, it is only the most intense peaks of speech which are compressed, rather than all or most of the speech signal.

In 1967, using a compressor hearing aid, Caraway and Carhart investigated the hypothesis that the intelligibility of speech can be increased by a reduction in its dynamic range, and that subjects with sensorineural hearing loss can benefit more from compressor action than can normal listeners.

Their rationale for the first hypothesis was as follows: Tillman, et al. (1963) found a span, in normal hearing listeners, of 24 dB for recognition from the least audible CNC words to the most audible. Based on this finding, Caraway and Carhart reasoned that, with 2:1 compression, this 24 dB range could be reduced to 12 dB, and, with 3:1 compression, it could be reduced to 8 dB. Much greater drops in input should be able to take place with compression, without a radical drop in intelligibility, increasing the usable dynamic range of input signals. They reasoned that at the various levels in which they were interested, all the phonemic elements comprising the most difficult words should be perceptible, leading to excellent discrimination at these reduced levels.

Their second hypothesis was based on the finding that patients with recruiting ears have a disrupted relationship of normal loudness, leading to a reduction in speech discrimination. They, therefore, need a reduced input dynamic range, as compared to the normal hearing listener, to perceive speech most advantageously.

Caraway and Carhart investigated the effect of compression on intelligibility of monosyllabic CNC words in quiet, using three compression ratios: 1:1 (no compression), 2:1 and 3:1. The speech signals were adjusted to achieve presentation levels of 0, 8, 16 and 24 dB re SRT, measured under each condition of compression. Their subjects consisted of 12 individuals with normal hearing, and 36 recruiting hearing-impaired individuals. There was a small but relatively consistent trend for performance during the non-compression condition to be poorer than performance during the condition of compression. Normal hearing listeners improved the most with compression, as much as 15% for 3:1 compression at 8 dB SL. The 2:1 and 3:1 compression conditions were essentially equivalent. For the recruiting listeners, they were unable to demonstrate any substantial enhancement in intelligibility with compression.

Some of the problems with Caraway and Carhart's study were as follows:

1. Although there were three separate channels for filtering and amplifying the speech spectrum (200-1000 Hz, 1000-2000 Hz, and 2000-5000 Hz), they did not take full advantage of them. Each of these channels could be controlled

independently. Nonetheless, Caraway and Carhart used one constant compression ratio for all of the three frequency channels during any one compression condition. However, despite the fact that only one compression ratio was used at a time, the fact that the signal passed through three separate frequency bands caused compression to occur at various times for one or more of the bands, but not for the other(s). For example, if a strong vowel passed through, it would drive the low, and perhaps also middle, frequency band into compression. However, there would not necessarily be a strong frequency component to drive the high (2000-5000 Hz) frequency band into compression, despite the fact that its compression ratio was set the same as for the other two bands. Their system, therefore, did not truly simulate a one-band compression system, in which any sound that drives the system into compression will compress all frequencies. Nor did it simulate a multiband compression system (discussed later in this review), in which different compression ratios for the various channels are deliberately chosen to match the needs of the individual listener. Thus, compression did not necessarily occur throughout the frequency range under the conditions of compression, but may have allowed non-compression amplification to persist in one or more of the frequency bands, even during the condition of compression.

2. There were high levels of harmonic distortion at all input levels, and especially for the low frequency band.

The deleterious effect of harmonic distortion on speech intelligibility has been discussed above.

3. All testing conditions were performed in quiet only, a condition that may not have been sufficiently sensitive to bring out differences among hearing aid variables.

4. Caraway and Carhart found no significant differences between the non-compression and compression conditions. However, discrimination scores were not measured at levels higher than 24 dB SL, in which case more of the intelligibility function would have been covered. Testing at higher levels would have shown whether the excellent scores obtained for all conditions would have exhibited a decrement in discrimination with increasing intensity.

5. Their reasoning for their first hypothesis was based on a 24 dB range required for complete recognition of CNC monosyllabic words for normal listeners (Tillman, et al., 1963). However, this 24 dB range may not have been appropriate for their hearing-impaired subjects.

Burchfield, et al. (1971) used a slightly modified version of the compressor employed by Caraway and Carhart, but they reduced the harmonic distortion by filtering. They measured intelligibility for 36 recruiting subjects, each with a unilateral sensorineural hearing loss, at 24 dB SL re SRT. The non-test ear routinely received broad band masking. Using compression ratios of 1:1, 2:1, and 3:1, Burchfield, et al. found improved intelligibility for the latter two conditions. The average improvement was from 63.1% to 73.7% for a compression ratio of 2:1 over the 1:1 condition, and to 75.4% for a compression ratio of 3:1. The subjects had been classified as to the amount of recruitment on the basis of the results of an ABLB test administered using noise with a spectrum like that of speech. Both groups, however, whether classified as having partial or complete recruitment, improved by similar amounts. Burchfield, et al therefore suggested that there was no systematic connection between compression and the amount of recruitment.

Since the study by Caraway and Carhart (1967) and that by Burchfield, et al. (1971) employed the same compressor amplifier, the results of the two studies appear to be paradoxical. However, Burchfield, et al. may have gotten improved discrimination, even though Caraway and Carhart did not, because of a low frequency filtering process that the former used. This process markedly reduced noise and harmonic distortion, and may have lessened the masking effect that the low frequency fundamental vowel energy is seen to exert on discrimination (Danaher and Pickett, 1973). In addition, individual differences among subjects in the two studies may have contributed to the differences in results.

Because of the discrepancy in the findings of Caraway and Carhart (1967) and Burchfield, et al. (1971), Vargo and Carhart (1972) conducted a study replicating the design of the former, using a different commercial compressor system. Their design was an attempt to validate the procedure of Caraway and Carhart, and to further resolve the differences between the two studies. Vargo and Carhart employed three compression ratios of 1:1, 2:1, and 5:1, and presented CNC monosyllables at input levels of 10, 20 and 30 dB SL at each compression ratio. Attack time was 100 nanoseconds, and release time was 20 msec. The monosyllables were recorded in groups of triplets, to simulate temporal sequencing as in connected discourse, and in order to compress the entire sentence-like length during processing action. The dynamic range of the words, before compression, was 24 dB, and was reduced to 12 dB for the 2:1 condition, and about 5 dB for the 5:1 condition. Two groups were tested, one consisting of 12 normal hearing subjects, and the other of 12 hearing-impaired subjects, otologically diagnosed as having Meniere's disease. Their results supported the conclusions of Caraway and Carhart: as sensation level increased, intelligibility increased. However, neither the normal hearing group nor the hearing-impaired recruiting group demonstrated increased intelligibility as a function of degree of amplitude compression, at any sensation level tested. Among the three conditions, scores differed by no more than approximately 3%, at any sensation level.

Vargo and Carhart did not use a low frequency filter, as did Burchfield, et al., so that the energy of the fundamental frequency may have exerted a detrimental masking effect on the intelligibility of the hearing-impaired listeners. In addition, although the former used a different compressor unit, the same problems that existed in the Caraway and Carhart (1967) study, mentioned above, existed in the study by Vargo and Carhart.

Research continued to be conducted as to the usefulness of compression amplification, and continued to yield paradoxical results among studies. Bearing in mind that speech discrimination loss is related to loudness recruitment (Hood and Poole, 1971), Fleming and Rice (1969) carried out a study on the effect of compression amplification on speech discrimination, over a limited compression range. Using a compressor type of aid, subjects listened to recorded test materials at several compression ratios. Their results suggested that compression was a positive way to improve speech discrimination. Since they did not find the optimum compression ratio to be the inverse of the recruitment slope line, they questioned the premise that the optimum compression ratio is related to the amount of recruitment. They found that a compression ratio of 2:1 or 3:1 improved speech discrimination for both normal listeners and the hearing impaired, and going from 3:1 to 5:1 compression ratio did not result in any further improvement. They pointed out that it is not important to restore the entire speech signal to audibility. Use of a high compression ratio is generally based on the desire to make the speech signal available where the hearing loss is worst, usually in the high frequency area. However, as noted, the result may be a degradation of speech intelligibility, rather than an improvement.

The finding that the optimum compression ratio is not necessarily the inverse of the recruitment slope line was confirmed by Trinder (1972). He attempted to restore the normal loudness relationships in unilaterally impaired listeners with sensorineural hearing loss, by designing a compression system that counteracted recruitment. His results indicated that although recruitment for pure tones was effectively corrected, measured by unaided versus aided ABLB tests, discrimination was improved by as much as 20%, but not restored to normal. The improvement of 20% was found in only one of five subjects; on the average, though, the amount of improvement was substantially less. He suggested that intermodulation components of both internal and external noise, and the non-linearities of the system, may have prevented his results from being better.

Another study in this area was conducted by Yanick (1973). He tested 12 subjects with a mild to moderate sensorineural hearing loss with a custom-fitted compression amplification hearing aid. The compression ratio was chosen as either 3:1, 2:1, or 1.3:1, depending on the subject's dynamic range between the Speech Reception Threshold and the Loudness Discomfort Level for speech. The attack and release times for all the compression ratios was 1 msec. and 30 msec., respectively. Yanick tested discrimination for PB-50's in sound field, under the following conditions: unaided; with the patient's own non-compression aid; and with the custom-fitted compression hearing aid, i.e. with input-output function matched to the patient's dynamic range. Discrimination scores were measured in quiet, at input SPL's of 45 and 70 dB. When scores for the latter two conditions were compared, the compression condition indicated an increased aided input dynamic range, improved tolerance for loud speech, and sharply improved discrimination score. At 45 dB SPL, scores went from 68.5% with the subject's own aids, to 87.5% with the compression aid. At 70 dB SPL, they rose from 39% to 91% with compression.

A problem with the experimental design was that Yanick compared a high quality compression amplifier to the subject's own hearing aids. The two conditions may not be comparable because of the difference in quality. In addition, the transmission characteristics between sound traveling through the earphone, as in the compression hearing aid condition, and through the earmold, as in the condition with the subjects' own hearing aids, should be taken into account if conditions are to be compared. Another problem may also have been that the frequency response for the compression condition may have been better suited to the subjects' needs than the frequency response for the non-compression condition.

It is evident that various studies have yielded contradictory results as to the benefits, if any, of compression amplification on speech intelligibility. Because of differences other than compression that may have influenced subjects' performance, such as low frequency cut-off and signal-to-noise ratio, some of the studies are not directly comparable. When researching how different electroacoustic parameters of hearing aids, whether with or without compression, influence performance, one should also take all possible sources of distortion into account. This factor becomes especially important when comparing various studies.

The amount that each possible source of distortion may be tolerated is one area of research in hearing aids. Since many signal parameters influence transient response characteristics in hearing aids, the performance of the compressor in actual situations becomes difficult to predict. The importance of different characteristics of compression therefore has to be made by psychoacoustical measurements, intelligibility tests, and quality judgments (Nabelek, 1973).

Nabelek measured the effects of transient distortion for seven hearing aids, using the Modified Rhyme Test as the stimulus. All measurements were performed with the hearing aid attached to a 2cc coupler in a Bruel & Kjaer test box, using a maximum input of 85 dB SPL, and abrupt changes of 30 dB (i.e. changes from 85 to 55 dB SPL, and from 55 to 85 dB SPL). Because harmonic distortion was observed in several hearing aids during attack and release times, Nabelek observed that steady-state measurements of harmonic distortion are not sufficient in describing performance of hearing aids with compression. Testing 10 normal hearing and 10 hard-of-hearing subjects in noise indicated that the greater the distortion, the poorer were the scores. This is a similar finding to that of Burchfield, et al. (1971), who found that, when comparing their results to those of Caraway and Carhart (1967), any benefit resulting from compression could be lessened by an increase of steady-state harmonic distortion.

In summary, as a form of amplitude limiting, compression amplification has definite advantages over non-compression amplification. These include compressing the output dynamic range, and limiting output without introducing the distortion introduced by peak clipping. However, as discussed in this review, differences between non-compression and compression amplification as a means of improving intelligibility have not been systematically shown. Problems within studies mentioned may have been responsible for the lack of superiority of the latter form of amplification over the former. In addition, it is possible that not all frequencies were properly compensated for in terms of amplitude, even with compression. Because recruitment is a frequency-dependent phenomenon, merely compressing the entire frequency range of speech may not be sufficient to compensate for the listener's speech discrimination difficulties. This concept has led to the use of compression in a multiband form.

Multiband Compression Amplification

As noted above, recruitment is a frequency dependent phenomenon. Therefore, by dividing the speech signal into an infinite number of frequency bands, and compressing the signal in each band independently, theoretically, ideal amplification might be realized. The amplification for each band after compression could then be equalized, that is, adjusted to offset the subject's recruitment at that frequency. Equalization would thus place this variably compressed speech at various desired levels over the frequency spectrum. This type of processing would compensate for problems caused by recruitment, as discussed by Villchur (1974). That is to say, those elements of speech falling below threshold would be amplified, and the expansion and treble attenuation experienced by the recruiting listener would be compensated for by the proper amounts of compression at various frequencies.

Villchur (1973) approached this theoretical construct by using a hearing aid with two frequency bands, plus post-compression equalization. The use of only two bands provided limited flexibility, but, nonetheless, proved adequate in his experiment for matching the frequency-dependent recruitment of his subjects. Villchur first ran a preliminary study testing his system on two normal hearing subjects who had a loss temporarily induced by shaped noise. He found that discrimination for final consonants of nonsense syllables rose from 38% to 84% with compression plus post-compression equalization, which was a significantly greater improvement than was realized from either compression alone or equalization alone. Villchur then tested six subjects with sensorineural loss, in both quiet and in noise, with a two-channel compressor plus post-compression equalization. The frequency-dependent compression ratio was adjusted to compensate the recruitment of the individual subjects. Calculations were made on the basis of pure tone thresholds, LDL, equal loudness contour at MCL, and characteristics of speech. Villchur compared the span of threshold and the equal loudness contour at MCL to the normal span between threshold and equal loudness contour. The compression ratio used was the ratio of normal to abnormal span. He first made the calculations himself, and found the mean compression ratio to equal 3.5:1 for the high frequency band, and 2.2:1 for the low frequency band. Subjects were then allowed to adjust the ratios themselves, and they chose 2.8:1 and 2.1:1 for the high and low bands, respectively.

The compressed speech was then subjected to frequency-selective amplification at each frequency co-ordinate of the speech band, similarly adapted to the subject. This manner of processing amplified each acoustical element of speech to a relative loudness for the hard-of-hearing subject corresponding to the relative loudness of that speech element as perceived by normal hearing listeners. Use of equalization allowed for a high frequency emphasis. The high frequency elements would otherwise have remained below the subject's hearing threshold, if the lower frequency elements, basically the vowels, had been kept at a comfortable level.

This type of multiband compression is theoretically advantageous, since it allows the amplitude ratio between simultaneous speech elements to be changed, and not just between successive elements. In addition, processing in Villchur's system continued up to 5.6 kHz, as opposed to most current hearing aids, whose high frequency cutoff is generally about 4 kHz. This additional high frequency information is important for the hearing aid user, since he has poor resistance to interference. Even normal listeners, who do not suffer great intelligibility losses with high frequency attenuation in a quiet environment, suffer a significant reduction in intelligibility when other distortions, such as reverberation, are added to the high frequency attenuation (Martin, et al., 1956). The hearing aid user listens in environments of reverberation, noise, and competing speech, combining external distortions with his already present internal perceptive aberrations. Compression without equalization in a hearing aid may enable him to understand continuous speech under ideal conditions, but, with interfering destructive influences, enough of the redundant information normally present in speech is absent. Villchur (1974) reasoned that compression plus equalization restores redundant recognition cues, making speech more resistant to interference than is the intelligibility of speech that has been compressed only.

Materials were recorded by a female speaker in a slightly reverberant environment. Villchur (1973) found that processing raised the intelligibility of CVC syllables for his six subjects in both quiet and in noise. Discrimination was measured at MCL, at MCL minus 10 dB, and at MCL minus 20 dB. The noise used was competing speech introduced before processing at a signal-to-noise ratio of 10 dB. Initial- or final-consonant recognition improved between 22% and 160% in quiet, and between 10% and 229% in noise.[2] Terminal consonants received the lowest unprocessed scores, possibly due to forward masking from the previous vowel, and/or the natural drop in the speaker's voice at the end of a syllable (Villchur, 1973). The terminal consonants showed the greatest improvement from processing. Vowels showed either no change, or slight improvement, with compression. Subjectively, processed speech was rated as equal to or superior to the unprocessed speech in pleasantness.

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[2] Villchur arrived at these percentage points by calculating as follows:

$$\frac{\text{compressed score} - \text{uncompressed score}}{\text{uncompressed score}}$$

Villchur's study was one of the few to use a high quality compressor, and to use compression and equalization that was individually matched to his subjects. He allowed his subjects to make their own adjustments of compression ratio, frequency equalization, and presentation level. In addition, his hearing aid had low distortion and noise characteristics, thus being of better quality than many of those used in previous studies.

In a similar study, Yanick (1976) measured the effects of signal processing with a two-channel compressor. He tested three groups of 12 subjects each: with normal hearing; with a flat-to-gradual sloping hearing loss; and with a ski-slope hearing loss. All subjects in the latter two groups had recruitment, as measured by relationships between thresholds of hearing, MCL contours, and Equal Loudness (EL) contours at suprathreshold levels, all for pure tones.

Subjects in Yanick's study adjusted the output for maximal clarity and a comfortable listening level. In addition, they adjusted the high band compression ratio, the low band compression ratio, the high band gain, the low frequency rolloff, and the high frequency response. Their choice for compression ratio was an average of 1.5:1 and 2.5:1 for the low and high bands, respectively, for the ski-slope group. For the flat loss group, it was 2.4:1 and 2.8:1 for the low and high bands, respectively. Bass rolloffs were 12 dB/octave for the flat-to-gradual slope group, and 6 dB/octave for the ski-slope group.

Intelligibility for processed speech was compared to unprocessed speech, in which subjects again chose an optimal output listening level based on intelligibility and comfort, and frequency equalization for low frequencies only. In addition, 12 of the subjects were tested with their own hearing aids, adjusted to a preferred level. All subjects were tested with 30 lists of Harvard sentences, as revised by IEEE (1969), recorded in a reverberant environment, with competing cafeteria noise on another track. The material was presented at signal-to-noise (S/N) ratios of 0 dB and 6 dB.

Yanick's results showed significant improvement in intelligibility for processed speech over unprocessed speech. At S/N = 6 dB, mean improvement was 28.2% for the ski-slope group, and 37.4% for the flat loss group. At S/N = 0 dB, mean improvement for the former group was 38.5%, and for the latter group, 39.4%. Subjectively, subjects reported that the processed speech had more clarity and was more comfortable to listen to. Intelligibility scores while wearing their hearing aids were within 2.1 - 2.4% of unprocessed scores.

Using a two-band compression system, Yanick and Drucker (1976) tested six subjects with 25 to 60 dB pure tone averages. All losses were ski-slope, with 15 to 25 dB loss per octave, starting in the midfrequencies. All subjects had recruitment present, determined by measurements of threshold, MCL, and equal loudness contours.

Test stimuli were the Harvard sentences (1969), as noted above. A 1000 Hz calibration tone was set equal to the average peaks of speech. Subjects were tested for both processed speech and unprocessed speech, the latter using the subject's own hearing aid, at signal-to-noise ratios of 0 dB and 6 dB. Experiments for processed speech were carried out with compression, and conventional amplification below compression threshold, and with compression, but with expansion below threshold. The latter condition was used to overcome the degradation inherent in compression systems, of the relative level of background competition increasing, especially during periods of silence.

Discrimination scores revealed the poorest scores with the unprocessed speech (subjects' own hearing aids), and the greatest improvement in the condition of compression, with expansion below compression threshold.

Here, too, as in Yanick's (1973) experiment, Yanick and Drucker compared a high quality compressor to the subjects' own hearing aids. The latter is generally of poorer quality, raising a question as to the validity of the use of the hearing aid as the referent condition. In addition, Yanick and Drucker did not use treble boost for unprocessed speech, but did for processed speech, creating even greater differences between the two conditions, other than the parameters under test. And, as noted before, the frequency response chosen may have been better suited for individual subjects.

Barfod (1976) conducted experiments on five bilaterally impaired hearing aid users with sensorineural hearing loss, using a multichannel compression system. His testing was based on the hypothesis that if the system was individually fitted to restore normal equal loudness contours, it would result in improved discrimination for the hearing impaired, as compared to non-compression amplification. All subjects had normal hearing at 500 Hz and below, and sharply sloping losses of at least 50 dB at 2000 Hz and above. Using the normal low frequency hearing as the reference, equal loudness contours at various levels were derived. Comparisons were made for four conditions: one non-compression condition that had a sharp low frequency rolloff below 750 Hz, and was then optimally fitted to the subject's hearing loss, and three compression conditions. The latter consisted of one non-compression low frequency channel for all three conditions, with the addition of one, two, or three high frequency compression channels. All parameters of the compression system, including low frequency and high frequency cutoffs, crossover frequencies between bands, compression ratios, and relative gain among channels (the latter being a rough form of equalization), were chosen to restore normal equal-loudness contours.

Testing was conducted with CVC nonsense syllables both in quiet and at various S/N ratios. All conditions employed a male speaker, with a fixed input at 65 dB SPL. Attack time varied for various frequencies, from 24 msec.

at low frequencies, to 6 msec. at high frequencies.

Barfod's results were as follows: discrimination scores for the three-channel compression system were the same as for the non-compression system, optimally chosen. Scores for the one- and two-channel compression systems were significantly worse than for the former conditions.

Barfod's results differ significantly from those of Villchur (1974), the latter finding that compression amplification was significantly superior to a non-compression system. Part of this difference may be accounted for by the difference in hearing losses among subjects in each of these two studies. As has been noted above, what is an optimum frequency response for one type of loss may not be optimum for a different type of loss (e.g. results of Pascoe, 1975 vs. Skinner, 1976). Other differences, such as differences in attack times between the two studies, and differences in input levels may have added to the different conclusions of the two studies.

In a study conducted concurrently with the research in the present study, Lippmann (1978) tested five sensorineurally impaired listeners with a 16 channel computer controlled compression system. Two compression conditions and four non-compression conditions comprised the experimental design. One compression system restored equal loudness contours for pure tones. The second employed a compression ratio of 1:1 at and below 500 Hz, and then increased to a maximum of 3:1 at and above 2000 Hz, based on preliminary experiments, informal listening, and past research on compression. The four non-compression systems employed were as follows: flat functional gain (as used by Pascoe, 1975), also known as orthotelephonic response; the other three all had high frequency emphasis, and consisted of mirroring of the subject's audiogram, most comfortable listening level, and restoration of normal loudness to the 10% levels of speech in each frequency band (the latter as suggested by Barfod, 1972).

The six systems were compared using CVC nonsense syllables, nonsense sentences, and standard word and sentence tests, spoken by male and female speakers, in quiet and in noise. The gain was adjusted by each subject to his respective MCL, and was later also presented at reduced levels.

Lippmann's results were as follows: the best non-compression systems were the three with high frequency emphasis, these three giving roughly equivalent results, and being substantially better than the orthotelephonic system. Performance with the compression system employing restoration of equal loudness contours was worse than the second compression system. And, perhaps most significantly, the scores obtained with the best non-compression system were usually greater than or equal to scores obtained with the better compression system.

These results once again conflict with Villchur's (1974) positive findings regarding multichannel compression amplification. However, if one looks at the differences in the referent, non-compression systems of the two studies, the difference in results may be accounted for. In addition, Lippmann found that when the input was reduced, compression performed better than non-compression amplification, a finding consistent with Villchur's results, the latter having used inputs not only at MCL, but also at levels of 10 dB and 20 dB below MCL.

## Summary

The review of the literature provides evidence that hearing aids are not providing optimum performance for their users. As has been discussed, there are many different ways of fitting hearing aids. They can be summarized as three basic methods. The first of these is individual selective amplification, which is amplification varying in amount at different frequencies, and fitted to the individual. This method includes fitting to equal loudness contours (e.g. Watson and Knudsen, 1940), audiogram mirroring (e.g. Redell and Calvert, 1966), and bisection and variations of bisection (e.g. Wallenfels, 1967; Victoreen, 1973). Another method is that of fixed frequency-gain characteristic, which is amplification utilizing uniform frequency characteristics for most cases of hearing loss. Amplification of this type has included flat frequency-gain, and fixed frequency slopes, such as 6 dB/octave slope (e.g. Davis, et al., 1947; MedResCo, 1944). Thirdly, there is adaptive fitting of a Master Hearing Aid, where a single hearing aid offers a range of hearing aid characteristics, with parameters that are adjustable to best suit the needs of each patient (e.g. Levitt and White, 1978).

Some of these methods, all using non-compression modes of amplification, have had, in the past, various methodological problems, such as lack of sensitivity of test materials (Shore, et al., 1960), and procedural differences among studies. To help avoid some of these research pitfalls, a good description of the electroacoustic parameters of the instrument used, including a description of all possible sources of distortion, is critical. An accurate description of the methods of measuring the parameters, such as frequency response, is also critical if one is to make valid comparisons among conditions tested and/or studies. Recall, for example, how, on the surface, the Harvard (Davis, et al., 1947) and MedResCo (1944) studies appear to have yielded similar optimum frequency response curves. On closer examination, however, when the methods of calibration are considered, there are large differences between the optimum frequency response curves of the two studies (Resnick, 1977).

A serious criticism of many of the early studies on non-compression amplification is that the true overall gain from sound field to eardrum was not specified accurately. Consequently, various frequency-gain responses were rejected. However, today, with more accurate methods of specifying overall gain (e.g. the functional gain method), the conclusions of these early studies are now being questioned. For example, Pascoe (1975) found an audiogram-mirroring technique, when measured in terms of functional gain, to be superior for his particular subjects. Barfod (1972) found that the best performance among several frequency-gain characteristics was achieved with a restoration of the 10% time levels of speech for the hearing impaired to levels that were about the same loudness as the 10% time levels of speech for normal hearing persons. His frequency responses were also measured in terms of functional gain. Skinner (1976) found that the best discrimination scores occurred when the slope of the functional gain between 500 and 1600 Hz mirrored the difference between each subject's threshold curve and the normal threshold curve. And Lippmann (1978) found that various high frequency emphasis methods, such as Barfod's (1972) method, were superior in terms of improving speech discrimination. Methods of frequency shaping such as those of Barfod and Lippmann have come close to Watson and Knudsen's (1940) original proposal, which was to put as much of the speech spectrum as possible into the residual auditory area of the listener.

Thus, despite the findings of the early classic studies, it now appears that significant improvements in performance can be obtained with individual adjustment of the frequency-gain characteristics. It should also be remembered that audiological characteristics of hearing aid wearers are generally different today than they were among the hearing-impaired listeners of the early studies. That is, there are fewer conductive hearing losses, and also more high frequency hearing losses.

Concurrent with recent research on non-compression amplification has been research on compression amplification. The simplest conventional hearing aid has three variables: frequency response, gain, and maximum power output. The optimum combination of these three variables for a hearing-impaired person has not been resolved, as has already been pointed out. When one introduces yet another set of variables, those of compression, the problem of specifying the optimum electroacoustic characteristics of the hearing aid becomes even more complex, and the predictability of intelligibility even more difficult.

Many experimental results have thus far failed to show the benefit of compression amplification. This seems surprising, considering the theoretical advantages of compression, and, in particular, of multiband compression, as pointed out by Villchur (1974). However, before 1970, compression amplifiers were of comparatively poor quality, often introducing noise and distortion. As pointed out by Lippmann (1975) researchers may have used too long an attack time, an improperly adjusted compression threshold, or may have failed to appropriately set other physical characteristics. Test materials may not have been sensitive enough to have sufficient resolving power. Interfering influences, such as background noise and reverberation, may have limited interpretation of results. In addition, properties of the amplifying system may have interacted, such as nonlinear distortion, thus not allowing the effect of individual parameters to be isolated and evaluated. And, perhaps most significantly, as mentioned above, two fundamental problems underlying much past research have been the problems of not providing an accurate specification of the true frequency-gain characteristics, and the lack of individualized selection strategy of the hearing aids. That is, it could be that the frequency-gain characteristics may not have been optimum for the compression conditions, therefore leading to the erroneous conclusion that compression is not beneficial.

Some of the basic questions asked with regard to compression amplification are: does this type of amplification improve discrimination, in addition to its other advantages, discussed above? Specifically, does it improve discrimination to a greater extent than a well-fitted non-compression hearing aid does? What characteristics of a compression hearing aid must be varied in order to provide optimum performance?

There is thus a need to study the effect of compression using high quality equipment, built to allow individual control of the various parameters of compression. Testing with this experimental hearing aid should attempt to determine to what extent the design objective of compression, plus post-compression equalization, tailored to the individual subject, and placing as much of the speech spectrum as possible into his residual area, can be beneficial to the hard of hearing. However, the simplest conventional hearing aid is still the most practical solution to amplification. Therefore, it is important to compare the more complex compression hearing aid, whether it is comprised of one or more channels, to a conventional hearing aid that has been determined to be well suited, if not optimum, for the subject's residual hearing. Matching the condition of compression to a superior non-compression hearing aid will allow one to judge the true relative merit of the former type of amplification. Some previous studies on compression amplification might have yielded substantially different results had they been compared to non-compression aids that were as well tailored to the subjects as were the compression aids. Villchur (1973), for example, did not use high frequency emphasis in his non-compression hearing aid (on the grounds that high frequency emphasis without compression might result in distress from environmental sounds). If one is to determine whether the complexities of the latter type of amplification are actually more beneficial than the relative simplicity of the former type of amplification, then they both must be optimally matched to the subject's residual hearing.

For these reasons, the present study was designed to investigate the potential value of both single- and multiband compression in improving intelligibility over a well-fitted non-compression hearing aid. As a start, it was felt that it was more important to compare a non-compression hearing aid with the simplest multiband compression system. It was believed that if multiband compression is beneficial, the largest improvement would come in going from a single- to a two-band system, and that it would therefore be preferable to investigate a two-band system in greater detail, than a more complex system superficially.

# CHAPTER III

## DESCRIPTION OF THE HEARING AID

### Compression Amplification

A conventional hearing aid amplifier is one in which the gain is independent of the signal being amplified, i.e. output = constant x input, or, expressed in dB,

output level (dB) = input level (dB) + gain (dB)

A compression amplifier functions such that a change in the level of the input signal modifies the system's gain. That is, the input-output ratio, or, the decibel change in output as a function of change in input, is not uniform over the entire range, i.e.,

output level(dB) = input level(dB) + gain(I)(dB)

where gain(I) = function of input. The parameters of compression amplification are defined below. Some of these parameters are graphically illustrated in Figure 2, which shows decibel changes in output level as a result of changes in input level.

### Definitions of Terms Related to Compression Amplification

#### Compression Ratio

The compression ratio is the decibel change in input level divided by the decibel change in output level. If, for example, a 20 dB input change results in an output

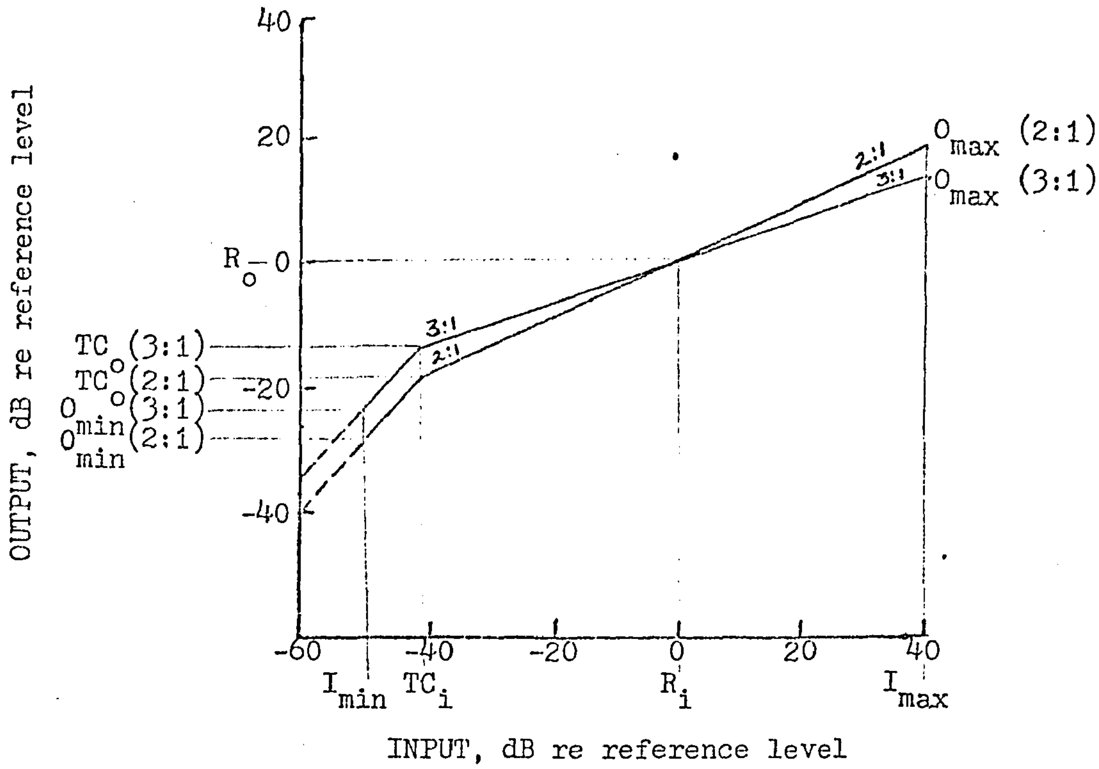


Fig. 2. Idealized input-output functions for a compression amplifier. Numbers on functions identify the respective compression ratios. Explanations of abbreviations used in figure are found in key, on following page.

Fig. 2--continued

KEY

| | |
|---|---|
| $I_{min}$ | Minimum input level (lowest level signal) detectable above internal noise. This intensity level is constant for all compression ratios for this particular circuit. |
| $O_{min}$ | Minimum output level detectable above internal noise. This level varies as a function of compression ratio, for this particular circuit. The numbers in parentheses identify the compression ratio. |
| $TC_i$ | Threshold of compression, at input. Constant for all compression ratios. |
| $TC_o$ | Threshold of compression, at output. Varies as a function of compression ratio, for this particular circuit. The numbers in parentheses identify the compression ratio. |
| $R_i$ | Reference level, at input. |
| $R_o$ | Reference level, at output. |
| $I_{max}$ | Maximum input level prior to overload of input circuit. Constant for all compression ratios, for this particular circuit. |
| $O_{max}$ | Maximum output level prior to overload of input circuit. Varies as a function of compression ratio, for this particular circuit. Other circuits could have their maximum determined by the output, making $O_{max}$ constant for all compression ratios. The numbers in parentheses identify the compression ratio. |

change of 5 dB, the compression ratio equals 20/5, or 4:1. If the input-output function has a slope of less than 1:1 (i.e. the angle θ in Fig. 2 is less than $45^{\circ}$), then compression is taking place. In a conventional amplifier, the ratio of input:output is 1:1 ($\theta = 45^{\circ}$), since the output level is equal to the input level plus a constant gain (in dB).

## Expansion

A change in the input level may result in an even greater change in the output level. If, for example, a 5 dB change in input results in 10 dB increase in output, then the ratio is 5/10, or 1:2. In this case, expansion is taking place ($\theta$ $45^{\circ}$).

## Compression Threshold

Compression threshold is that level at which compression begins. The compression threshold can be identified in two ways: the input level at which compression begins ($TC_i$ in Fig. 2), or the corresponding output level ($TC_o$ in Fig. 2). Below the compression threshold, the compression ratio is, theoretically, 1:1.

## Reference Level

In order to specify the characteristics of a compression hearing aid conveniently, the concept of reference level is introduced (Villchur, 1977). At a specific input level, regardless of any change in compression ratio, there will be no change in output level. This is seen graphically in Fig. 2 as the point where all input-output functions cross each other ($R_i = R_o$), and is known as the reference level (note that in Fig. 2, $R_i = R_o$ because the gain beyond the compressor has been arbitrarily set to 0 dB at the reference level). In the compression amplifier used in the present study, the voltage of the input signal at the reference level was 100 mv.

At all levels, the less intense the input, the greater is the relative gain. However, the gain of a signal below the reference level will be increased, whereas the gain of a signal above the reference level will be decreased. The above holds for a compression amplifier; for an expansion amplifier, the reverse is the case. This concept is discussed below, and is exemplified in Table 1.

If, at the input, the peak levels of the speech signal are set equal to the reference level (in this study, 100 mv.), there will be no amplification by the compressor (i.e. 0 dB gain for those peak levels). For those speech levels that fall below $R_i$, the gain will be greater than 0 dB. In addition, the closer the speech levels are to $R_i$, the closer the gain will be to 0 dB. Similarly, for speech levels above the reference level, the gain will be less than 0 dB, and the further the speech levels are from the reference level, the greater the attenuation (i.e., the gain becomes more negative). Table 1 gives numerical

Table 1. Numerical examples of the amount of gain obtained with a compression amplifier, for various inputs. Examples are given for two different compression ratios.

(a) Amount of gain obtained for various inputs, when the gain beyond the reference level is 0 dB ($R_o = R_i$), and the compression ratio is 2:1 and 3:1, respectively.

| Compression Ratio | Input, dB Relative to $R_i$ | Output, dB Relative to $R_o$ | Gain, dB (Output - Input) |
|---|---|---|---|
| 2:1 | -20 | -10 | 10 |
| 2:1 | -10 | -5 | 5 |
| 2:1 | 0 | 0 | 0 |
| 2:1 | 10 | 5 | -5 |
| 2:1 | 20 | 10 | -10 |
| 3:1 | -20 | -6.7 | 13.3 |
| 3:1 | -10 | -3.3 | 6.7 |
| 3:1 | 0 | 0 | 0 |
| 3:1 | 10 | 3.3 | -6.7 |
| 3:1 | 20 | 6.7 | -13.3 |

(b) Amount of gain obtained for various inputs, when the gain beyond the reference level is 10 dB ($R_o = R_i + 10$), and compression ratio is 2:1 and 3:1, respectively.

| Compression Ratio | Input, dB Relative to $R_i$ | Output, dB Relative to $R_o$ | Gain, dB (Output - Input) |
|---|---|---|---|
| 2:1 | -20 | 0 | 20 |
| 2:1 | -10 | 5 | 15 |
| 2:1 | 0 | 10 | 10 |
| 2:1 | 10 | 15 | 5 |
| 2:1 | 20 | 20 | 0 |
| 3:1 | -20 | 3.3 | 23.3 |
| 3:1 | -10 | 0 | 16.7 |
| 3:1 | 0 | 10 | 10 |
| 3:1 | 10 | 13.3 | 3.3 |
| 3:1 | 20 | 16.7 | -3.3 |

examples of these concepts. The gain of the total system can be set to the desired value, depending on the gain of the components other than the compressor. Table 1 provides examples of the input-output relations when the gain beyond the reference is 10 dB as well as 0 dB. As shown in the last column in the table, the gain decreases systematically with increasing input level. For an expansion system, the gain would decrease with an increase in input level.

## Maximum Output Level

All compressors have a limitation on the maximum output level. Typically, beyond a pre-determined input level, no additional output will be obtained despite further increases in the input level. However, additional output at any given input level can be obtained by post-compressor gain. This maximum output level ($0_{max}$ in Fig. 2) determines the Maximum Power Output (MPO) of the hearing aid (i.e. power = k(level)$^2$). The maximum output level may be measured by monitoring output on a voltmeter as a function of change in input. $0_{max}$ is that output level beyond which no additional output can be obtained despite increases in input level. The input level corresponding to $0_{max}$ is depicted in Fig. 2 as $I_{max}$.

## Dynamic Range

The dynamic range of the compressor at the input is the range from the minimum input level detectable above the internal noise ($I_{min}$ in Fig. 2) to the maximum input level prior to overload of the input circuit. The value obtained, $I_{max} - I_{min}$, specifies the range of lowest to highest input that the hearing aid can amplify. On some compressors, there may be an intermediate point, $I_{max} - \Delta$, beyond which further amplification is possible, but with an unacceptable level of distortion. If the compression threshold, $TC_i$, of the instrument is above the minimum detectable input, then the range of compression will be less than the total dynamic range of the instrument. The former range extends from compression threshold to $I_{max}$, i.e. $I_{max} - TC_i$, and specifies the range of lowest to highest input that the hearing aid will compress. In terms of output, the dynamic range of the compressor is $O_{max} - O_{min}$, and the dynamic range of compression is $O_{max} - TC_o$.

Time Constants

When there is an abrupt increase in input level, a finite amount of time is required for a compression amplifier to adjust to its new gain. The attack time is the time taken from the onset of the increase in signal level (by 25 dB) to that instant when the amplifier gain stabilizes to within 2 dB of the steady-state value. The release time, or recovery time, is the time elapsed from the decrease in signal level (by 25 dB) to the instant the amplifier gain stabilizes to within 2 dB of the steady-state value (ANSI, 1976). For convenience of measurement, attack and release times are specified in terms of the change in output signal level. A schematic diagram showing time constants is illustrated in Figure 3.

## The Experimental Hearing Aid

All experiments were carried out with a special-purpose compression amplifier. A sketch of the control panel is shown in Figure 4. As can be seen on the sketch, the compression system consisted of eight units. Unit 1 consisted of two pre-compression amplifiers: there were two potentiometers and two input jacks, potentiometer A corresponding to input A, and potentiometer B to input B. The range of pre-compression amplification was from 5 dB to 40 dB. The next four units (2 through 5) comprised the four individual channels of the multichannel compression system. Each panel had two plug-in modules for setting the filter response. The upper module corresponded to the low-cut filter, and the lower module corresponded to the high-cut filter. Below the plug-in modules was a control knob for the compression threshold. Beneath that was an input selector switch, corresponding either to input A or input B. The control for the compression ratio was situated at the bottom of the panel, with a range from 1:4 (expansion) to 10:1 (compression). Unit 6 was the frequency shaper, consisting of two switches, to

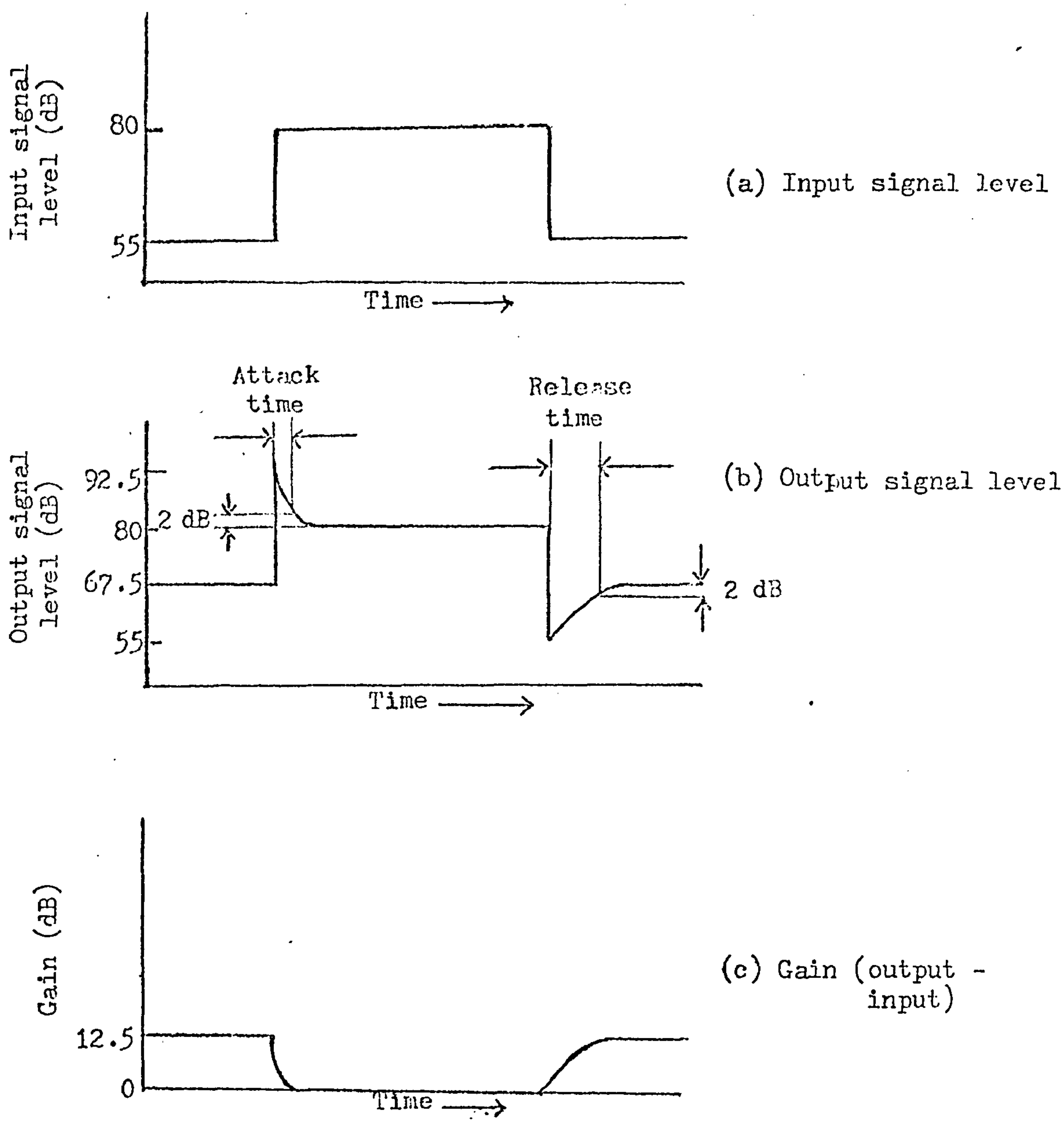


Fig. 3. Schematic diagram showing the effect of time constants on input to hearing aid, at a compression ratio of 2:1.

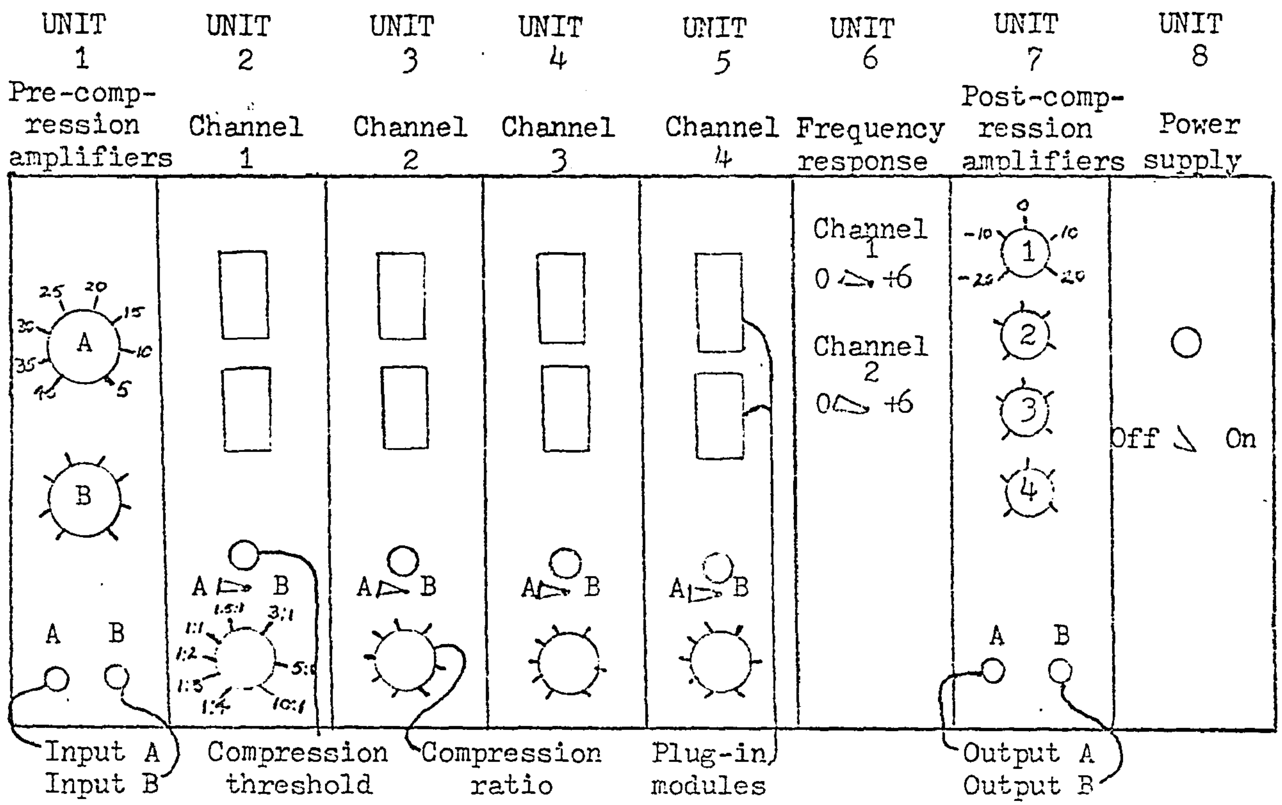


Fig. 4. Sketch of the control panel of the multichannel compression system. Shown are the eight units comprising the system. See text for full explanation of controls.

provide a 0 dB/octave slope or 6 dB/octave slope for channels 1 and 2, respectively. The four post-compression amplifiers, each having its own potentiometer, corresponded to each of the four channels, respectively. The range of post-compression amplification was from 20 dB to -20 dB. If the control knob was pulled in an outward position, it corresponded to input A, whereas an inward position of the knob corresponded to input B. There were two output jacks, again corresponding to either input A or B, respectively. Unit 8 was a power supply. The panel contained a power switch and a light to indicate when the system was on. Controls for the attack and release times were internal, and were held constant for each channel over all experimental conditions.

The amplification system was in a table-mounted box with outer dimensions of 17" x 9" x 5¼". The signals were transmitted via a tape recorder to the amplifier, and the amplified signals were delivered to the subject by means of TDH-39 earphones, mounted in MX/41-AR cushions. Block diagrams of the entire hearing aid, as used in the pilot study and the main experiment, are shown in Figures 17 and 24, respectively.

## Calibration of the Hearing Aid

The multichannel compression hearing aid was calibrated and checked by obtaining specific information, outlined below, and detailed in the following sections:

1. Gain of pre- and post-compression amplifiers;
2. Harmonic distortion;
3. Signal-to-noise (S/N) ratio;
4. Input-output curves at various compression ratios;
5. Minimum input level ($I_{min}$) and maximum output level ($O_{max}$);
6. Frequency response curves as a function of changes in low cutoff and high cutoff frequencies;
7. Attack and release time measurements;
8. Graphic level recordings of peak speech levels at the output of the hearing aid;
9. Spectrograms and vertical cross-sectional spectra of speech signals at the output of the hearing aid.

To facilitate calibration procedures, the compression amplification system was calibrated electrically, for all the relevant combinations of parameter values, followed by a final acoustic calibration of the overall system. Following are the details of the calibration procedures.

## Gain of Pre- and Post-Compression Amplifiers

Two pre-compression amplifiers controlled the overall input gain to all four channels, each amplifier corresponding to a separate input (Unit 1 in Fig. 4). The gain could be adjusted over the range of 0 to 40 dB. In addition, each channel had a post-compression amplifier that allowed the gain to be set individually for each respective channel

(Unit 7 in Fig. 4). The post-compression amplifiers could also be adjusted to each of two inputs, with the amount of post-compression gain ranging from -20 dB to 20 dB. This allowed for either attenuation or amplification of the frequencies passing through each channel.

To calibrate both the pre- and post-compression amplifiers, the frequency of the sinusoidal source was adjusted to 1000 Hz, and the setting of the gain control of each amplifier, respectively, was varied systematically. The output signal level was plotted against the input signal level for each compression ratio. All levels were specified relative to the reference level ($R_i$ in Fig. 2), which was 100 mv.

The effects of adjusting the pre-compression and post-compression gain, respectively, are best understood by referring to Table 2 and to Figure 5, the latter graphically illustrating the values depicted in Table 2. As illustrated, changing the pre-compression gain produced a horizontal shift of the input-output function of the overall compression system. Changing the post-compression gain shifted the input-output function of the entire amplification system vertically.

## Harmonic Distortion

Harmonic distortion was measured with input frequencies of 500 Hz, 800 Hz, and 1600 Hz. At a compression ratio of 3:1, the harmonic distortion was .22% for input

Table 2. Numerical examples of the effects of adjusting the pre-compression gain and the post-compression gain of the compression amplifier, at a compression ratio of 2:1.

| | A | | B | | C | | D |
|---|---|---|---|---|---|---|---|
| | Input to overall system, dB re reference level | Gain of pre-compressor amplifier, dB | Input to compressor, dB re reference level | Gain of compressor, dB, at compression ratio of 2:1 | Output of compressor, dB re reference level | Gain of post-compressor amplifier, dB | Output of overall system, dB re reference level |
| | -70 | 10 | -60 | 20 | -40 | 20 | -20 |
| | -60 | 10 | -50 | 20 | -30 | 20 | -10 |
| Threshold of Compression | -50 | 10 | -40 | 20 | -20 | 20 | 0 |
| | -40 | 10 | -30 | 15 | -15 | 20 | 5 |
| | -30 | 10 | -20 | 10 | -10 | 20 | 10 |
| | -20 | 10 | -10 | 5 | -5 | 20 | 15 |
| Reference Level | -10 | 10 | 0 | 0 | 0 | 20 | 20 |
| | 0 | 10 | 10 | -5 | 5 | 20 | 25 |
| | 10 | 10 | 20 | -10 | 10 | 20 | 30 |

C vs. B = Input-output of compressor

C vs. A = Input-output of compressor + pre-compressor amplifier

D vs. B = Input-output of compressor + post-compressor amplifier

(Note: C vs. B is the same as D vs. A if gain of pre-compressor amplifier = 0 dB, and gain of post-compressor amplifier = 0 dB)

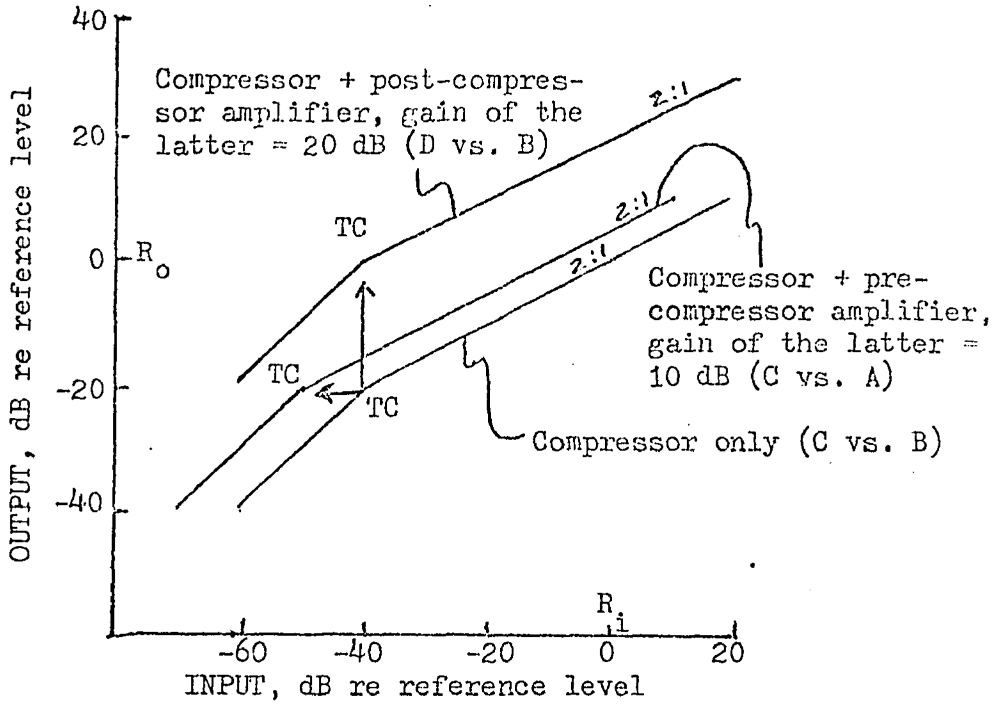


Fig. 5. Idealized input-output functions illustrating effects of adjusting pre-compression gain and post-compression gain, at a compression ratio of 2:1. Numerical values of A through D are derived from Table 2, columns A through D, respectively. Numbers on functions refer to the compression ratio. Arrows indicate the direction of shift of the input-output function, with the respective changes. Explanations of abbreviations used in figure are found in key, on following page.

Fig. 5--continued

KEY

| | |
|---|---|
| TC | Threshold of compression |
| A | Input to overall system, dB re reference level |
| B | Input to compressor, dB re reference level |
| C | Output of compressor, dB re reference level |
| D | Output of overall system, dB re reference level |

(Note: Numerical values of A through D are derived from Table 2, columns A through D, respectively.)

frequency of 500 Hz, .155% for input frequency of 800 Hz, and .09% for input frequency of 1600 Hz.

Signal-to-Noise (S/N) Ratio

The S/N ratio of the instrument was measured at a compression ratio of 1:1, with a probe frequency of 1000 Hz. The pre-compression and post-compression amplifiers were both turned to the maximum possible gain, and the output voltage was measured. The probe tone was then introduced at a level sufficient to produce $O_{max}$, and the noise generated internally by the instrument was measured with the input short-circuited. The difference between the two values, the maximum possible output minus the noise floor, i.e. $O_{max} - O_{min}$, was defined as the S/N ratio of the instrument. Table 3 gives the values of these measurements.

Input-Output Functions as Various Compression Ratios

The compression amplifier used in the present study had a continuously variable control for changing compression ratio over the range from compression ratios of 1:4 (expansion) to 10:1 (compression). Compression ratio was calibrated using a pure tone of 1000 Hz, and measuring the output voltage on a voltmeter, over the range from the minimum input level, $I_{min}$, to the overload level, $I_{max}$. A block diagram of the equipment used in measuring compression ratio is illustrated in Figure 6.

The input-output curves of the three ratios utilized in the present study, 1:1, 1.5:1, and 3:1, are illustrated

Table 3. Signal-to-noise (S/N) ratio of the compression amplifier, at a compression ratio of 1:1, for each of four channels.

| Channel | 1000 Hz tone output, dBm,* at maximum setting of the compression amplifier | Noise floor, dBm | S/N ratio, dB |
|---|---|---|---|
| 1 | 17.2 | -59 | 76.2 |
| 2 | 16.5 | -62 | 78.5 |
| 3 | 18.5 | -60 | 78.5 |
| 4 | 18.2 | -60 | 78.2 |

*dBm employs as its reference, i.e. 0 dBm, 1 milliwatt of power in a 600Ω circuit.

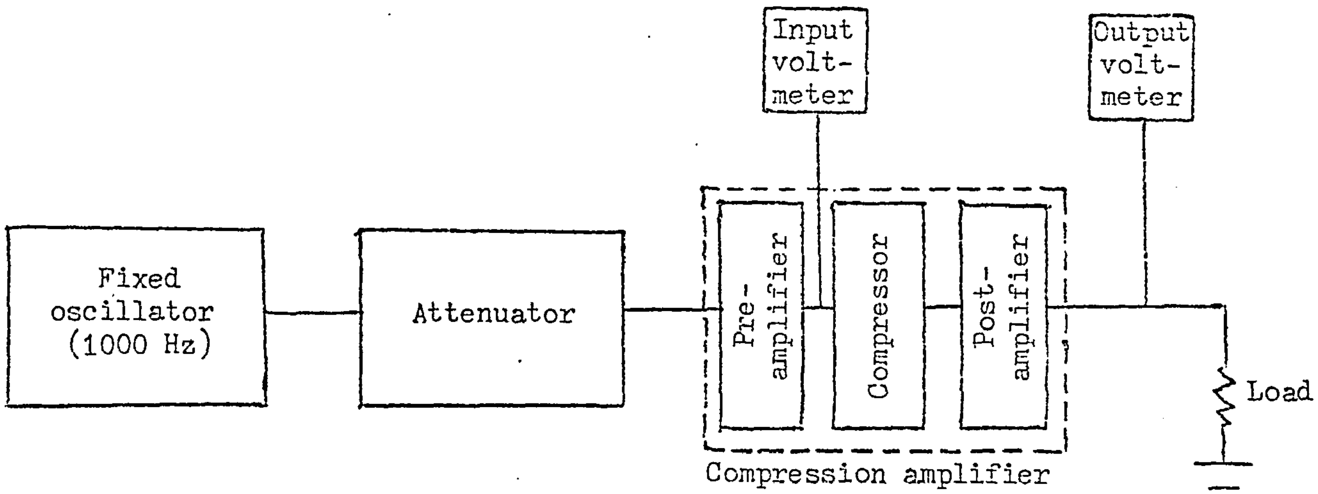


Fig. 6. Block diagram of the equipment used in calibrating compression ratio of the compression amplifier.

in Figure 7. This figure plots decibels of input change versus decibels of output change. The intensity values are specified in dB re the reference level, $R_i$, which was 100 mv. Both an input and an output of 100 mv. are therefore equivalent to 0 dB, since the gain of the compressor is 0 dB at the reference level.

At inputs less than approximately -40 dB re the reference level, the compression ratios are greater than they are above -40 dB. Thus, this input level is taken to be the compression threshold. Although, theoretically, below the threshold of compression, TC, the compression ratio should be 1:1, the compression ratio was greater than this because of internal noise.

<u>Minimum Input Level and Maximum Output Level</u>

The lowest input level, $I_{min}$, of the hearing aid was -60 dB re the reference level (see Fig. 7). Because of the internal electrical noise of the system at low levels, as noted above, output level increased as input level decreased to less than -60 dB. The upper limit, $I_{max}$, of the hearing aid was 40 dB re the reference level. Above $I_{max}$ there was virtually no change in output despite any amount of change in input. This level was the maximum output level, $O_{max}$.

The input dynamic range was from -60 dB to 40 dB, re the reference level, i.e.

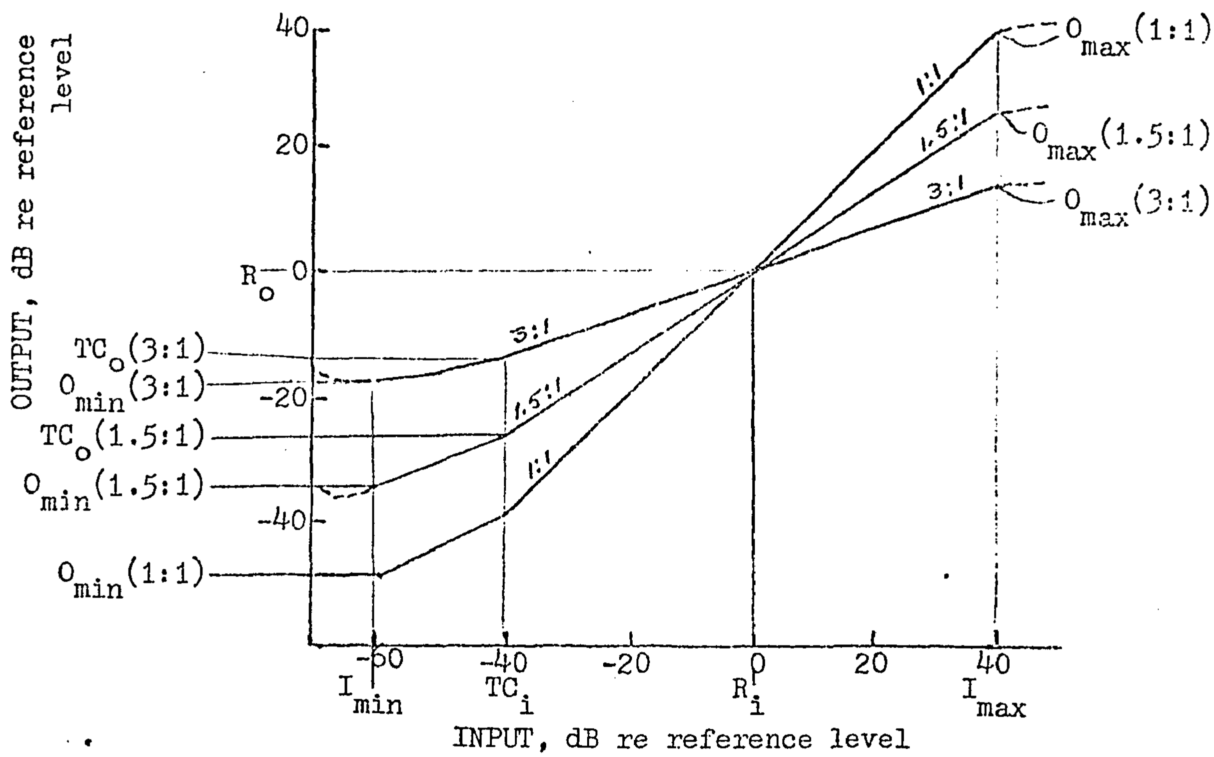


Fig. 7. Measured input-output functions for various compression ratios of the compression amplifier. Numbers on functions identify the respective compression ratios. Explanations of abbreviations used in figure are found in key, on following page.

Fig. 7--continued

KEY

| | |
|---|---|
| $I_{min}$ | Minimum input level detectable above internal noise. |
| $O_{min}$ | Minimum output level detectable above internal noise. Varies as a function of compression ratio. Number in parentheses identify the compression ratio. |
| $TC_i$ | Threshold of compression, at input. |
| $TC_o$ | Threshold of compression, at output. Varies as a function of compression ratio. The numbers in parentheses identify the compression ratio. |
| $R_i$ | Reference level, at input. Equivalent to 100 mv. |
| $R_o$ | Reference level, at output. Equivalent to 100 mv. |
| $I_{max}$ | Maximum input level prior to overload of input circuit. Virtually no change in output with change in input above this level. |
| $O_{max}$ | Maximum output level prior to overload of input circuit. Varies as a function of compression ratio. Numbers in parentheses identify the compression ratio. |

$$I_{max} - I_{min} = 40 - (-60) \text{ dB}$$
$$= 100 \text{ dB.}$$

The dynamic range of compression, with the compression threshold set as illustrated in Fig. 7, which was the setting employed in the present study, was from an input of -40 dB to 40 dB, re the reference level, i.e.

$$I_{max} - TC_i = 40 - (-40) \text{ dB}$$
$$= 80 \text{ dB.}$$

A change in the compression threshold will make the dynamic range of compression either greater or smaller. Input-output functions of the compression amplifier at compression thresholds other than those used in the present study can be found in Appendix A.

Frequency Response Curves

There were several factors controlling frequency response. These were as follows: Each channel had a low cutoff frequency and a high cutoff frequency, whose respective values could be changed by plug-in filter modules (found on Units 2 through 5, as shown in Fig. 4). Channels 1 and 2 also had a switch that allowed the addition of a 6 dB/octave change in the slope of the frequency response (found on Unit 6, as shown in Fig. 4). In addition, the post-compression gain could be controlled individually for each of the four channels (Unit 7, as shown in Fig. 4).

The frequency response of the compression amplifier was checked using swept sine wave analysis, recorded on a graphic level recorder, and by obtaining real time spectral analysis using a white noise input. In the former method, the 1000 Hz probe tone input to the hearing aid was set at the reference level of 100 mv. The electrical output of the compression amplifier was coupled to a graphic level recorder, set at a writing speed of 100 mm/second, and at a paper speed of 10 mm/second. Figure 8 illustrates an example of the frequency response at a ratio of 1:1 (non-compression), for each of the two channels used in the study, measured individually, and the resultant frequency response when the output was recorded through both channels combined. The relatively smooth overall frequency response was obtained by means of fine adjustment to the phase shifters[3] between channels 1 and 2.

Measurements of the frequency response of the compression amplification system at compression ratios other than 1:1, and for input levels above and below the reference level, are given in Appendix B.

The second method by which the frequency response of the compression amplifier was checked was that of a real time spectral analysis, using white noise as the input. In this method, white noise was delivered to an impedance

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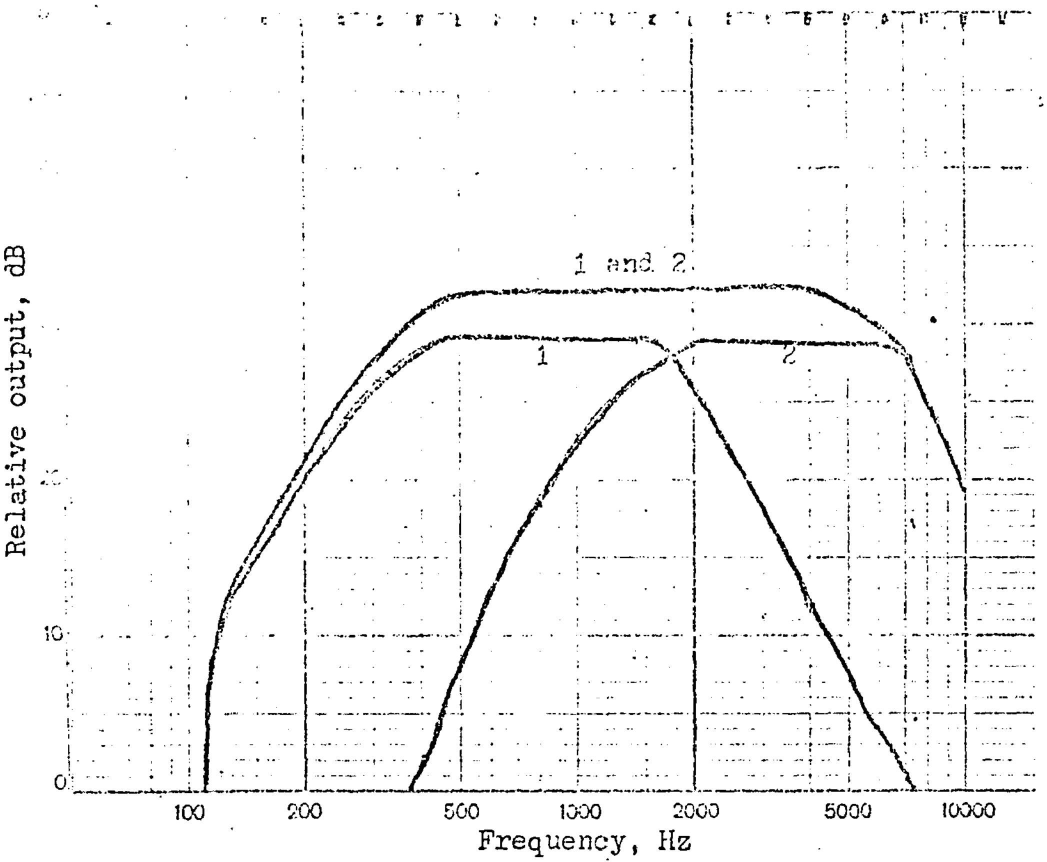
[3] A phase shifter was inserted to smooth out the frequency response in the region of the crossover frequency.




| Channel | Frequency, Hz Low cutoff | High cutoff |
|---|---|---|
| 1 | 250 | 1500 |
| 2 | 1500 | 6000 |
| 1 and 2 | 250 | 6000 |

Fig. 8. Graphic level recording of frequency responses of channels 1 and 2, individually and in combination. Input was the reference level, 100 mv. Compression ratio was 1:1. Numbers on curves indicate channels.

splitter, whose purpose was to maintain the proper input impedances to the two passive filters that followed. Since passive filters were used, impedance matching was important for proper operation of the filters. The output of a noise generator with a flat noise power spectrum was split into two bands by means of the filters, attenuated as desired, mixed, and fed into the compression amplifier. The measurement system thus replicated bandwidths of the channels of the compression amplifier. This was done to check the functioning of each channel of the amplifier, although it was not essential for its calibration. Frequency response was recorded on an X-Y plotter coupled to a real time spectrum analyzer. Figure 9 shows a block diagram of the equipment used for the real time analysis method of measuring frequency response.

Figure 10 depicts an example of real time analysis of white noise, with the compression amplifier set at a ratio of 1:1 and a slope of 0 dB/octave. The frequency range for channel 1 was 125 Hz to 1000 Hz, and for channel 2, 1000 Hz to 6000 Hz. Figure 11 depicts the real time spectrum analysis with the inclusion of the 6 dB/octave slope to channels 1 and 2. Other variables remained the same.

For additional frequency response curves of other experimental conditions, see Appendices C and D.

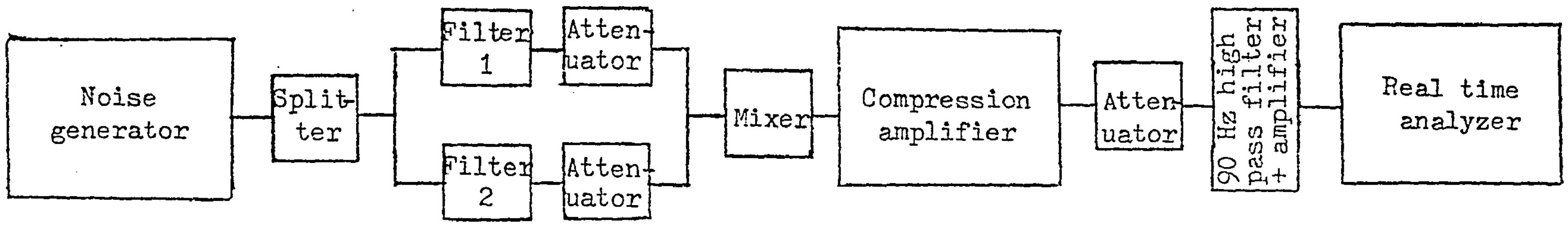



Fig. 9. Block diagram of equipment used for real time spectrum analysis method of measuring frequency response.

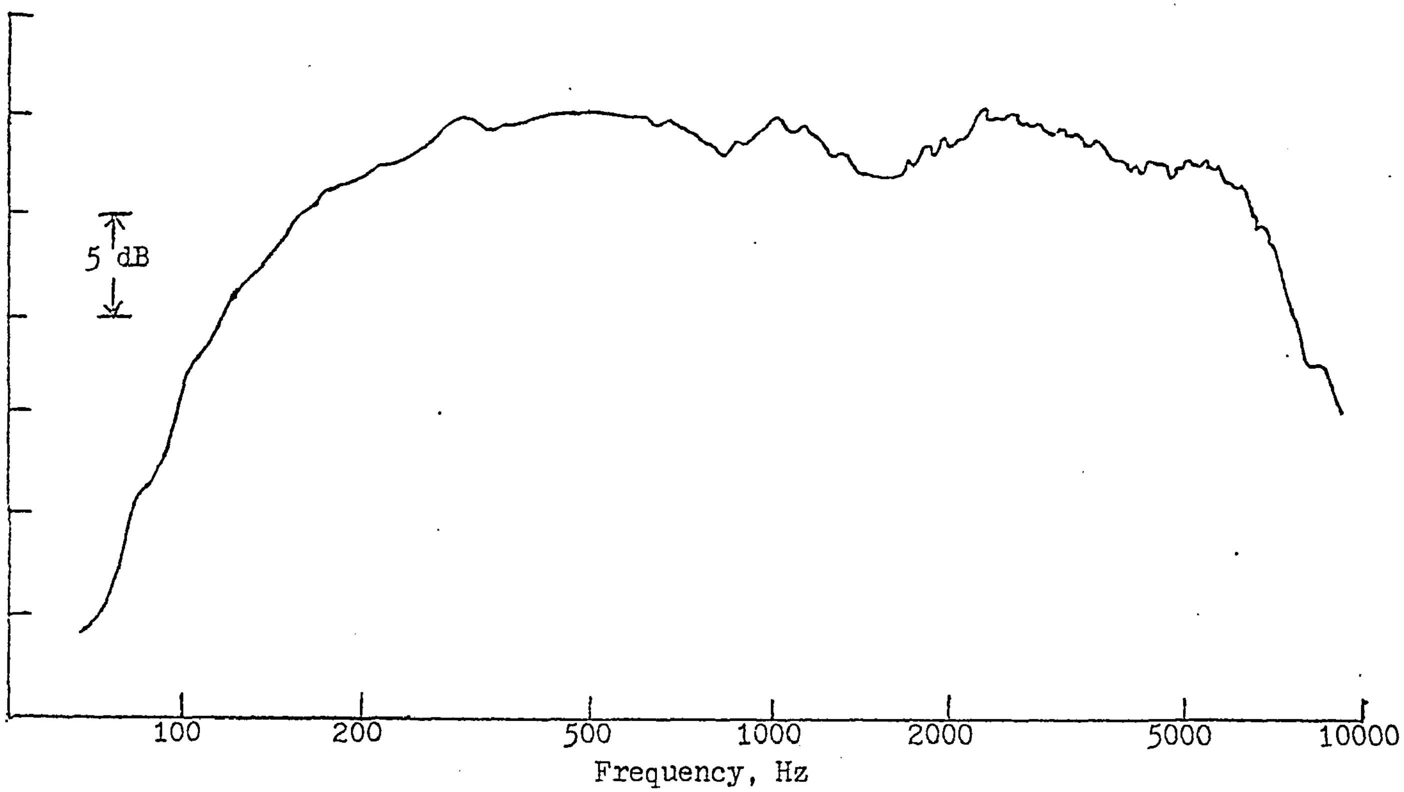



Fig. 10. Real time spectrum analysis: flat frequency response. White noise was passed through channels 1 and 2 combined. Channel 1 was set to a bandwidth of 125 Hz to 1000 Hz, and channel 2 was set to a bandwidth of 1000 Hz to 6000 Hz. Compression ratio was 1:1.


Fig. 11. Real time spectrum analysis: frequency response of +6 dB/octave. White noise was passed through channels 1 and 2 combined. Channel 1 was set to a bandwidth of 125 Hz to 1000 Hz, and channel 2 was set to a bandwidth of 1000 Hz to 6000 Hz. Compression ratio was 1:1.

Time Constants

Using pure tone input signals of 500, 1000 and 2000 Hz, which alternated periodically and abruptly between levels of 55 and 80 dB SPL, the attack and release times were observed on an oscilloscope and determined from an oscillographic tracing. The values of 55 dB and 80 dB approximately encompass the range of sound pressure level during a conversation.

The attack time is defined as the time interval between the abrupt increase from 55 dB to the level where stabilization to within 2 dB of the steady state value for the 80 dB input level has occurred. The release time is defined as the time interval between the abrupt drop from 80 dB to the level where stabilization to within 2 dB of the steady state value for the 55 dB input has occurred (ANSI, 1976). Recording the output signal on an oscillographic tracing allowed the attack and release times to be measured with a precision of $\pm 1$ msec., using a paper speed of 40 inches/sec. The tracings produced showed the signal envelope in linear units (as opposed to decibels). From this diagram, the peak-to-peak envelope could be read easily. A change of 2 dB corresponds to a 26% increase in peak-to-peak voltage, i.e. from 100% to 126%. A change of -2 dB corresponds to a 21% reduction in peak-to-peak voltage, i.e. from 100% to 79%. Figure 12 shows a block dia-

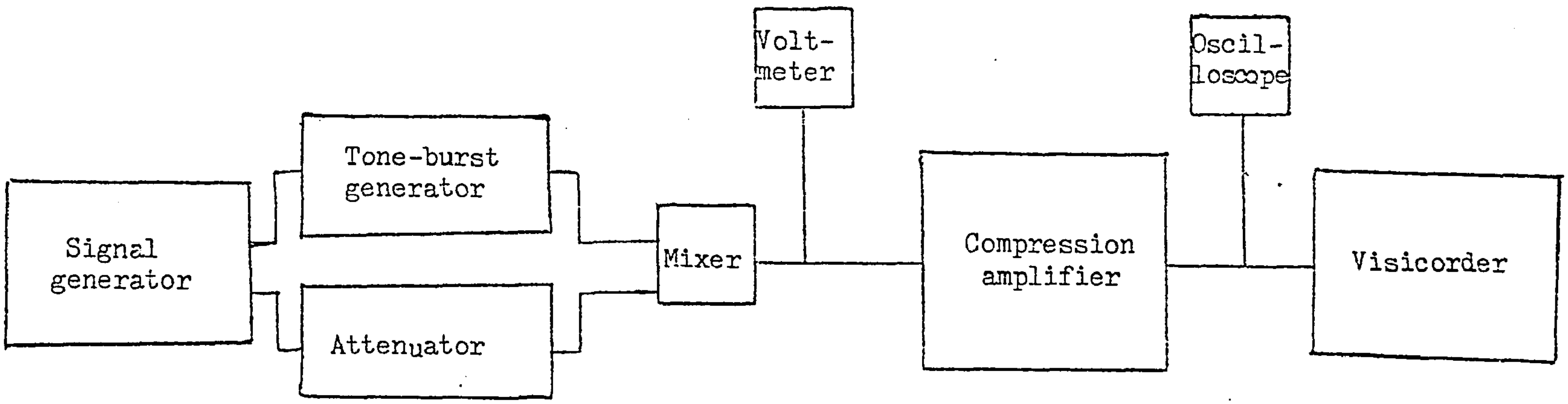



Fig. 12. Block diagram of the apparatus used for measurement of attack and release times.

gram of the apparatus used in making these measurements.[4] Figures 13 and 14 show the envelope of the output signal as recorded with the oscillographic tracing, illustrating attack and release times, respectively, at 2000 Hz.

The measured attack time of all four channels, at the three frequencies measured, ranged between 1.5 and 3.75 msec., for a nominal setting of 2 msec. Measured release times ranged from 16.0 to 21.75 msec., for a nominal setting of 20 msec. All measures were made at both 1.5:1 and 3:1 compression ratios. Table 4 specifies the values for attack and release times at three frequencies, and for two compression ratios, in all four channels.

Graphic Level Recordings with Speech as the Stimulus

Graphic level recordings were made of the Nonsense Syllable Test, which was used to measure speech discrimination in the study. This measure was made to check the dynamic range of the test items, which was estimated to be about 25 dB.

Spectrograms and Cross-Sections of Speech

Spectrographic analyses, including cross-sections of individual phonemes under various experimental conditions of compression, were made. This was done in order to ensure

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[4] Note that an attenuator is used in parallel with the tone-burst generator to determine the level of the output signal. That is, the signal going through the attenuator determines the steady background voltage, $v_1$. The sum of the output of the attenuator and the tone-burst generator determines the new steady state value, $v_4$.

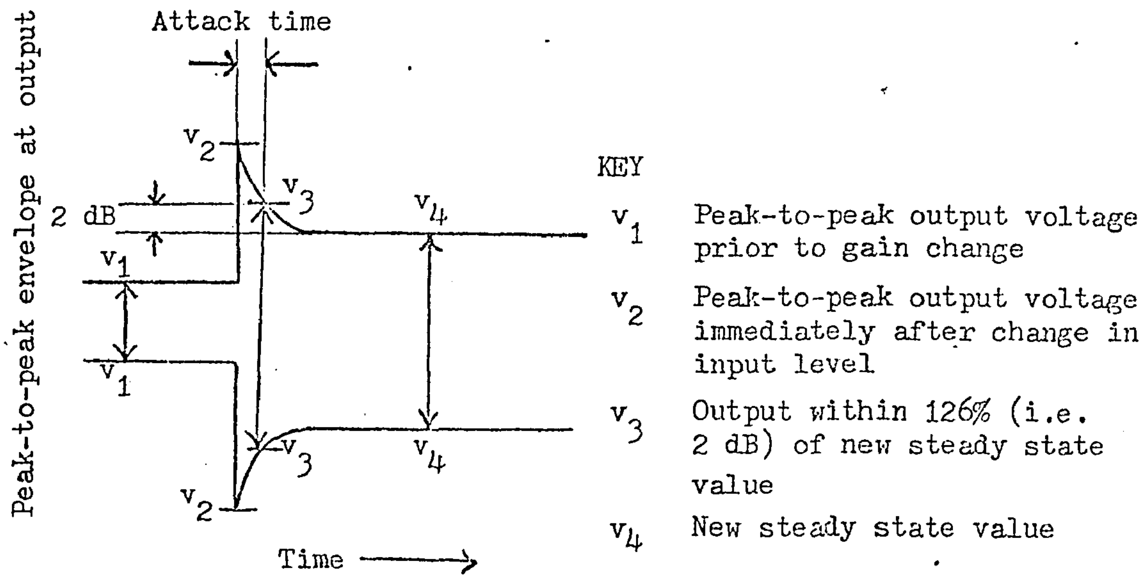
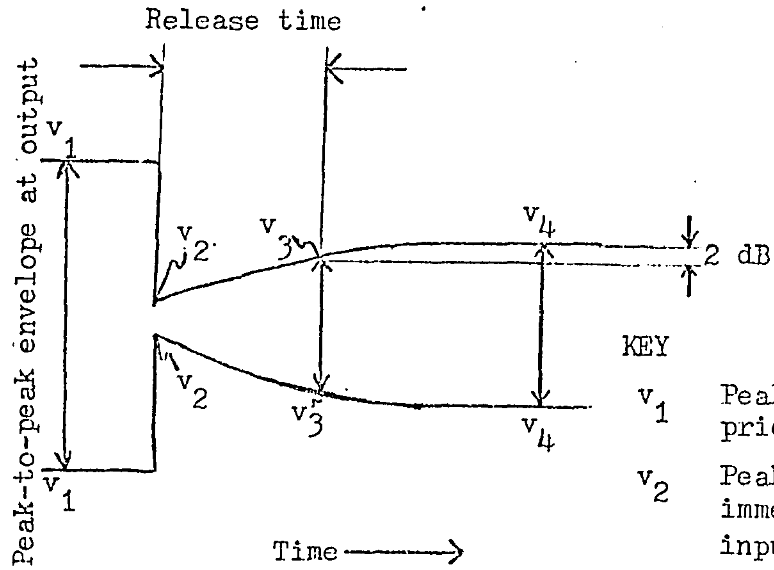



Fig. 13. Measured envelope of a typical transient response of the compression amplifier. An abrupt increase of 25 dB in the input sound pressure level caused the output to change rapidly upwards, compression set in, and the output then gradually approached the corresponding steady-state voltage, measured to within 2 dB, i.e. 126%, of final stabilization. The frequency shown is 2000 Hz, and the compression ratio 3:1.




KEY

$v_1$ Peak-to-peak output voltage prior to gain change

$v_2$ Peak-to-peak output voltage immediately after change in input level

$v_3$ Output within 79% (i.e. -2 dB) of new steady state value

$v_4$ New steady state value

Fig. 14. Measured envelope of a typical transient response of the compression amplifier. An abrupt decrease of 25 dB in the input sound pressure level caused the output to change rapidly in the opposite direction, compression was released, and the output then gradually approached the corresponding steady state voltage, measured to within -2 dB, i.e. 79%, of final stabilization. The frequency shown is 2000 Hz, and the compression ratio 3:1.

Table 4. Attack and release times of the multichannel compression amplifier by channel, in milliseconds, for each of two compression ratios, at three frequencies.

| Channel | Compression ratio | | Frequency, Hz 500 | 1000 | 2000 |
|---|---|---|---|---|---|
| Channel 1 | 1.5:1 | Attack time/ Release time | 2.50/19.75 | 1.50/20.00 | 1.75/18.00 |
| | 3:1 | Attack time/ Release time | 3.75/24.75 | 3.00/24.00 | 3.25/25.75 |
| Channel 2 | 1.5:1 | Attack time/ Release time | 3.00/16.25 | 2.00/20.25 | 1.75/19.00 |
| | 3:1 | Attack time/ Release time | 3.50/20.75 | 3.75/24.00 | 2.75/26.50 |
| Channel 3 | 1.5:1 | Attack time/ Release time | 1.50/16.50 | 1.50/16.00 | 1.50/16.00 |
| | 3:1 | Attack time/ Release time | 2.00/26.75 | 2.00/22.50 | 2.00/27.75 |
| Channel 4 | 1.5:1 | Attack time/ Release time | 2.75/17.50 | 2.00/21.75 | 2.00/21.00 |
| | 3:1 | Attack time/ Release time | 3.50/26.50 | 3.50/27.00 | 3.50/25.50 |

that, after all the above calibrations were made, the multichannel compression amplifier actually processed speech as it had been designed to.

# CHAPTER IV

# THE PILOT STUDY

## Description of the Subjects

A group of hearing aid wearers with binaural sensori-neural hearing losses which had been medically diagnosed, in addition to having been confirmed by audiologic evaluations, were initially selected as potential subjects. All hearing losses had been acquired post-lingually, and all subjects had been wearing compression hearing aids that had been prescribed by audiologists. The purpose of using experienced hearing aid wearers was to avoid confounding the results due to improvements obtained from a general adaptation to a hearing aid, rather than from improvements attributable to the various settings of the experimental hearing aid.

The group selected from the pool of potential subjects consisted of two females and three males, aged 25 to 66 years (mean age, 43 years). These subjects were selected on the basis of audiologic testing performed during the first session. The criteria used for both subject and ear selection are given below. All baseline testing was conducted in a two-room IAC booth using a Grason-Stadler 1701 audiometer, TDH-39 earphones set in MX/41-AR cushions, and a Madsen Z070 electroacoustic bridge. Equipment was calibrated to ANSI 1969 standards.

Both audiometric and impedance measures were taken as part of the initial baseline measurements. The audiometric measures, made under earphones (all testing was done for both ears, with masking where indicated), included the following:

1. Pure tone air-conduction thresholds at one-octave intervals (or, where indicated by a sharp drop in the audiogram, at half-octave intervals), from 125 Hz through 8000 Hz. Thresholds were obtained using the Hughson-Westlake technique (Carhart and Jerger, 1959).

2. Pure tone bone-conduction thresholds at one-octave intervals (or, where indicated, at half-octave intervals), from 250 Hz through 4000 Hz. Thresholds were obtained using the Hughson-Westlake technique (Carhart and Jerger, 1959).

3. Speech Reception Threshold (SRT), using taped C.I.D. Auditory Test W-1.

4. Word Discrimination Score (WDS), using taped C.I.D. Auditory Test W-22. The test was administered at SRT + 35 dB, or lower, the latter only if the setting of SRT +35 dB was judged by the subject to be too loud.

5. Most Comfortable Loudness (MCL), measured with taped W-22 word lists. A simple up-down adaptive procedure (Levitt, 1971), using 5 dB steps, starting at an intensity of 5 dB above SRT, was utilized to measure MCL. Instructions to the subject were adapted from Morgan, et al.

(1974), and were as follows:

> The purpose of this test is to find and maintain a loudness at which words are most comfortable to listen. We want you to decide when the sound or the phrase is at a level which you feel is the most comfortable listening level.
>
> Say "yes" when the sound is at your most comfortable listening level. Say "no" when the sound is louder or softer than the most comfortable level. Following each phrase, you must respond either "yes" or "no".

6. Loudness Discomfort Level (LDL), for speech, using taped W-22 word lists; for white noise bursts; and for bursts of narrow bands of noise from 250 Hz through 8000 Hz, at one-octave intervals. An ascending procedure, using 5 dB steps, with initial presentation at MCL, was used to measure LDL. Instructions to the subjects were as follows (adapted from Morgan, et al., 1974):

> This is a test in which you will be hearing sounds. We want you to decide when the sound is at a level that you think is uncomfortably loud or unpleasantly loud. By "uncomfortably loud or unpleasantly loud" we mean when the sound is so loud that you would choose not to listen to it for any period of time.
>
> Say "yes" when the sound is at a loudness to which you would not choose to listen. Say "no" when the sound is below that level. Following each sound you must respond either "yes" or "no".

Following audiometric measures, impedance measures were taken. These included tympanometry at 220 Hz, static compliance (cc), and contralateral stapedial reflexes for pure tone activating signals of 250, 500, 1000, 2000, and 4000 Hz, and for white noise.

From the initial group of subjects, tested as described above, the criteria for both subject selection and ear selection for the pilot study (and later, for the main experiment) were as follows (although all subjects were binaurally impaired, only one ear was tested in the study):

The type of hearing loss was sensorineural, as indicated by air-conduction and bone-conduction thresholds equal to within 10 dB. Impedance measurements were obtained as part of the baseline audiological measurements, to identify any active middle ear pathology. All impedance data correlated with pure tone audiometric data, that is, not suggestive of any middle ear pathology.

The amount of hearing loss was at least 45 dB HTL in the speech frequencies, as measured by the SRT. To avoid participation of the non-test ear by the possibility of crossover of the test stimuli, the difference in SRT, or threshold at any frequency, between the two ears did not exceed 35 dB, if the test ear had the greater impairment. The Word Discrimination Score (WDS) was less than 80%.

Acoustic reflexes for at least two frequencies were less than 60 dB SL, suggestive of recruitment (Schwartz and Bess, 1975). There was also a small range from MCL to LDL of 20 dB or less, also suggestive of recruitment (Hood and Poole, 1966) (all but one subject met this last criterion, with his range being 27 dB).

Table 5 summarizes the air-conduction thresholds of the subjects for the test ear (that is, the experimental ear), and Table 6 summarizes this information for the non-test ear. The standard deviations and the lowest to highest thresholds are also given. Figure 15 plots, in audiogram form, the values given in Table 5. The average audiogram for the test ear was mildly downward sloping. Figure 16 plots, in audiogram form, the values given in Table 6. The average audiogram for the non-test ear approximated that of the test ear.

Tables 7 and 8 summarize the results of speech tests performed on the five subjects, for the test ear and the non-test ear, respectively. Mean Pure Tone Averages, Speech Reception Thresholds, and Word Discrimination Scores are given for each ear, in addition to Most Comfortable Loudness Levels and Loudness Discomfort Levels for speech stimuli. The standard deviations and the lowest to highest values are also shown.

## Procedure

The first testing session for all subjects involved a diagnostic evaluation, detailed above. Subjects then reported for four additional two-hour sessions. During the first of these subsequent sessions, air-conduction thresholds at 250 Hz through 4000 Hz and Speech Reception Thresholds were remeasured to determine the stability of the subjects' losses. All measures were within 5 dB of the values

Table 5. Mean pure tone thresholds for the test ear, for five subjects. Standard deviations and lowest to highest thresholds are also included.

| | Frequency, Hz | | | | | | |
|---|---|---|---|---|---|---|---|
| | 125 | 250 | 500 | 1000 | 2000 | 4000 | 8000 |
| Mean threshold, dB HTL | 35.0 | 41.0 | 59.0 | 67.0 | 70.0 | 69.0 | 63.75* |
| Standard deviation, dB | 18.7 | 18.5 | 8.2 | 11.5 | 18.0 | 16.3 | 23.2 |
| Lowest - highest thresholds, dB HTL | 10-60 | 20-65 | 50-70 | 50-75 | 50-95 | 55-95 | 40-NR** |

*average of four ears only (one ear elicited no response at the maximum output of the audiometer)

**no response at the maximum output of the audiometer

Table 6. Mean pure tone thresholds for the non-test ear, for five subjects. Standard deviations and lowest to highest thresholds are also included.

| | Frequency, Hz | | | | | | |
|---|---|---|---|---|---|---|---|
| | 125 | 250 | 500 | 1000 | 2000 | 4000 | 8000 |
| Mean threshold, dB HTL | 26.25* | 38.0 | 56.0 | 65.0 | 70.0 | 66.25* | 60.0* |
| Standard deviation, dB | 21.4 | 27.8 | 11.9 | 12.3 | 17.7 | 19.3 | 26.1 |
| Lowest - highest thresholds, dB HTL | 10-NR** | 15-75 | 40-70 | 50-80 | 40-85 | 45-NR** | 25-NR** |

*average of four ears only (one ear elicited no response at the maximum output of the audiometer)

**no response at the maximum output of the audiometer

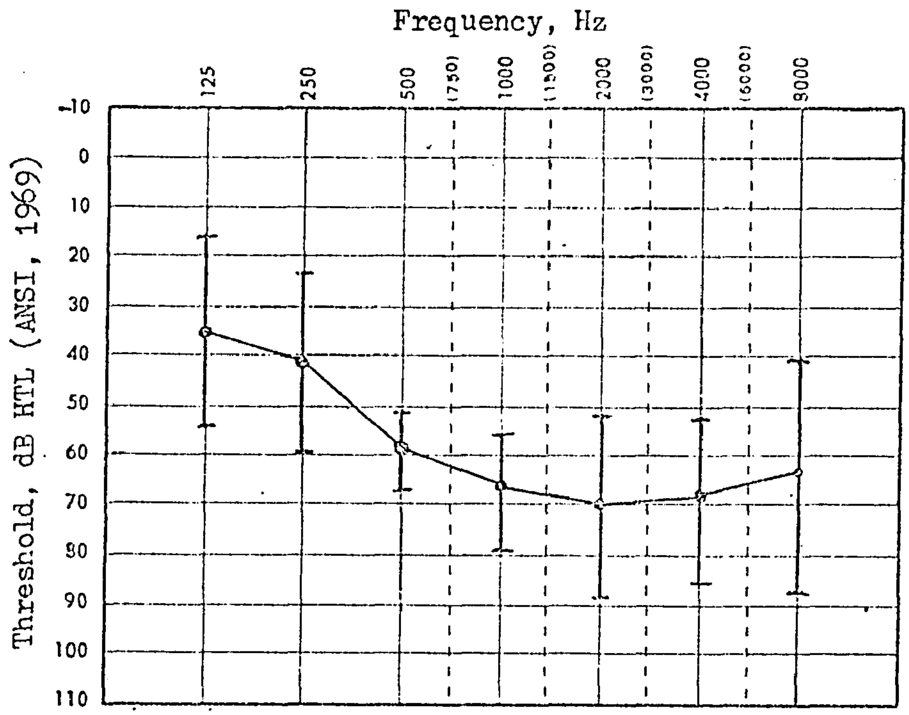



Fig. 15. Mean pure tone thresholds of the test ear, for five subjects. Bars indicate standard deviations.

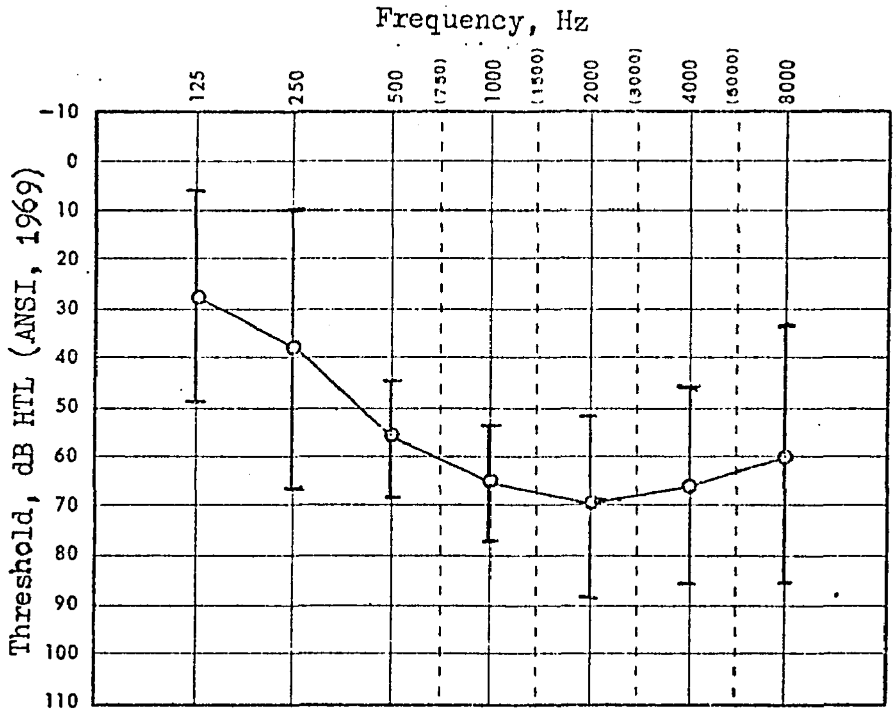



Fig. 16. Mean pure tone thresholds of the non-test ear, for five subjects. Bars indicate standard deviations.

Table 7. Results of speech audiometric tests of the test ear, for five subjects. Pure Tone Average, means, standard deviations, and lowest to highest values are also given.

| Audiological measures | Mean | Standard deviation | Lowest--highest values |
|---|---|---|---|
| Pure Tone Average (PTA),* dB HTL | 62.5 | 7.7 | 52-73 |
| Speech Reception Threshold (SRT), dB HTL | 53.0** | 5.7 | 45-60 |
| Word Discrimination Score (WDS), percent | 69.6 | 6.4 | 62-78 |
| Most Comfortable Loudness (MCL), dB HTL | 72.4 | 5.1 | 67-80 |
| Loudness Discomfort Level (LDL), dB HTL | 90.4 | 9.7 | 75-100 |
| Dynamic range (SRT - LDL), dB | 37.4 | 11.9 | 20-47 |
| Range of MCL to LDL, dB | 18.0 | 8.6 | 8-27 |

*For two ears with sharply downward sloping hearing loss, two-frequency PTA was used, instead of standard three-frequency PTA.

**Because two of the hearing losses were so sharply downward sloping, mean SRT did not agree closely with mean PTA, even when two-frequency PTA was used.

Table 8. Results of speech audiometric tests of the non-test ear, for five subjects. Pure Tone Average, means, standard deviations, and lowest to highest values are also given.

| Audiological measures | Mean | Standard deviation | Lowest-highest values |
|---|---|---|---|
| Pure Tone Average (PTA),* dB HTL | 61.4 | 12.9 | 43-77 |
| Speech Reception Threshold (SRT), dB HTL | 50.0** | 12.9 | 35-65 |
| Word Discrimination Score (WDS), percent | 70.5** | 8.2 | 60-80 |
| Most Comfortable Loudness (MCL), dB HTL | 69.0 | 7.4 | 60-80 |
| Loudness Discomfort Level (LDL), dB HTL | 88.4 | 8.1 | 80-97 |
| Dynamic range (SRT - LDL), dB*** | 35.0 | 15.5 | 13-55 |
| Range of MCL to LDL, dB | 19.4 | 8.6 | 10-30 |

*For two ears with sharply downward sloping hearing loss, two-frequency PTA was used, instead of standard three-frequency PTA.

**Average of four ears only: one subject had such poor discrimination, that even an SRT could not be obtained.

***For subject for whom SRT could not be obtained, range used was from PTA to LDL.

obtained in the diagnostic evaluation.

At the start of each test session, the Loudness Discomfort Level (LDL) for each subject for speech, using taped W-22 word lists, was measured under the following conditions (the procedure used to measure LDL is decribed above):

| | | |
|---|---|---|
| Frequency response | 1. 0 dB/octave slope | |
| | 2. 6 dB/octave slope | |
| Frequency range | 1. 125 - 6000 Hz ($L_b$) | |
| | 2. 125 - 1000 Hz ($L_1$) | Note that the cross-over frequency was 1000 Hz |
| | 3. 1000 - 6000 Hz ($L_2$) | |

Separate LDL measurements were obtained for each of the 2 x 3, i.e. 6, combinations of the above two factors. The measurements obtained for each of the five subjects are given in Appendix E. These measures were used in determining the level of test administration for all conditions. Since six LDL measurements were obtained, the specific LDL used in any experimental condition was the one obtained with the desired frequency range and slope, varying with the combination of factors used in the experimental condition. In the case of experimental conditions involving two channels, the level was adjusted according to the following strategy: the difference in LDL between high and low channels, i.e. $L_2 - L_1$, was computed. The output of the channel having the lower LDL was attenuated by $|L_2 - L_1|$.

The summed output of the two channels was then fed to the ear, and the output was set at $L_b$ - 5 dB. This method ensured that the stimulus in both the low frequency channel and the high frequency channel were at 5 dB below LDL.

For all experimental conditions, a 400 Hz calibration tone was used. The tone corresponded to the average maximum RMS level of the word "mark" in the carrier phrase, and was adjusted so that the power of the speech peak was at the determined level. On the second track of each test tape, cafeteria noise was recorded for simultaneous presentation with the speech materials. When noise was also used, the 400 Hz calibration tone on the noise track was adjusted so that the peak power was at the determined level.

For peak clipped conditions, the input intensity was gradually increased until the first sign of peak clipping at the output could be observed on an oscilloscope. This intensity level was considered the threshold of peak clipping. The intensity was then adjusted to the threshold of peak clipping +10 dB.

During the entire experiment, attack time was held constant at a nominal setting of 2 msec., and release time at a nominal setting of 20 msec., as discussed in Chapter III.

The pilot study consisted of an 8 x 2 x 2, i.e. 32, factorial design. The variables considered were as follows: Amplitude Mode, which included non-compression amplification both with and without peak clipping, and single- and two-channel compression amplification; Frequency Response, which included a 0 dB/octave slope and a 6 dB/octave slope; and Listening Mode, which included both quiet and noise at a signal-to-noise (S/N) ratio of 10 dB. Table 9 summarizes the experimental conditions considered in the pilot study.

The experimental protocol was designed based on the following:

It was desired to compare non-compression amplification, both with and without peak clipping, to various modes of compression amplification, such as single- and two-band compression. Using both single- and two-band compression was aimed at seeing if varying the compression ratio in each band of the multiband system had an effect on intelligibility, as compared to a single-band system. In addition, it was aimed at seeing if there was any difference between a single wide band employing a specific compression ratio, and two bands, each employing the same compression ratio as the single-band condition. The difference between the latter two conditions is as follows: In a one-band compression system, any sound that drives the system into compression will compress all frequencies within its frequency range. In a two-band compression system, even if the compression ratio is the same for both bands, compression does not always occur in both bands at the same time. For example, the low frequency band of the two-band system may be driven into compression by a vowel. However, there may not be a

Table 9. The experimental variables considered in the 8 x 2 x 2, i.e. 32, factorial design of the pilot study.

| AMPLITUDE MODE | Compression ratio | Frequency range, Hz | Input level to subject |
|---|---|---|---|
| Non-compression, no peak clipping | Channel 1: 1:1 | Channel 1: 125-6000 | LDL - 5 dB |
| Non-compression, with peak clipping | Channel 1: 1:1 | Channel 1: 125-6000 | LDL - 5 dB |
| Compression, single channel | Channel 1: 1.5:1 | Channel 1: 125-6000 | LDL - 5 dB |
| Compression, single channel | Channel 1: 3:1 | Channel 1: 125-6000 | LDL - 5 dB |
| Compression, dual channel | Channel 1: 1.5:1<br>Channel 2: 1.5:1 | Channel 1: 125-1000<br>Channel 2: 1000-6000 | LDL - 5 dB |
| Compression, dual channel | Channel 1: 3:1<br>Channel 2: 3:1 | Channel 1: 125-1000<br>Channel 2: 1000-6000 | LDL - 5 dB |
| Compression, dual channel | Channel 1: 1.5:1<br>Channel 2: 3:1 | Channel 1: 125-1000<br>Channel 2: 1000-6000 | LDL - 5 dB |
| Compression, dual channel | Channel 1: 3:1<br>Channel 2: 1.5:1 | Channel 1: 125-1000<br>Channel 2: 1000-6000 | LDL - 5 dB |

FREQUENCY RESPONSE

0 dB/octave

6 dB/octave

LISTENING MODE

Quiet

S/N = 10 dB

strong frequency component to drive the high frequency band into compression, too. In the single-band compression hearing aid, however, this same vowel would be compressed over its entire frequency range.

The compression ratios chosen were based on average data derived from the literature. The data entailed those approximate ratios that appeared to have yielded maximum benefit regarding speech intelligibility.

Since a fixed frequency-gain characteristic was chosen, a comparison between flat frequency response and mild high frequency emphasis of 6 dB/octave was desired. Because the study was limited to only two frequency responses, the latter was chosen because it was often selected by subjects in the Wearable Master Hearing Aid (WMHA) study (Levitt and White, 1978), and not only because it was recommended by the Harvard study (Davis, et al., 1947).

Conditions that were believed would optimize differences among hearing aids were chosen. For example, experience has shown that noise, such as that encountered by hearing aid users, is helpful in differentiating among hearing aids. At a S/N ratio of 10 dB, intelligibility is generally reduced, but not to so great an extent that use of a hearing aid is no longer possible.

And, finally, the female version of the Nonsense Syllable Test (NST), discussed below, was used, because it was found to be more difficult than the male version in the WMHA study (Levitt and White, 1978). Consequently, it was believed it might be more effective in discriminating among the various hearing aid settings employed in the study.

In any study, only a finite number of conditions can be tested, because of practical constraints. The above variables were chosen since it was felt that they would be the most likely to highlight the effect, if any, of single- or multichannel compression amplification on intelligibility, as compared to non-compression amplification, the latter both with and without peak clipping.

## Test Equipment

All testing was conducted in a two-room double-walled IAC test booth, whose inner room measured 100" x 108" x 78". The apparatus used for testing speech intelligibility for the NST under the various experimental conditions were as follows: The speech and noise (the latter, when used) were passed from a tape recorder/reproducer to two attenuators that allowed intensity control for each track independently. The speech and noise were then mixed (i.e. if noise was being used for the specific experimental condition) and processed by the compression amplifier. The input to the compression amplifier was monitored by use of a voltmeter. The 400 Hz calibration tone on each track of each test tape was adjusted to the reference level of 100 mv. The signals passed from the compression amplifier to an attenuator, and then to a Grason-Stadler 1701 audiometer.

The input to the audiometer was visually monitored on an oscilloscope. The audiometer was used for intensity control of the input to the subject. The subject was seated in the inner room of the test booth, and wore TDH-39 earphones set in MX-41/AR cushions. Both speech and noise were presented from the same earphone, placed on the test ear. A block diagram of the apparatus used for testing is shown in Figure 17.

## Measurement of Speech Discrimination

The speech discrimination test used in the present study was the Nonsense Syllable Test (NST), developed for use in the Wearable Master Hearing Aid (WMHA) study (Levitt and White, 1978). It met certain basic requirements, such as low test-retest variability and minimal learning effects.

The test was of the closed-response-set type and used nonsense syllables as its test material. It consisted of seven subtests, each subtest embodying a set of seven to nine nonsense syllables of the consonant-vowel (CV) or vowel-consonant (VC) type. The choice of subtests was weighted to have a greater proportion of those sounds with which hearing aid users tend to have difficulty. Sounds such as voiceless consonants in the final position are relatively effective in discriminating among various hearing aid characteristics, since these consonants are difficult, but not impossible, to understand (Levitt and White, 1978).

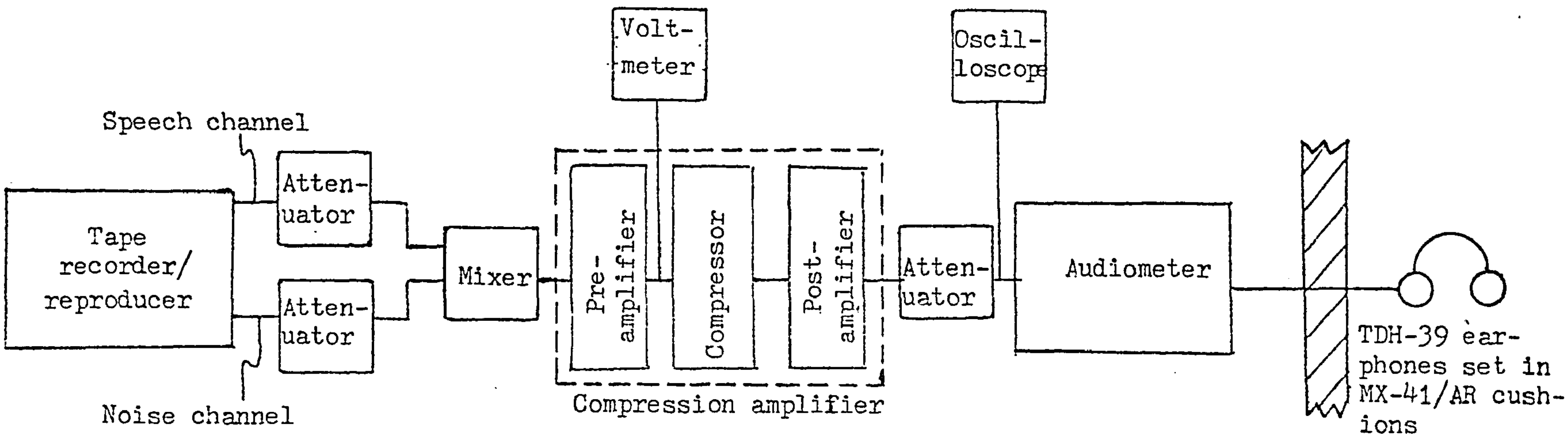



Fig. 17. Block diagram of apparatus used for speech reproduction through the multi-channel hearing aid in the pilot study.

The test items that made up the NST are shown in Table 10. Each column composes one subtest. Subtests differed in three characteristics: the class of consonants represented, the position of the consonants, and the vowel context. The entire test consisted of a total of 62 items, with eight randomized versions of the test.

The test was recorded by both a male and female speaker "without marked regional characteristics" (Levitt and White, 1978). Since the female version was found to be the more difficult, both among normal hearing listeners and hearing-impaired listeners, only the female recording of the test was used in the present study. This was in an attempt to maximize the possible advantages of various settings of the experimental hearing aid.

Each syllable was embedded within the carrier phrase "You will mark _______, please." The level of the word "mark" in the carrier phrase was equated for each test item by selectively introducing attenuation in the overall recording. The level during the most intense vowel of any test item was measured to be about 25 dB greater than the level during the weakest consonant of any test item.

Cafeteria noise was recorded on the second track of each test tape. This allowed for a competing signal to be presented simultaneously with the speech, a communication condition that hearing aid wearers often find themselves in.

The working test tapes were copied directly from the

Table 10. Test items that made up the seven subtests of the Nonsense Syllable Test (NST), as utilized in the pilot study.

| Subtest 1 | 2 | 3 | 4 | 5 | 6 | 7 |
|---|---|---|---|---|---|---|
| at | ut | it | ad | ta | da | da |
| ak | uk | ik | ag | ka | ga | ga |
| ap | up | ip | ab | pa | ba | ba |
| af | uf | if | av | fa | la | va |
| as | us | is | az | sa | ra | za |
| aθ | uθ | iθ | að | θa | ja | ða |
| aʃ | uʃ | iʃ | an | ʃa | wa | na |
| | | | am | tʃa | dʒa | ma |
| | | | aŋ | ha | | |

eight respective randomizations of the Sub-Master tapes, which were copies of the Master tapes made for the WMHA study (Levitt and White, 1978). Each test tape was carefully checked by listening and by graphic level recordings to check for any possible distortions in the test tapes.

The subject's response to a syllable within a given subtest was limited to syllables within the same subtest. All the syllables within a particular subtest served as the response foils. Thus, within any subtest, voicing errors were not possible, although errors with respect to place and/or manner of articulation were possible. Voicing errors were not considered among the response foils, since the feature of voicing tends to be well-perceived even under extremely adverse conditions, such as filtering out of the high frequencies (Miller and Nicely, 1955). The response form used by the subjects is shown in Appendix F (note that responses were written by the subjects).

## Results

An analysis of variance was performed on the data obtained in the pilot study. The main factors in this analysis were Amplitude Mode, Frequency Response, Listening Mode, Subjects, and Subtests (see Table 10 for listing of the experimental conditions). Table 11 shows the results of this analysis.

Statistically significant effects were found at the .001 level for Listening Mode, Subjects, and Subtests, and at the .05 level for Frequency Response. There were also

Table 11. Analysis of variance of data obtained in the pilot study.

| Source of variation | Sums of squares | Degrees of freedom | Mean squares | F-ratio |
|---|---|---|---|---|
| Amplitude Mode (A) | 0.29 | 7 | 0.041 | 1.70 |
| Frequency Response (B) | 0.10 | 1 | 0.109 | 4.47 * |
| Listening Mode (C) | 14.25 | 1 | 14.254 | 582.28 ** |
| AB interaction | 0.10 | 7 | 0.014 | 0.60 |
| AC interaction | 0.39 | 7 | 0.057 | 2.33 * |
| BC interaction | 0.30 | 1 | 0.307 | 12.56 ** |
| ABC interaction | 0.29 | 7 | 0.042 | 1.72 |
| Subjects (S) | 7.86 | 4 | 1.967 | 80.36 ** |
| AS interaction | 0.91 | 28 | 0.032 | 1.32 |
| BS interaction | 0.87 | 4 | 0.219 | 8.96 ** |
| CS interaction | 1.09 | 4 | 0.272 | 11.14 ** |
| ABS interaction | 0.71 | 28 | 0.025 | 1.04 |
| ACS interaction | 1.56 | 28 | 0.055 | 2.28 ** |
| BCS interaction | 0.24 | 4 | 0.061 | 2.50 * |
| ABCS interaction | 1.22 | 28 | 0.043 | 1.78 * |
| Modules or subtests (M) | 2.17 | 6 | 0.362 | 14.82 ** |
| AM interaction | 1.01 | 42 | 0.024 | 0.98 |
| BM interaction | 0.23 | 6 | 0.039 | 1.59 |
| CM interaction | 0.67 | 6 | 0.112 | 4.61 ** |
| ABM interaction | 0.67 | 42 | 0.016 | 0.66 |
| ACM interaction | 0.92 | 42 | 0.021 | 0.89 |
| BCM interaction | 0.16 | 6 | 0.027 | 1.11 |
| ABCM interaction | 0.99 | 42 | 0.023 | 0.96 |
| SM interaction | 4.13 | 24 | 0.172 | 7.03 ** |
| ASM interaction | 4.15 | 168 | 0.024 | 1.00 |
| BSM interaction | 1.04 | 24 | 0.043 | 1.77 * |
| CSM interaction | 0.88 | 24 | 0.036 | 1.50 |
| ABSM interaction | 2.92 | 168 | 0.017 | 0.71 |
| ACSM interaction | 3.63 | 168 | 0.021 | 0.88 |
| BCSM interaction | 0.80 | 24 | 0.033 | 1.37 |
| ABCSM | 4.11 | 168 | 0.024 | |
| Total | 58.80 | 1119 | | |

*significant at 0.05 level
**significant at 0.001 level

significant interactions at the .001 level between Subjects and the following factors: Subtests, Listening Mode and Frequency Response. At the .05 level a significant interaction was seen between Amplitude Mode and Listening Mode.

Three-way interactions were seen among Amplitude Mode, Listening Mode, and Subject (at the .001 level); Subtest, Frequency Response, and Subject (at the .05 level); and Listening Mode, Frequency Response, and Subject (also at the .05 level).

Table 12 shows means for the main effects: Amplitude Mode, Frequency Response, and Listening Mode.

The data showed that Frequency Response had a significant effect. On the average, the 0 dB/octave slope elicited a slightly better score than the 6 dB/octave slope (60.8% for flat frequency response, 58.8% for the 6 dB/octave slope). However, although the average differences across subjects was small, there were large individual differences among subjects. Three out of five subjects scored better with a flat frequency response. For example, subject LS, who had a mildly downward sloping hearing loss, showed a gain in score of 11 percentage points for flat frequency response over 6 dB/octave slope. For Listening Mode, all subjects scored consistently better in quiet than in noise (71.1% in quiet, averaged over all conditions, and 48.5% in noise, averaged over all conditions).

Although there was no significant difference in the

Table 12. Percent correct scores for subjects as a function of Amplitude Mode, Frequency Response, and Listening Mode. Mean scores are also shown. The scores for each Main Effect have been averaged over the remaining factors.

| AMPLITUDE MODE | SUBJECT KH | LS | RM | RS | RG | Mean | Means Across Main Effects |
|---|---|---|---|---|---|---|---|
| Non-compression, no peak clipping | 69.2 | 65.8 | 59.4 | 51.5 | 46.5 | 58.9 | 59.4 (non-compression) |
| Non-compression, with peak clipping | 69.9 | 61.2 | 61.0 | 61.4 | 45.3 | 59.8 | |
| Compression, 1.5:1 | 74.0 | 70.5 | 61.0 | 57.2 | 48.9 | 62.3 | 60.0 (single-channel compression) |
| Compression, 3:1 | 68.7 | 66.1 | 50.9 | 60.7 | 41.4 | 57.6 | |
| Compression, 1.5:1, 1.5:1 | 71.4 | 69.8 | 60.2 | 54.6 | 48.5 | 61.0 | 60.1 (two-channel compression) |
| Compression, 3:1, 3:1 | 68.0 | 66.0 | 60.3 | 55.3 | 44.6 | 58.8 | |
| Compression, 1.5:1, 3:1 | 72.6 | 70.4 | 63.0 | 54.0 | 49.4 | 61.9 | |
| Compression, 3:1, 1.5:1 | 73.4 | 60.5 | 51.6 | 60.0 | 47.8 | 58.7 | |
| **FREQUENCY RESPONSE** | | | | | | | |
| 0 dB/octave | 72.0 | 71.8 | 56.2 | 59.1 | 44.9 | 60.8 | |
| 6 dB/octave | 69.8 | 60.8 | 60.6 | 54.6 | 48.2 | 58.8 | |
| **LISTENING MODE** | | | | | | | |
| Quiet | 85.4 | 78.8 | 66.2 | 64.2 | 60.8 | 71.1 | |
| S/N = 10 dB | 56.4 | 53.8 | 50.7 | 49.5 | 32.3 | 48.5 | |

effect of Amplitude Mode averaged over subjects, there was a three-way interaction among Amplitude Mode, Listening Mode, and Subject. These data are given in Table 13, showing significant individual differences with respect to the joint effects of Amplitude Mode and Listening Mode. For example, for subject RM, who had a sharply downward sloping hearing loss, his best score in quiet was achieved with non-compression amplification, no peak clipping, but his best score in noise was achieved with multichannel compression. The reverse held true for subject RS, who had a flat hearing loss, whose best score in quiet was achieved with multiband compression amplification, and in noise with non-compression amplification, no peak clipping. Regarding differences among two-channel compression conditions, in quiet, all subjects achieved their best score when at least one channel had a compression ratio of 3:1. For four of the five subjects, this channel was the high frequency one. In noise, only three of the five subjects scored best when at least one channel had a compression ratio of 3:1.

Table 14 shows scores, by subject, for the seven subtests of the NST. Subjects scored better, on the average, for voiced consonants than for voiceless (61.3% for voiced subtests, 58.6% for voiceless). The best score, across subjects, was obtained for Subtest 6, voiced consonant preceding /a/. This subtest included the plosives and semi-vowels among its consonants. The poorest score, averaged over subjects, was obtained for Subtest 5, voiceless consonant pre-

Table 13. Percent correct scores for subjects as a function of Amplitude Mode and Listening Mode. Mean scores are also shown. The scores for each main effect have been averaged over the remaining factors.

| AMPLITUDE MODE | LISTENING MODE | SUBJECT KH | LS | RM | RS | RG | Mean |
|---|---|---|---|---|---|---|---|
| Non-compression, no peak clipping | Quiet | 82.8 | 82.9 | 72.9 | 60.5 | 60.6 | 71.9 |
| | S/N = 10 dB | 55.6 | 48.7 | 45.9 | 42.6 | 32.4 | 45.0 |
| Non-compression, with peak clipping | Quiet | 80.2 | 64.6 | 64.7 | 66.4 | 57.4 | 66.7 |
| | S/N = 10 dB | 59.5 | 57.8 | 57.4 | 56.4 | 33.2 | 52.9 |
| Compression, 1.5:1 | Quiet | 86.3 | 78.5 | 65.9 | 67.2 | 72.9 | 74.2 |
| | S/N = 10 dB | 61.5 | 62.6 | 56.2 | 47.3 | 24.9 | 50.5 |
| Compression, 3:1 | Quiet | 82.8 | 73.1 | 59.6 | 66.0 | 60.6 | 68.4 |
| | S/N = 10 dB | 54.7 | 59.2 | 42.3 | 55.4 | 22.3 | 46.8 |
| Compression, 1.5:1, 1.5:1 | Quiet | 86.8 | 80.6 | 68.0 | 58.0 | 63.1 | 71.3 |
| | S/N = 10 dB | 56.1 | 59.0 | 52.4 | 51.1 | 34.0 | 50.5 |
| Compression, 3:1, 3:1 | Quiet | 88.9 | 85.5 | 69.2 | 60.9 | 53.1 | 71.5 |
| | S/N = 10 dB | 47.2 | 46.3 | 51.5 | 50.0 | 36.0 | 46.2 |
| Compression, 1.5:1, 3:1 | Quiet | 88.7 | 85.9 | 65.1 | 64.7 | 63.7 | 73.6 |
| | S/N = 10 dB | 56.5 | 54.9 | 61.0 | 43.1 | 35.1 | 50.1 |
| Compression, 3:1, 1.5:1 | Quiet | 86.6 | 79.4 | 64.1 | 70.0 | 55.2 | 71.1 |
| | S/N = 10 dB | 60.2 | 41.5 | 39.0 | 50.0 | 40.3 | 46.2 |

Table 14. Percent correct scores for each subtest of the Nonsense Syllable Test (NST). Data are shown for each subject, as well as the mean scores across subjects. All scores are averaged over the remaining factors.

| SUBTEST | | | SUBJECT | | | | | |
|---|---|---|---|---|---|---|---|---|
| Number | | Description | KH | LS | RM | RS | RG | Mean |
| 1 | /av̄/ | voiceless consonant following /a/ | 71.6 | 65.4 | 55.2 | 52.1 | 46.9 | 58.2 |
| 2 | /uv/ | voiceless consonant following /u/ | 66.7 | 64.8 | 62.5 | 50.8 | 62.2 | 61.4 |
| 3 | /iv̄/ | voiceless consonant following /i/ | 71.1 | 61.7 | 62.2 | 47.7 | 49.5 | 58.4 |
| 4 | /av/ | voiced consonant following /a/ | 66.6 | 64.6 | 64.8 | 55.9 | 33.8 | 57.1 |
| 5 | /v̄a/ | voiceless consonant preceding /a/ | 74.6 | 65.4 | 44.9 | 52.1 | 45.1 | 56.4 |
| 6 | /va/ | voiced consonant preceding /a/* | 83.9 | 69.0 | 64.3 | 80.0 | 52.7 | 70.0 |
| 7 | /va/ | voiced consonant preceding /a/** | 61.8 | 73.0 | 55.1 | 59.3 | 35.7 | 57.0 |

*included semi-vowels, plosives, and an affricate
**included nasals, plosives, and fricatives

ceding /a/, which included the plosives, a glottal, and several fricatives among its consonants.

## Discussion

In analyzing the data obtained in the pilot study, it was necessary to differentiate among several factors:

1. Those affecting the subjects' performance with the compression amplifier, i.e. Amplitude Mode, Listening Mode and Frequency Response.

2. Those factors affecting individual subjects, indicating whether or not there were individual differences, and, if there were, the nature of these differences.

3. The test itself, which offered some insight into how hearing-impaired people perceive sounds.

4. And, lastly, the electroacoustic characteristics of the hearing aid.

The data obtained indicated which of the experimental variables significantly affected hearing aid performance, and the relative magnitude of these effects. In order of their relative magnitudes, the major effects were: Listening Mode (i.e. quiet vs. noise), differences among subjects, subtests of the NST, and Frequency Response (i.e. 0 dB/octave vs. 6 dB/octave slope). Amplitude Mode was not found to be a significant variable. Significant interactions existed between Subjects and Frequency Response, indicating large individual differences as to which conditions provided the best performances. All this information was used in considering which variables to eliminate, and which to retain or modify, in order to extract the most information in the main experiment. Each of these factors mentioned above, and its influence on the design of the main experiment, is discussed below.

Factors Affecting Performance with the Hearing Aid

Although it was a primary variable of interest, Amplitude Mode was not statistically significant. However, it did interact significantly with Listening Mode, and, in a three-way interaction, with Listening Mode and Subject. The interaction with Subject was of importance for this study, since it showed that different subjects obtained their highest scores under different sets of conditions. This result emphasized the importance of individualized adjustment for hearing aid fitting.

Frequency Response was found to be a significant variable. In addition, it interacted significantly with Subject, and with Listening Mode, indicating its selective effect. These findings also supported the notion that individualized fitting should be used in the main experiment, in order to assess the relative merits of the various hearing aid settings under optimum conditions. The frequency-gain characteristics that had been used in the pilot study were 0 dB/octave slope and 6 dB/octave slope. However, as mentioned, because of the Subject by Frequency interaction obtained for the different Listening Modes, it was felt that individual frequency-gain shaping should be included in the main experiment.

Accordingly, a method of frequency shaping was developed (Levitt, 1977) which would place as much of the speech spectrum into the listener's residual hearing area as possible, with an adjustment to reduce upward spread of masking. Details of the procedure are described in Chapter V.

Listening Mode showed the largest effect of the experimental variables, and interacted significantly with other factors, i.e. Frequency Response, Subject, and Amplitude Mode. Because of these interactions, it was felt that Listening Mode should continue to be used for each combination of experimental variables in the main experiment.

Factors Affecting Individual Subjects

Subjects were found to differ significantly from one another. In addition, there were large two-way interactions between Subjects and the following factors: Subtests, Listening Mode, and Frequency Response. Several three-way interactions were also significant. These were Subjects and: Listening Mode and Amplitude Mode; Frequency Response and Subtest; and Frequency Response and Listening Mode. From these many complex interactions, the best scores for each subject were obtained on different combinations of conditions, again stressing the importance of individualized hearing aid fittings. In addition, it was also important that each subject be tested on all permutations of the experimental variables in the main experiment (i.e. a factorial design should be used).

Use of the Nonsense Syllable Test (NST) as the Test Instrument

Although the test instrument used, the NST, had low test-retest variability, it was not entirely free of learning effects. On several occasions where a specific test condition had been repeated, a score that was lower by an average of 6 percentage points was consistently found on the first administration of the test, leading to a concern about learning effects. It was thus necessary to introduce a check for learning. The main study was therefore extended to include four measurements for each experimental condition, instead of only one. In addition, with four measurements, it was possible to analyze consonant confusions with greater reliability than with only one measurement.

Because of practical limitations of the subjects' time, the use of repeated measurements called for a reduction in the size of the overall protocol, resulting in more observations per subject, but with fewer test conditions. Some variables were consequently reduced in size, such as the reduction of four conditions of two-band compression to two conditions. In addition, the size of the NST was reduced from seven subtests to four subtests.

Electroacoustic Characteristics of the Hearing Aid

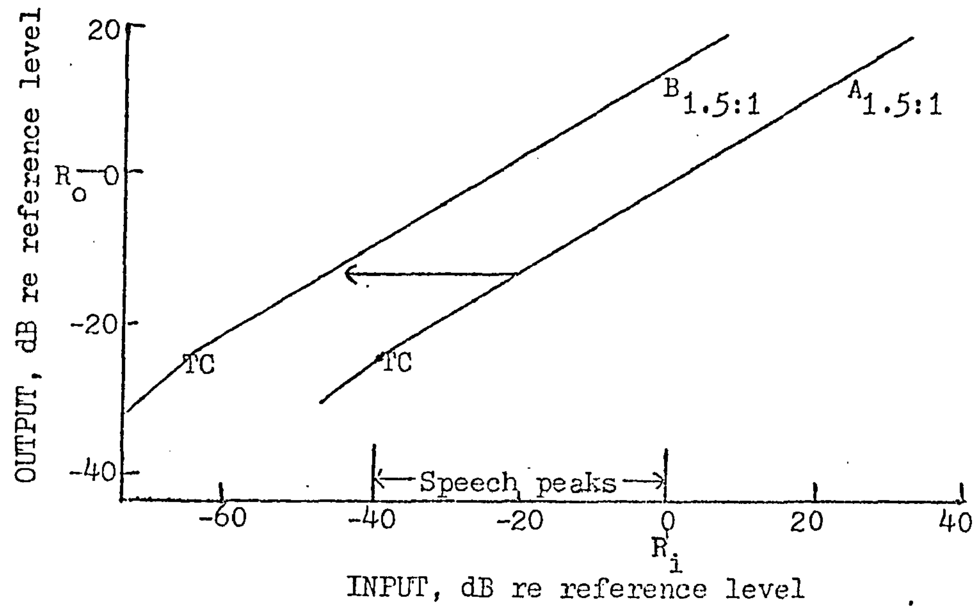
After completion of the pilot study and analysis of the results, the instrument was recalibrated and checked again completely, as described in Chapter III. All calibrations confirmed proper functioning of the hearing aid.

However, since the variable Amplitude Mode (i.e. conditions of non-compression and compression) had not been found to be statistically significant, it was considered that the operating conditions of the compression amplifier may have interfered with this factor. That is, there was concern that the internal noise of the hearing aid may have interfered with low level signals. If the speech peaks span a range of roughly 30 to 40 dB, then, with the higher speech peaks set at the reference level ($R_1$ in Fig. 7), the lower speech levels may be close to, or even below, the level of the internal noise. As can be seen in Fig. 7, as the compression ratio increases, the noise level at the output increases. Therefore, for the main experiment, the operating level of the peaks of speech was shifted upwards by 25 dB. That is, 25 dB of pre-compression gain was introduced, in order to keep the lower speech levels well above the internal noise.

The effect of 25 dB of pre-compression gain is to shift the input-output curve 25 dB to the left (this is illustrated in Fig. 5). But if, in combination with a change in pre-compression gain, there is also a difference in the compression ratio of the two channels, the higher speech peaks of the two channels will no longer be at a common output level, as they originally were (see Fig. 7).

This condition existed in the main experiment, under the condition of two-channel compression, with compression ratio of 1.5:1 in channel 1, and 3:1 in channel 2. Thus, to ensure that the higher speech peaks were always at the same level in both channels, despite differences in compression ratios, post-compression attenuation of 8.4 dB was imposed on the channel utilizing a compression ratio of 1.5:1. This value was derived as follows: at a compression ratio of 3:1, with pre-compression gain of 25 dB re the reference level, the output was 25 dB/3, or 8.3 dB. At a compression ratio of 1.5:1, with the same amount of pre-compression gain, the output was 25 dB/1.5 or 16.7 dB. The difference between the two levels was 8.4 dB. In order to produce a common reference level, so that the higher speech peaks fell at both the same input and output ($R_i = R_o$), post-compression attenuation of 8.4 dB was imposed on the channel utilizing a compression ratio of 1.5:1. As is conceptually illustrated in Fig. 5, post-compression attenuation shifted the input-output curve downwards. The combined effect of these two steps is illustrated in Figures 18 through 20, where the input-output functions are computed for several experimental conditions.

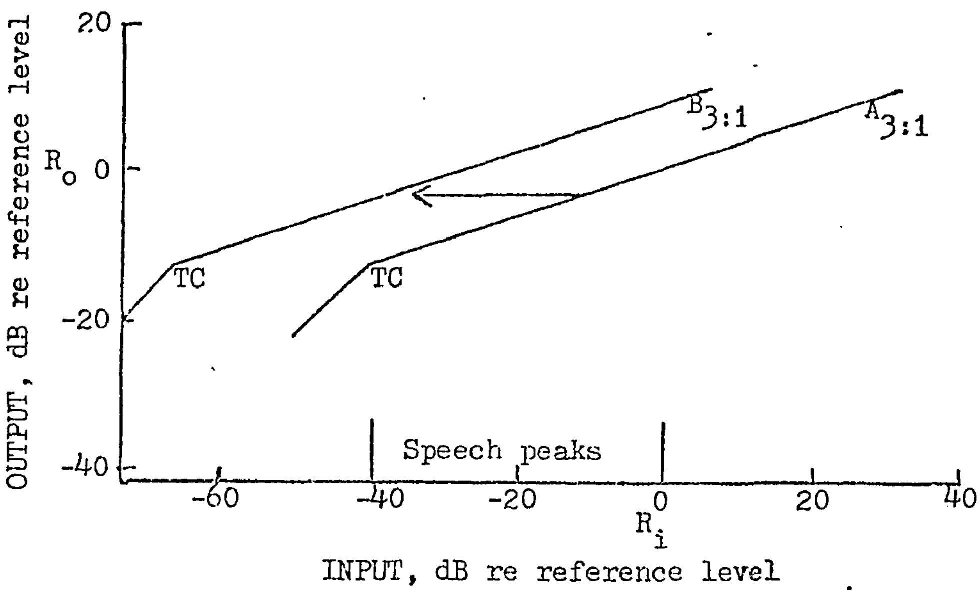
Another change imposed on the electroacoustic characteristics of the hearing aid was in the frequency range utilized. In the pilot study, when one channel was used, the frequency range was 125 - 6000 Hz. With the use of two channels, the frequency range was the same, with a




KEY

| | |
|---|---|
| $R_i$ | Reference level at input |
| $R_o$ | Reference level at output |
| TC | Threshold of compression |
| $A_{1.5:1}$ | Input-output function before 25 dB of pre-compression gain, at compression ratio of 1.5:1 |
| $B_{1.5:1}$ | Input-output function after 25 dB of pre-compression gain, at compression ratio of 1.5:1 |

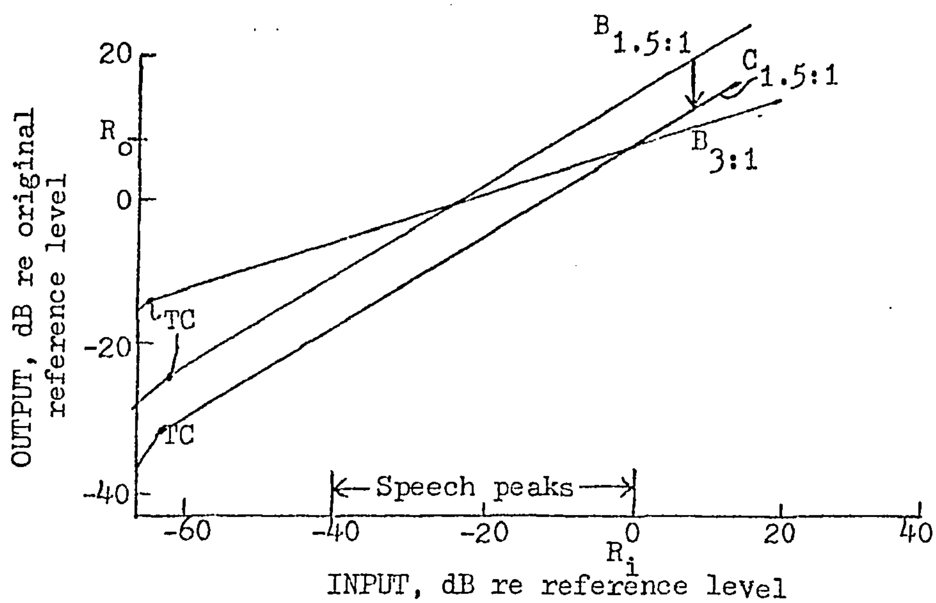
Fig. 18. Idealized input-output functions at compression ratio of 1.5:1, before and after 25 dB of pre-compression gain. Arrow indicates the direction of shift with the change in gain.




KEY

| | |
|---|---|
| $R_i$ | Reference level at input |
| $R_o$ | Reference level at output |
| TC | Threshold of compression |
| $A_{3:1}$ | Input-output function before 25 dB of pre-compression gain, at compression ratio of 3:1 |
| $B_{3:1}$ | Input-output function after 25 dB of pre-compression gain, at compression ratio of 3:1 |

Fig. 19. Idealized input-output functions at compression ratio of 3:1, before and after 25 dB of pre-compression gain. Arrow indicates the direction of shift with the change in gain.




KEY

| | |
|---|---|
| $R_i$ | Reference level at input |
| $R_o$ | Reference level at output |
| TC | Threshold of compression |
| $B_{1.5:1}$ | Input-output function after 25 dB of pre-compression gain, at compression ratio of 1.5:1 (refer to Fig. 18) |
| $B_{3:1}$ | Input-output function after 25 dB of pre-compression gain, at compression ratio of 3:1 (refer to Fig. 19) |
| $C_{1.5:1}$ | Input-output function after additional 8.4 dB of post-compression attenuation to function $B_{1.5:1}$ |

Fig. 20. Idealized input-output functions at compression ratios of 1.5:1 ($B_{1.5:1}$) and 3:1 ($B_{3:1}$), with 25 dB of pre-compression gain for each compression ratio (refer to Figs. 18 and 19, respectively). Also shown is the idealized input-output function ($C_{1.5:1}$) at compression ratio of 1.5:1, with an additional post-compression attenuation of 8.4 dB to function $B_{1.5:1}$. Arrow indicates the direction of shift with the attenuation.

crossover frequency of 1000 Hz. Davis, et al. (1947) had found that discrimination was best with a low frequency cutoff that was no lower than 200 Hz, and that there was no significant change in performance if the low frequency cutoff was raised to 400 Hz. Empirical evidence thus favors a lower frequency limit of amplification, perhaps because of the possibility of upward spread of masking. Further, the method of individualized frequency shaping, as shall be described, would typically raise the gain in the low frequencies. This would raise the likelihood of upward spread of masking. To counteract the possibility, it was decided to raise the lower cutoff frequency to 250 Hz in the main experiment. It was also felt that in the two-band conditions, the low frequency band might then be too narrow (i.e. 250 Hz to 1000 Hz), and the crossover frequency was therefore raised to 1500 Hz. The frequency range was now 250 Hz to 6000 Hz, with a crossover frequency of 1500 Hz in the two-channel condition.

## Summary

In summary, the pilot study suggested the following: First, it was desirable to make both electroacoustic changes and changes in the speech discrimination test for the main study. Second, it was suggested that one of the most important conditions necessary for optimizing the various hearing aid settings might be individualization of the frequency-gain shaping. Details of the experimental procedure utilized in the main experiment are described in the following chapter.

# CHAPTER V

## METHODOLOGY OF THE MAIN EXPERIMENT

### Description of the Subjects

A group of four subjects, two females and two males, aged 45 to 69 years (mean age 58.5 years), was chosen for the main experiment. The criteria used for both subject and ear selection were the same as those used in the pilot study, and are described in Chapter IV, pages 117-121. Table 15 summarizes the air-conduction thresholds of the subjects for the test ear (that is, the experimental ear), and Table 16 summarizes this information for the non-test ear. The standard deviations and the ranges of lowest to highest threshold values are also shown. Figure 21 plots, in audiogram form, the values given in Table 15. The average audiogram for the test ear was mildly downward sloping. Figure 22 plots, in audiogram form, the values given in Table 16. The average audiogram for the non-test ear approximated that of the test ear. Figures 43 through 47 of Appendix G show the audiograms obtained for each of the four subjects.

Tables 17 and 18 summarize the results of audiometric speech tests performed on the four subjects, for the test ear and non-test ear, respectively. Mean Pure Tone Averages, Speech Reception Thresholds, and Word Discrimination scores

Table 15. Mean pure tone thresholds for the test ear, for four subjects. Standard deviations and lowest to highest thresholds are also included.

| | Frequency, Hz | | | | | | |
|---|---|---|---|---|---|---|---|
| | 125 | 250 | 500 | 1000 | 2000 | 4000 | 8000 |
| Mean threshold, dB HTL | 27.75 | 37.5 | 53.75 | 61.25 | 63.75 | 65.0 | 76.25 |
| Standard deviation, dB | 16.5 | 16.8 | 14.3 | 14.3 | 8.0 | 14.4 | 8.5 |
| Lowest - highest thresholds, dB HTL | 10-50 | 20-60 | 45-75 | 45-80 | 55-70 | 50-85 | 65-85 |

Table 16. Mean pure tone thresholds for the non-test ear, for four subjects. Standard deviations and lowest to highest thresholds are also included.

| | Frequency, Hz | | | | | | |
|---|---|---|---|---|---|---|---|
| | 125 | 250 | 500 | 1000 | 2000 | 4000 | 8000 |
| Mean threshold, dB HTL | 35.0 | 36.25 | 50.0 | 62.5 | 62.5 | 72.5 | 67.5 |
| Standard deviation, dB | 20.4 | 16.5 | 20.4 | 25.3 | 20.2 | 17.0 | 15.5 |
| Lowest - highest thresholds, dB HTL | 20-65 | 25-60 | 35-80 | 45-100 | 45-90 | 65-95 | 50-85 |

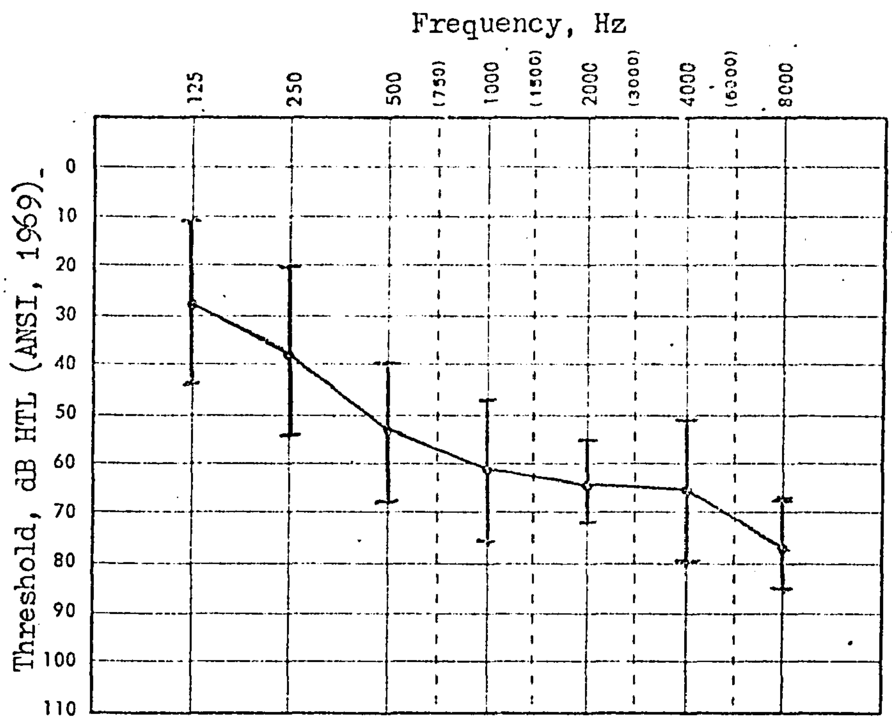



Fig. 21. Mean pure tone thresholds of the test ear, for four subjects. Bars indicate standard deviations.

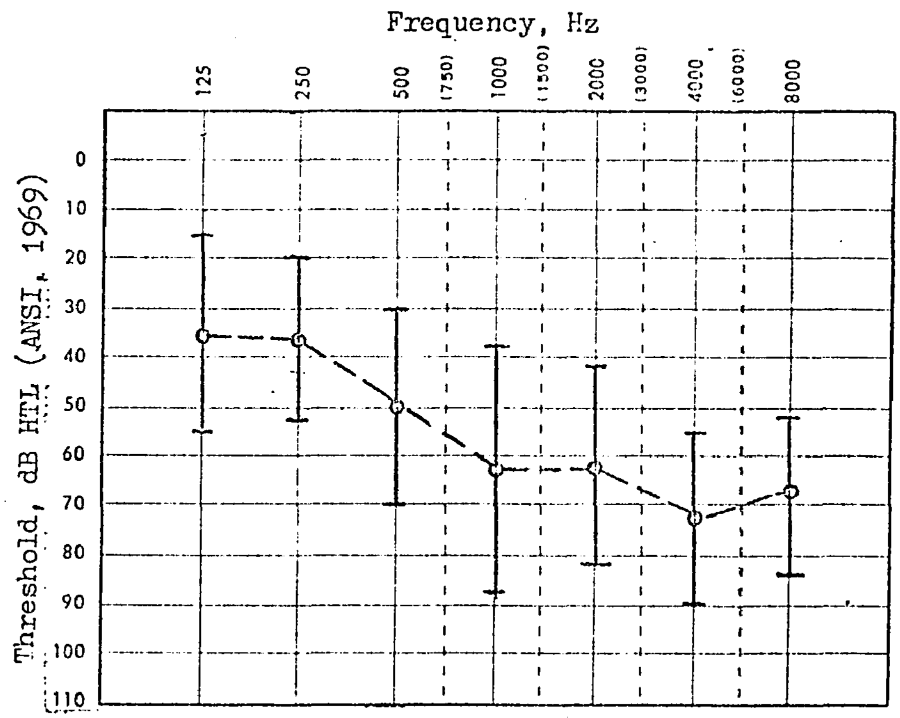



Fig. 22. Mean pure tone thresholds of the non-test ear, for four subjects. Bars indicate standard deviations.

Table 17. Results of speech audiometric tests of the test ear, for four subjects. Pure Tone Average, means, standard deviations, and lowest to highest values are also given.

| Audiological measures | Mean | Standard deviation | Lowest-highest values |
|---|---|---|---|
| Pure Tone Average (PTA), dB HTL | 59.5 | 11.6 | 53.3-75 |
| Speech Reception Threshold (SRT), dB HTL | 60.0 | 14.7 | 45-80 |
| Word Discrimination Score (WDS), percent | 65.5 | 6.7 | 56-70 |
| Most Comfortable Loudness (MCL), dB HTL | 80.0 | 7.1 | 75-90 |
| Loudness Discomfort Level (LDL), dB HTL | 97.5 | 2.8 | 95-100 |
| Dynamic range (SRT - LDL), dB | 37.5 | 12.8 | 20-50 |
| Range of MCL to LDL, dB | 17.5 | 5.0 | 10-20 |

Table 18. Results of speech audiometric tests of the non-test ear, for four subjects. Pure Tone Average, means, standard deviations, and lowest to highest values are also given.

| Audiological measures | Mean | Standard deviation | Lowest-highest values |
|---|---|---|---|
| Pure Tone Average (PTA), dB HTL | 58.4 | 21.2 | 45-90 |
| Speech Reception Threshold (SRT), dB HTL | 57.5 | 18.2 | 45-85 |
| Word Discrimination Score (WDS), percent | 62.0 | 17.1 | 36-76 |
| Most Comfortable Loudness (MCL), dB HTL | 82.5 | 9.6 | 75-95 |
| Loudness Discomfort Level (LDL), dB HTL | 97.5 | 8.6 | 90-105 |
| Dynamic range (SRT - LDL), dB | 40.0 | 13.5 | 20-50 |
| Range of MCL to LDL, dB | 15.0 | 8.6 | 10-20 |

are given for each ear, in addition to Most Comfortable Loudness levels and Loudness Discomfort Levels for speech stimuli. The standard deviations and the ranges of lowest to highest values are also shown.

Measurement of Speech Discrimination

The speech discrimination test utilized throughout the study was the Nonsense Syllable Test (NST). The format of the test and preparation of the test tapes are described in Chapter IV. As discussed, there were changes made in the number of subtests utilized for the main experiment. Only four subtests were used, instead of the original seven, of which two were taken from the original NST. These two were subtests number 2 and 3, consisting of voiceless consonant following /u/ and voiceless consonant following /i/. The remaining two subtests were taken from an Optional version of the NST, which consisted of five subtests. The two used in the main experiment were voiced consonant following /u/ and voiced consonant following /i/. These vowels were chosen because they represent two extremes among the vowels regarding their overall spectra: /u/ has most of its energy concentrated in the low frequencies, whereas /i/ has its energy spread out both in the low frequencies and the high frequencies. The test items that made up the new version of the NST are shown in Table 19.

Subtests differed in two characteristics: the class of consonants represented, and the vowel context. The

Table 19. Test items that made up the four subtests of the Nonsense Syllable Test (NST), as utilized in the main experiment.

| Subtest | | | |
|---|---|---|---|
| 1 | 2 | 3 | 4 |
| ud | ut | it | id |
| ug | uk | ik | ig |
| ub | up | ip | ib |
| uv | uf | if | iv |
| uz | us | is | iz |
| uð | uθ | iθ | ið |
| un | uʃ | iʃ | in |
| um | | | im |
| uŋ | | | iŋ |

entire test consisted of a total of 32 items, with eight randomized versions of the test. The female version of the test was used, as discussed in Chapter IV.

Measurement of LDLs Using One-Third Octave Bands of Noise

Loudness Discomfort Levels (LDLs) for one-third octave bands of noise were measured using a specially prepared tape. The purpose of this tape was to provide a means for calibrating the frequency-gain characteristic that would generate LDLs at the ear in a way that could be related to standard 6cc coupler measurements (a 6cc coupler was chosen in order to be able to relate the data obtained in this study to other published data, such as the Harvard study (Davis, et al., 1947)). The tape was prepared in a way that when it was played through the nominally flat response of the system it generated third octave bands of noise of equal sound pressure level, as measured in a 6cc coupler. Using this tape, the LDLs of third octave bands of noise were found, i.e. the frequency-gain characteristic needed to generate LDL at the ear was specified with respect to a stimulus that had a flat one-third octave band spectrum in the coupler.

It should be noted that the noise on the tape(which was of one-half hour duration), was wide-band, i.e. all of the bands were recorded simultaneously, and subsequent measurements were made through an external multifilter, in order to isolate individual one-third octave bands.

The details of how the tape was prepared are described below.

The first step in making a tape whose output was equal at each third octave band was to correct for the bandwidth effect of white noise, which was used in preparing the tape. Since white noise has a constant power density (i.e. equal power/Hz), and the bandwidth of the standard one-third octave bands increases proportionally with the center frequency, the total power within a band increases at the rate of 3 dB/octave, with increasing center frequency. However, the noise used may not have been true white noise. Therefore, as a check, it was passed through a multifilter set at 0 dB attenuation for all bands, and the intensity for each one-third octave band was measured on a voltmeter. A slight rolloff above 2500 Hz was found. Thus, a correction for this spectral variation was required, in addition to the correction for the bandwidth effect. The correction factors used may be found in Table 20, column (a).

The second step in preparing the tape concerned applying correction factors for two instruments in the experimental set-up that imposed frequency changes on stimuli passing through them. They were the compression amplifier and the earphone. Corrections for these two instruments were made as follows:

Correction for Compression Amplifier

With the frequency range of the compression amplifier set at 250 Hz to 6000 Hz, at a compression ratio of 1:1, the

Table 20. Separate effects of (a)bandwidth and spectral variation of white noise, (b)compression system, and (c)earphone, and (d)total combined effect. These were the effects that had to be corrected in order to make the noise signal flat, i.e. equal in level, by one-third octave bands, in a 6cc coupler. Values given are the corrections, in dB.

| MULTIFILTER CHANNEL | | (a) | (b) | (c) | (d) |
|---|---|---|---|---|---|
| Number | Center frequency, Hz | Correction for bandwidth effect and spectral variation, dB | Correction for compression amplifier response, dB | Correction for earphone response, dB | Overall correction (a+b+c), dB |
| 23 | 200 | 3 | 9 | 0 | 12 |
| 24 | 250 | 2 | 7 | 0 | 9 |
| 25 | 315 | 1 | 4 | 0 | 5 |
| 26 | 400 | 0 | 2 | 0 | 2 |
| 27 | 500 | -1 | 2 | 0 | 1 |
| 28 | 630 | -2 | 1.5 | 0 | -0.5 |
| 29 | 800 | -3 | 0.5 | -1 | -3.5 |
| 30 | 1000 | -4 | 1 | -2 | -5 |
| 31 | 1250 | -5 | 0 | -2 | -7 |
| 32 | 1600 | -6 | 0 | 0 | -6 |
| 33 | 2000 | -7 | 0 | 0 | -7 |
| 34 | 2500 | -8 | 0.5 | -2 | -9.5 |
| 35 | 3150 | -8 | 0.5 | -6 | -13.5 |
| 36 | 4000 | -9 | 0 | -4 | -13.5 |
| 37 | 5000 | -10 | 0 | -8 | -18 |
| 38 | 6300 | -10 | 1 | 4 | -5 |

white noise was passed through the compression amplifier and then through the multifilter.[5] The multifilter attenuator setting were set as noted in Table 20, column (a), these values correcting for the bandwidth effect and spectral variation of the one-third octave band filters, as described above. The output for one-third octave bands was then measured with a voltmeter, yielding the relative intensities shown in Table 20, column (b). These values were the correction factors necessary, by one-third octaves, to flatten the signal as it passed through the compression amplifier.

Correction for Earphone

With the frequency range of the compression amplifier set at 250 Hz to 6000 Hz, at a compression ratio of 1:1, the white noise was passed through the compression amplifier and then the multifilter, the latter with its attenuator settings to correct for the combined effect of the filter bandwidth and the compression amplifier response, i.e. values of Table 20, columns (a) + (b). The output for one-third octave bands was then measured at the earphone, in a 6cc coupler, yielding the relative intensities shown in Table 20, column (c). These values were a direct measure of the effect of the

[5] Since the skirts of the compression amplifier filters had a shallower rate of attenuation than the multifilter, the multifilter's frequency range was adjusted to be just beyond the frequency range of the compression amplifier, i.e. 200 Hz to 6300 Hz. Thus, although the frequency range of the hearing aid was nominally from 250 Hz to 6000 Hz, effectively, it was from the one-third octave band centered at 200 Hz to the one-third octave band centered at 6300 Hz.

earphone. A composite correction was then obtained for the entire system and compared to the sum of the individual corrections. The composite correction, which equalled the sum of the individual corrections, is presented in Table 20, column (d).

Table 20 summarizes each of the effects for which compensation was necessary in order to yield an output of filtered noise that had equal levels at each one-third octave band. In practice, one need go only to the end effect (Table 20, column (d)), but the separate effects, as described, were measured in order to determine the relative effect of each part of the system.

In summary, when wide band noise was passed through the experimental apparatus, it yielded one-third octave bands of noise that were equal in level as measured in a standard 6cc coupler. A tape recording was made of this frequency shaped noise. This tape, referred to as the calibrated noise tape, was used for LDL measurements, described below.

## Procedure

Before the experiment itself was begun, preliminary information consisting of baseline audiometric data was obtained. The experiment then consisted of two stages. These were determination of Loudness Discomfort Levels, and comparison of compression versus non-compression conditions. The acquisition of the preliminary information, and the stages of the experiment are described below.

Baseline Audiometric Data

The first session for all subjects involved a diagnostic evaluation, detailed in Chapter IV. Subjects then reported for six additional two-hour sessions, during which the data for the main experiment was obtained. During the first of these subsequent sessions, air-conduction thresholds at 250 Hz through 4000 Hz and Speech Reception Thresholds were remeasured to determine the stability of the subjects' hearing losses. All measures were within 5 dB of the values obtained in the diagnostic evaluation (given in Tables 15 and 17).

Determination of Loudness Discomfort Levels

During the first session, LDLs for one-third octave bands of noise were measured for each subject using the calibrated noise tape, described above. Instructions to the subjects were the same as those used in measuring LDL for speech and noise in the baseline session, and are given in Chapter IV. Measurements were obtained for one-third octave bands from 200 Hz through 6300 Hz in steps of one octave. LDLs for the intervening one-third octave bands were obtained by linear interpolation. Using the LDL measured at the third octave band centered at 400 Hz as the reference, the relative gain needed for each octave band of the calibrated noise tape to reach LDL was calculated. The frequency-gain function thus derived was referred to as the LDL frequency response curve (for flat noise), and was used in shaping the speech spectrum for several of the experimental conditions, as described below. Note that the LDL in any given third octave band is equal to the LDL in the 400 Hz band plus the value of the LDL frequency response curve for that one-third octave.

LDLs obtained for each one-third octave band represent, to a first approximation, the LDL for a broad band stimulus as a function of third octave bands (this argument is discussed in Chapter VII). However, the measurements of LDL were made using the calibrated noise tape, which had a flat one-third octave spectrum. Speech, which was the test stimulus used, does not have a flat one-third octave spectrum. A second set of adjustments was therefore determined so as to bring the speech spectrum up to the LDL level in each one-third octave band.

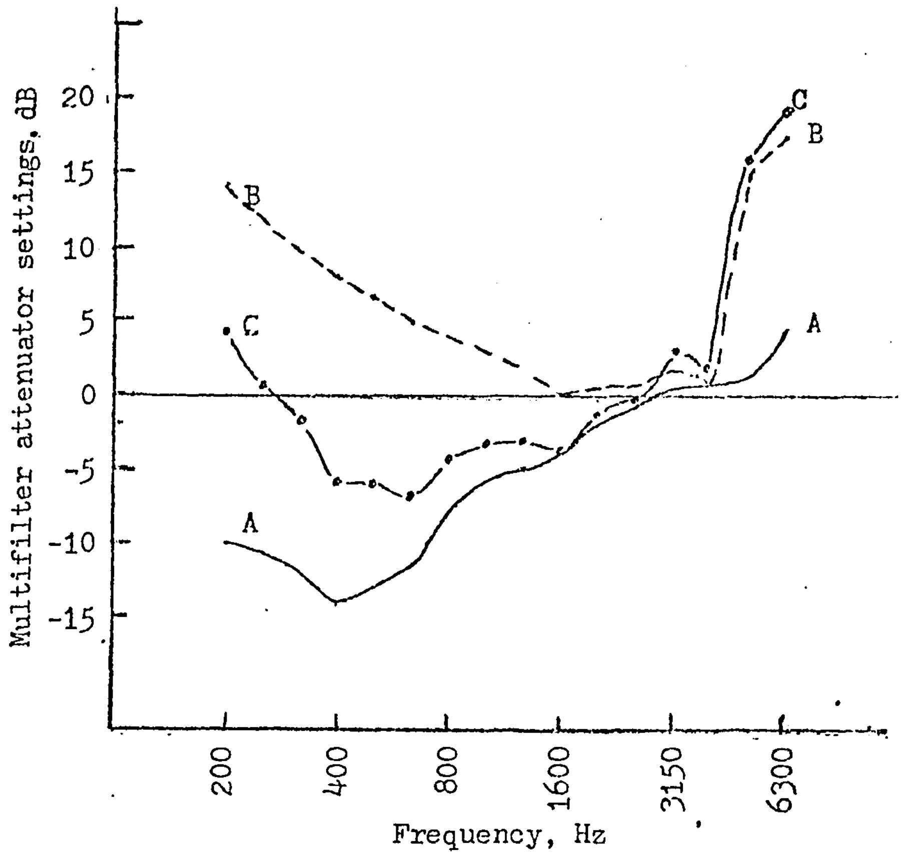
There are many ways in which the distribution of speech energy over the frequency range may be expressed. Dunn and White (1940) made several types of measurements using "normal conversational speech" at 30 cm. from the mouth for all measurements. Using bands one octave wide below 500 Hz, and a half-octave wide above that frequency, they measured the peak sound pressures in each of these bands, up to 12,000 Hz, for five male and five female speakers. The data of the female speakers were of direct interest to the present study. A 1/8th second averaging time was used in these measurements. They then calculated those band levels, as a function of frequency, that were exceeded for various percentages of time. Of interest for our purposes were those band levels which were exceeded 10% of the time by the speech peaks. Using these 10% peak pressure levels, the relative gain needed for each third octave band to flatten the speech spectrum was calculated, with the band centered at 400 Hz as the reference (note that the spectral flattening required is essentially the same for third- or half-octave band measurements). The values obtained are given in Table 21. These values were added to each subject's LDL frequency response curve. The end result was thus a frequency response curve in which each one-third octave band of the speech spectrum was at the individual subject's LDL. The speech was then presented to the subject at 5 dB less than his LDL for wide band noise. This procedure maximized the amount of amplification at each frequency at the subject's ear, by placing all frequencies at a constant below their respective LDLs.

Because of the complexity of the calculations involved, an example of the effect of flattening of the speech spectrum on the frequency response is presented in Figure 23. In this figure, the corrections necessary to flatten the speech spectrum are shown (curve A), which are then added to the LDL frequency response curve (for flat noise) for subject MS (curve B). The net effect was the multifilter attenuator settings required to produce LDL for typical speech at the ear for subject MS (curve C). Curve C is referred to as the LDL frequency response curve for speech

Table 21. Values, in relative dB, derived from the 10% peak pressure levels, at mid-frequencies of one-octave width below 500 Hz, and one-half octave width above 500 Hz (from Dunn & White, 1940). The values shown represent those necessary to flatten the speech spectrum, and were added to each subject's LDL frequency response curve.

| Mid-frequency | Values derived from 10% peak pressure levels, relative dB |
|---|---|
| 200 | 4 |
| 250 | 3* |
| 315 | 2* |
| 400 | 0 |
| 500 | 1* |
| 630 | 2 |
| 800 | 6 |
| 1000 | 8* |
| 1250 | 9 |
| 1600 | 10* |
| 2000 | 12* |
| 2500 | 13 |
| 3150 | 15 |
| 4000 | 15* |
| 5000 | 15 |
| 6300 | 19 |

*these values are interpolated




KEY

A—— Multifilter settings required to flatten speech spectrum. Settings are given as relative values, derived from Table 22 (from Dunn & White, 1940). For practical purposes, the 400 Hz band has been arbitrarily set to -14 on the multifilter.

B--- Relative settings required to produce LDL at the ear for calibrated noise tape, i.e. LDL frequency response curve, uncorrected for speech spectrum, for subject MS.

C•—• Multifilter settings required to produce LDL for typical speech at the ear, for subject MS, i.e. speech spectrum at LDL for all frequencies. This curve is equal to the sum of A + B. The levels are relative, with the overall gain determined by the subject's wide-band LDL for speech. This curve is the LDL frequency response curve for speech.

Fig. 23. Example of the effect of "flattening" of speech spectrum on LDL frequency response curve, for subject MS.

for subject MS. (Henceforth, the term LDL frequency response curve implies the curve for speech.) As mentioned above, the speech was then presented to the subject at 5 dB less than his LDL for wide band noise.

The LDL frequency response curves for the four subjects, with the applied corrections for the speech spectrum, and the subjects' pure tone thresholds, are given in Appendix H.

Comparison of Compression v. Non-Compression Conditions

At the start of each test session, the LDL for speech, using taped W-22 word lists, was measured for each subject under the following conditions (the procedure used to measure LDL is described in Chapter IV):

| | |
|---|---|
| Frequency Response | 1. 0 dB/octave slope |
| | 2. LDL frequency response curve |
| Frequency Range | 1. 250 - 6000 Hz ($L_b$) |
| | 2. 250 - 1500 Hz ($L_1$) |
| | 3. 1500 - 6000 Hz ($L_2$) |

Note that the cross-over frequency was 1500 Hz.

Separate LDL measurements were obtained for each of the 2 x 3, i.e. 6, combinations of the above two factors. The measurements obtained for each of the four subjects are given in Appendix I. The strategy used for determination of the level of test administration to the subjects, i.e. LDL - 5 dB, was based on these six LDL measurements, and is described in Chapter IV, pages 121, 128.

For all experimental conditions, a 400 Hz calibration tone, corresponding to the average peaks of speech, was adjusted so that the power of the speech peak was at the determined level. When noise was also used, the 400 Hz calibration tone on the noise track was adjusted so that the peak power was at the determined lvel.

For peak clipped conditions, the input intensity was gradually increased until the first sign of peak clipping at the output could be observed on an oscilloscope. This intensity level was considered the threshold of peak clipping. The intensity was then raised to the threshold of peak clipping +10 dB.

The experimental consisted of two overlapping factorial designs, using the format shown in Table 22 (although there were multiple test conditions, the design was not fully balanced because of time constraints). The first design consisted of three factors: Amplitude Mode, which included non-compression amplification, both with and without peak clipping, and single-channel compression amplification; Frequency Response, which included a 0 dB/octave slope, and the LDL frequency response curve, the latter with correction for the shape of the speech spectrum; and Listening Mode, which included quiet, and noise at a signal-to-noise (S/N) ratio of 10 dB. There were a total of 3 x 2 x 2, i.e. 12 conditions, with four replications per condition. Table 23 summarizes the variables in the first design.

Table 22. Experimental conditions considered in the main experiment.

| AMPLITUDE MODE | LISTENING MODE | FREQUENCY RESPONSE |
|---|---|---|
| Non-compression, no peak clipping | Quiet | 0 dB/octave<br>LDL frequency response |
| | S/N = 10 dB | 0 dB/octave<br>LDL frequency response |
| Non-compression, with peak clipping | Quiet | 0 dB/octave<br>LDL frequency response |
| | S/N = 10 dB | 0 dB/octave<br>LDL frequency response |
| Compression, single channel<br>Compression ratio: 3:1 | Quiet | 0 dB/octave<br>LDL frequency response |
| | S/N = 10 dB | 0 dB/octave<br>LDL frequency response |
| Compression, two channels<br>Compression ratio:<br>3:1, low frequency channel<br>3:1, high frequency channel | Quiet | LDL frequency response |
| | S/N = 10 dB | LDL frequency response |
| Compression, two channels<br>Compression ratio:<br>1.5:1: low frequency channel<br>3:1, high frequency channel | Quiet | LDL frequency response |
| | S/N = 10 dB | LDL frequency response |

Table 23. The experimental variables considered in the first factorial design of the main experiment, consisting of 3 x 2 x 2, i.e. 12, conditions. There were four replications for each condition.

| AMPLITUDE MODE | Compression ratio | Frequency range, Hz | Input level to subject |
|---|---|---|---|
| Non-compression, no peak clipping | Channel 1: 1:1 | Channel 1: 250-6000 | LDL - 5 dB |
| Non-compression, with peak clipping | Channel 1: 1:1 | Channel 1: 250-6000 | LDL - 5 dB |
| Compression, single channel | Channel 1: 3:1 | Channel 1: 250-6000 | LDL - 5 dB |
| FREQUENCY RESPONSE | | | |
| 0 dB/octave | | | |
| LDL frequency response | | | |
| LISTENING MODE | | | |
| Quiet | | | |
| S/N = 10 dB | | | |

The second design consisted of two factors: Amplitude Mode, which included non-compression amplification, both with and without peak clipping, and single- and two-channel compression amplification; and Listening Mode, which included quiet, and noise at a S/N ratio of 10 dB. There were a total of 5 x 2, i.e. 10 conditions, with four replications per condition (Frequency Response was not a factor, since only the LDL frequency response curve was used for all conditions.) Table 24 summarizes the variables in the second design.

Test Equipment

The apparatus used to deliver the signals to the subjects in the main experiment was similar to that of the pilot study, with several changes: the speech and noise (if noise was being used for the specific experimental condition) were passed from a tape recorder/reproducer to two attenuators that allowed independent intensity control for each track. The speech and noise were then mixed, and processed by the compression amplifier. The input to the compression amplifier was monitored by a voltmeter. The 400 Hz calibration tone on each track of each test tape was adjusted to the reference level of 100 mv. From the compression amplifier, the signals passed through an attenuator, and then to a multifilter. From there, they passed through another attenuator, and then to a Grason-Stadler 1701 audiometer. The input to the audiometer was visually monitored on an

Table 24. The experimental variables considered in the second factorial design of the main experiment, consisting of 5 x 2, i.e. 10, conditions. Frequency Response was not a variable, since LDL frequency response was used for all conditions. There were four replications for each condition.

| AMPLITUDE MODE | Compression ratio | Frequency range, Hz | Input level to subject |
|---|---|---|---|
| Non-compression, no peak clipping | Channel 1: 1:1 | Channel 1: 250-6000 | LDL - 5 dB |
| Non-compression, with peak clipping | Channel 1: 1:1 | Channel 1: 250-6000 | LDL - 5 dB |
| Compression | Channel 1: 3:1 | Channel 1: 250-6000 | LDL - 5 dB |
| Compression | Channel 1: 3:1 | Channel 1: 250-1500 | LDL - 5 dB |
| | Channel 2: 3:1 | Channel 2: 1500-6000 | |
| Compression | Channel 1: 1.5:1<br>Channel 2: 3:1 | Channel 1: 250-1500<br>Channel 2: 1500-6000 | LDL - 5 dB |
| LISTENING MODE | | | |
| Quiet | | | |
| S/N = 10 dB | | | |

oscilloscope. All test equipment through this point was in the outer room of a two-room double-walled IAC test booth. The subject was seated in the inner room of the test booth, and wore TDH-39 earphones set in MX 41/AR cushions. Both speech and noise were presented from the same earphone, placed on the test ear. A block diagram of the apparatus used for testing is shown in Figure 23.

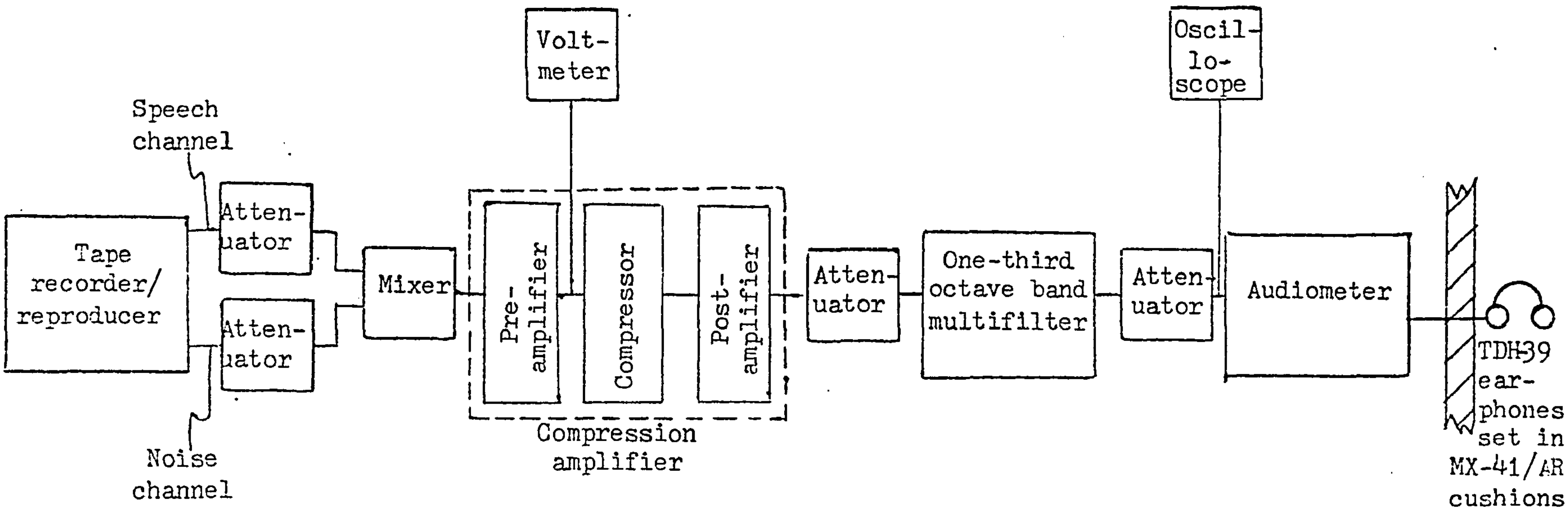



Fig. 24. Block diagram of apparatus used for speech reproduction through the multichannel hearing aid, in the main experiment.

# CHAPTER VI

## RESULTS OF THE MAIN EXPERIMENT

As discussed in Chapter V, the experimental conditions were organized into two overlapping factorial designs. Analyses of variance were performed on the data obtained. In the first factorial design, the main factors in the analysis of variance were Amplitude Mode, Frequency Response, Listening Mode, Subjects, Subtests, and Replications. In the second factorial design, the main factors in the analysis of variance were Amplitude Mode, Listening Mode, Subjects, Subtests, and Replications. In these analyses, the proportions were transformed to inverse sine units.[6] Tables 25 and 26 show the respective results of these analyses.

In the first factorial design, consisting of six factors, statistically significant effects were found at the .001 level for Amplitude Mode, Frequency Response, Listening Mode, Subjects and Subtests, and at the .05 level for Replications. There were significant inter-

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[6] The transformation used was $y = 2 \arcsin\sqrt{p/100}$, where p is the score as the percent correct, and the transformed variable, y, is a measure of the score in inverse sine units. Whereas the variance of p is a function of the value of p, the variance of y is roughly constant, and is approximately 1/n, where n is approximately the total number of syllables in each subtest (Brownlee, 1965).

Table 25. Analysis of variance of data obtained in first factorial design, i.e. for single-channel conditions (both non-compression and compression) for flat frequency response and LDL frequency response.

| Source of variation | Sums of squares | Degrees of freedom | Mean Squares | F-ratio |
|---|---|---|---|---|
| Amplitude Mode (A) | 3.61 | 2 | 1.805 | 20.96 ** |
| Frequency Response (B) | 2.21 | 1 | 2.210 | 25.65 ** |
| Listening Mode (C) | 58.17 | 1 | 58.179 | 675.34 ** |
| AB interaction | 0.06 | 2 | 0.030 | 0.35 |
| AC interaction | 0.28 | 2 | 0.144 | 1.68 |
| BC interaction | 0.04 | 1 | 0.049 | 0.57 |
| ABC interaction | 0.57 | 2 | 0.289 | 3.36 * |
| | | | | |
| Subjects (S) | 18.39 | 3 | 6.130 | 71.16 ** |
| AS interaction | 1.11 | 6 | 0.186 | 2.16 |
| BS interaction | 3.10 | 3 | 1.036 | 12.03 ** |
| CS interaction | 1.89 | 3 | 0.633 | 7.35 ** |
| ABS interaction | 0.67 | 6 | 0.112 | 1.30 |
| ACS interaction | 0.39 | 6 | 0.065 | 0.76 |
| BCS interaction | 0.41 | 3 | 0.139 | 1.61 |
| ABCS interaction | 0.30 | 6 | 0.051 | 0.59 |
| | | | | |
| Modules or subtests (M) | 37.66 | 3 | 12.556 | 145.75 ** |
| AM interaction | 1.16 | 6 | 0.194 | 2.26 |
| BM interaction | 0.05 | 3 | 0.018 | 0.21 |
| CM interaction | 2.10 | 3 | 0.701 | 8.14 ** |
| ABM interaction | 1.52 | 6 | 0.254 | 2.95 * |
| ACM interaction | 0.19 | 6 | 0.032 | 0.37 |
| BCM interaction | 0.15 | 3 | 0.052 | 0.61 |
| ABCM interaction | 1.03 | 6 | 0.173 | 2.00 |
| | | | | |
| Replications | 1.46 | 3 | 0.489 | 5.68 * |
| AR interaction | 1.00 | 6 | 0.167 | 1.94 |
| BR interaction | 0.95 | 3 | 0.319 | 3.71 * |
| CR interaction | 0.04 | 3 | 0.013 | 0.15 |
| ABR interaction | 1.00 | 6 | 0.167 | 1.94 |
| ACR interaction | 0.88 | 6 | 0.147 | 1.70 |
| BCR interaction | 0.54 | 3 | 0.182 | 2.11 |
| ABCR interaction | 0.47 | 6 | 0.079 | 0.92 |

*significant at 0.05 level
**significant at 0.001 level

Table 25--continued

| Source of variation | Sums of squares | Degrees of freedom | Mean squares | F-ratio |
|---|---|---|---|---|
| SM interaction | 7.46 | 9 | 0.829 | 9.63 ** |
| ASM interaction | 2.99 | 18 | 0.166 | 1.93 * |
| BSM interaction | 1.93 | 9 | 0.214 | 2.49 * |
| CSM interaction | 3.30 | 9 | 0.367 | 4.26 ** |
| ABSM interaction | 1.49 | 18 | 0.083 | 0.96 |
| ACSM interaction | 2.81 | 18 | 0.156 | 1.81 * |
| BCSM interaction | 1.99 | 9 | 0.222 | 2.57 * |
| ABCSM interaction | 0.83 | 18 | 0.046 | 0.53 |
| SR interaction | 1.22 | 9 | 0.136 | 1.58 |
| ASR interaction | 1.83 | 18 | 0.101 | 1.18 |
| BSR interaction | 0.48 | 9 | 0.053 | 0.62 |
| CSR interaction | 0.80 | 9 | 0.089 | 1.04 |
| ABSR interaction | 2.03 | 18 | 0.113 | 1.31 |
| ACSR interaction | 1.32 | 18 | 0.073 | 0.85 |
| BCSR interaction | 0.92 | 9 | 0.103 | 1.19 |
| ABCSR interaction | 2.66 | 18 | 0.147 | 1.71 |
| MR interaction | 0.59 | 9 | 0.066 | 0.76 |
| AMR interaction | 2.48 | 18 | 0.138 | 1.60 |
| BMR interaction | 0.45 | 9 | 0.050 | 0.59 |
| CMR interaction | 0.33 | 9 | 0.037 | 0.43 |
| ABMR interaction | 1.10 | 18 | 0.061 | 0.71 |
| ACMR interaction | 0.97 | 18 | 0.054 | 0.62 |
| BCMR interaction | 1.12 | 9 | 0.125 | 1.45 |
| ABCMR interaction | 1.39 | 18 | 0.077 | 0.89 |
| MSR interaction | 4.32 | 27 | 0.160 | 1.86 |
| AMSR interaction | 3.94 | 54 | 0.073 | 0.84 |
| BMSR interaction | 3.30 | 27 | 0.122 | 1.42 |
| CMSR interaction | 2.42 | 27 | 0.089 | 0.43 |
| ABMSR interaction | 3.93 | 54 | 0.072 | 0.84 |
| ACMSR interaction | 5.05 | 54 | 0.093 | 1.08 |
| BCMSR interaction | 1.47 | 27 | 0.054 | 0.63 |
| ABCMSR | 4.65 | 54 | 0.086 | |
| Total | 213.34 | 767 | | |

*significant at0.05 level
**significant at 0.001 level

Table 26. Analysis of variance of data obtained in second factorial design, i.e. for single-channel conditions (both non-compression and compression) and two-channel conditions (compression only), with LDL frequency response.

| Source of variation | Sums of squares | Degrees of freedom | Mean squares | F-ratio |
|---|---|---|---|---|
| Amplitude Mode (A) | 1.59 | 4 | 0.397 | 5.21 ** |
| Listening Mode (C) | 51.07 | 1 | 51.075 | 669.52 ** |
| AC interaction | 0.59 | 4 | 0.148 | 1.94 |
| Subjects (S) | 9.28 | 3 | 3.093 | 40.55 ** |
| AS interaction | 1.52 | 12 | 0.127 | 1.67 |
| CS interaction | 1.36 | 3 | 0.455 | 5.97 ** |
| ACS interaction | 2.40 | 12 | 0.200 | 2.62 * |
| Modules or subtests (M) | 32.07 | 3 | 10.691 | 140.14 ** |
| AM interaction | 3.15 | 12 | 0.262 | 3.44 ** |
| CM interaction | 3.01 | 3 | 1.005 | 13.18 ** |
| ACM interaction | 2.59 | 12 | 0.216 | 2.83 * |
| Replications (R) | 3.98 | 3 | 1.329 | 17.42 ** |
| AR interaction | 1.64 | 12 | 0.137 | 1.80 |
| CR interaction | 0.67 | 3 | 0.223 | 2.92 * |
| ACR interaction | 1.12 | 12 | 0.093 | 1.22 |
| SM interaction | 5.13 | 9 | 0.570 | 7.47 ** |
| ASM interaction | 3.66 | 36 | 0.101 | 1.33 |
| CSM interaction | 2.46 | 9 | 0.273 | 3.58 ** |
| ACSM interaction | 6.50 | 36 | 0.180 | 2.36 ** |
| SR interaction | 0.82 | 9 | 0.091 | 1.20 |
| ASR interaction | 4.32 | 36 | 0.120 | 1.57 * |
| CSR interaction | 0.56 | 9 | 0.062 | 0.81 |
| ACSR interaction | 4.60 | 36 | 0.127 | 1.67 * |
| MR interaction | 0.61 | 9 | 0.068 | 0.90 |
| AMR interaction | 3.87 | 36 | 0.107 | 1.41 |
| CMR interaction | 1.28 | 9 | 0.143 | 1.87 |
| ACMR interaction | 3.36 | 36 | 0.093 | 1.22 |
| SMR interaction | 5.70 | 27 | 0.211 | 2.77 ** |
| ASMR interaction | 9.51 | 108 | 0.088 | 1.15 |
| CSMR interaction | 2.77 | 27 | 0.102 | 1.34 |
| ACSMR interaction | 8.23 | 108 | 0.076 | |
| Total | 179.56 | 639 | | |

*significant at 0.05 level
**significant at 0.001 level

actions at the .001 level between Subjects and the following factors: Frequency Response, Listening Mode, and Subtests; and between Subtests and Listening Mode. Between Replications and Frequency Response, a significant interaction was seen at the .05 level.

At the .001 level, three-way interactions were seen among Subjects, Subtests, and Listening Mode. At the .05 level, three-way interactions were seen among Amplitude Mode, Frequency Response, and both Listening Mode and Subtests. They were also seen among Subjects, Subtests, and both Amplitude Mode and Frequency Response, also at the .05 level.

In the second factorial design, consisting of five factors, statistically significant effects were found at the .001 level for all five factors: Amplitude Mode, Listening Mode, Subjects, Subtests, and Replications. Interactions, significant at the .001 level, were seen between Listening Mode and Subjects; and between Subtests and the following factors: Amplitude Mode, Listening Mode, and Subjects. At the .05 level, the interaction between Listening Mode and Replications was significant.

Also at the .001 level, three-way interactions were seen among Subjects, Subtests, and both Replications and Listening Mode. Three-way interactions, significant at the .05 level, were seen among Amplitude Mode, Listening Mode, and both Subjects and Subtests; and among Amplitude Mode, Subjects and Replications. A four-way interaction was seen among Amplitude Mode, Listening Mode, Subjects, and Subtests, significant at the .001 level.

Tables 27 and 28 give scores, by subject, as a function of Amplitude Mode, Frequency Response, and Listening Mode, for each of four replications. Means and standard deviations are also given.

## The Effect of Amplitude Mode

The data showed that Amplitude Mode, Listening Mode, and Frequency Response were significant in the first factorial design, and that the first two of these variables were also significant in the second factorial design (Frequency Response was not a variable in the latter design). These scores, averaged over both subjects and replications, are shown in Figure 25. The scores are also given in tabular form in Appendix J, Table 30. All conditions in quiet yielded better scores than all conditions in noise, averaging 27% better. LDL frequency response was consistently better for any specific Amplitude Mode than was flat frequency response, averaging 5% better over all conditions.

In quiet, the maximum score obtained for both frequency responses was under the condition of non-compression, without peak clipping (66% for LDL frequency response, 61% for flat frequency response). The poorest score was obtained under the condition of non-compression, with peak clipping (57% for LDL frequency response, 50% for flat

Table 27. Percent correct scores for subjects as a function of Amplitude Mode, Frequency Response, and Listening Mode, for each of four replications. Means and standard deviations are also given. The scores for each Main Effect have been averaged over remaining factors.

| Main Effect | Replication | SUBJECT MS | HS | FS | BB | Row Mean | Row Standard Deviation |
|---|---|---|---|---|---|---|---|
| Non-compression, no peak clipping | 1 | 39 | 59 | 45 | 40 | 45.75 | 9.2 |
| | 2 | 50 | 65 | 43 | 42 | 50.00 | 10.6 |
| | 3 | 48 | 58 | 47 | 36 | 47.25 | 9.0 |
| | 4 | 48 | 71 | 53 | 44 | 54.00 | 11.9 |
| | Column Mean | 46.25 | 63.25 | 47.0 | 40.5 | | |
| | Column Standard Deviation | 4.9 | 6.0 | 4.3 | 3.4 | | |
| Non-compression, with peak clipping | 1 | 33 | 41 | 40 | 34 | 37.00 | 7.1 |
| | 2 | 36 | 48 | 46 | 42 | 43.00 | 5.3 |
| | 3 | 40 | 54 | 41 | 34 | 42.25 | 8.4 |
| | 4 | 35 | 60 | 43 | 31 | 42.25 | 12.8 |
| | Column Mean | 36.0 | 50.75 | 42.5 | 35.25 | | |
| | Column Standard Deviation | 2.9 | 8.1 | 2.6 | 4.7 | | |
| Compression, 3:1 | 1 | 43 | 59 | 42 | 31 | 43.75 | 11.5 |
| | 2 | 44 | 55 | 36 | 41 | 44.00 | 8.0 |
| | 3 | 46 | 67 | 47 | 40 | 50.00 | 11.7 |
| | 4 | 50 | 61 | 45 | 35 | 47.75 | 10.8 |
| | Column Mean | 45.75 | 60.5 | 42.5 | 36.75 | | |
| | Column Standard Deviation | 3.1 | 5.0 | 4.8 | 4.6 | | |
| Flat frequency response | 1 | 39 | 57 | 40 | 32 | 42.00 | 10.6 |
| | 2 | 40 | 57 | 35 | 40 | 43.00 | 9.6 |
| | 3 | 43 | 61 | 37 | 32 | 43.25 | 12.7 |
| | 4 | 42 | 64 | 37 | 30 | 43.25 | 14.7 |
| | Column Mean | 41.0 | 59.75 | 37.25 | 33.5 | | |
| | Column Standard Deviation | 1.8 | 3.4 | 2.1 | 4.4 | | |

Table 27--continued

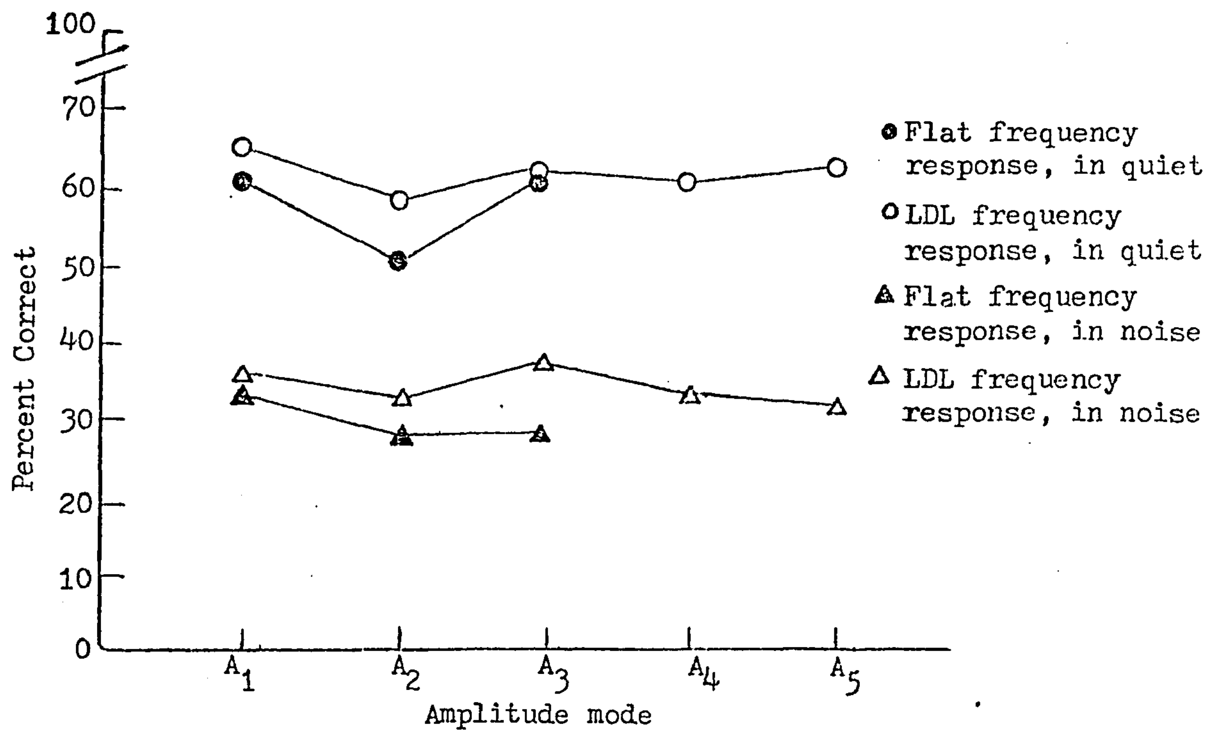
| Main Effect | Replication | SUBJECT MS | HS | FS | BB | Row Mean | Row Standard Deviation |
|---|---|---|---|---|---|---|---|
| LDL frequency response | 1 | 38 | 48 | 44 | 38 | 42.00 | 4.9 |
| | 2 | 47 | 55 | 49 | 43 | 48.50 | 5.0 |
| | 3 | 46 | 58 | 53 | 41 | 48.50 | 7.5 |
| | 4 | 46 | 64 | 57 | 43 | 52.50 | 9.7 |
| | Column Mean | 44.25 | 56.25 | 50.75 | 41.25 | | |
| | Column Standard Deviation | 4.2 | 6.7 | 5.6 | 2.4 | | |
| Quiet | 1 | 47 | 62 | 61 | 51 | 55.25 | 7.4 |
| | 2 | 56 | 66 | 57 | 57 | 59.00 | 4.7 |
| | 3 | 58 | 71 | 59 | 55 | 60.75 | 7.0 |
| | 4 | 54 | 76 | 61 | 53 | 61.00 | 10.6 |
| | Column Mean | 53.75 | 68.75 | 59.5 | 54.0 | | |
| | Column Standard Deviation | 4.8 | 6.1 | 1.9 | 2.6 | | |
| Noise | 1 | 30 | 43 | 25 | 20 | 29.50 | 9.9 |
| | 2 | 31 | 46 | 27 | 27 | 32.75 | 9.0 |
| | 3 | 32 | 48 | 31 | 20 | 32.75 | 11.5 |
| | 4 | 35 | 50 | 33 | 21 | 34.75 | 11.9 |
| | Column Mean | 32.0 | 46.75 | 29.0 | 22.0 | | |
| | Column Standard Deviation | 2.2 | 3.0 | 3.7 | 3.4 | | |

Table 28. Percent correct scores for subjects as a function of Amplitude Mode and Listening Mode, for each of four replications. Frequency Response was LDL frequency response for all conditions. Means and standard deviations are also given. The scores for each Main Effect have been averaged over the remaining factors.

| Main Effect | Replication | SUBJECT MS | HS | FS | BB | Row Mean | Row Standard Deviation |
|---|---|---|---|---|---|---|---|
| Non-compression, no peak clipping | 1 | 34 | 51 | 48 | 46 | 44.75 | 7.5 |
| | 2 | 55 | 61 | 51 | 39 | 51.50 | 9.3 |
| | 3 | 53 | 52 | 61 | 36 | 50.50 | 10.5 |
| | 4 | 51 | 71 | 63 | 47 | 58.00 | 11.5 |
| | Column Mean | 48.25 | 58.75 | 55.75 | 42.0 | | |
| | Column Standard Deviation | 9.6 | 9.3 | 7.4 | 5.4 | | |
| Non-compression, with peak clipping | 1 | 34 | 36 | 44 | 40 | 38.50 | 4.4 |
| | 2 | 45 | 48 | 55 | 42 | 47.50 | 5.6 |
| | 3 | 35 | 52 | 40 | 38 | 41.25 | 7.5 |
| | 4 | 38 | 58 | 57 | 41 | 48.50 | 10.5 |
| | Column Mean | 38.0 | 48.5 | 49.0 | 40.25 | | |
| | Column Standard Deviation | 5.0 | 9.3 | 8.3 | 1.7 | | |
| Compression, 3:1 | 1 | 44 | 58 | 40 | 28 | 42.50 | 12.4 |
| | 2 | 41 | 55 | 42 | 48 | 46.50 | 6.5 |
| | 3 | 51 | 69 | 56 | 49 | 56.25 | 9.0 |
| | 4 | 49 | 62 | 52 | 41 | 51.00 | 8.7 |
| | Column Mean | 46.25 | 61.0 | 47.5 | 41.5 | | |
| | Column Standard Deviation | 4.6 | 6.1 | 7.7 | 9.7 | | |
| Compression, 3:1, 3:1 | 1 | 33 | 59 | 30 | 33 | 38.75 | 13.6 |
| | 2 | 48 | 61 | 48 | 49 | 51.50 | 6.4 |
| | 3 | 41 | 60 | 55 | 39 | 48.75 | 10.3 |
| | 4 | 46 | 59 | 52 | 31 | 47.00 | 11.9 |
| | Column Mean | 42.0 | 59.75 | 46.25 | 38.0 | | |
| | Column Standard Deviation | 6.7 | 1.0 | 11.2 | 8.1 | | |

Table 28—continued

| Main Effect | Replication | SUBJECT MS | HS | FS | BB | Row Mean | Row Standard Deviation |
|---|---|---|---|---|---|---|---|
| Compression, 1.5:1, 3:1 | 1 | 39 | 43 | 44 | 36 | 40.50 | 3.7 |
| | 2 | 36 | 58 | 47 | 48 | 47.25 | 9.0 |
| | 3 | 50 | 57 | 52 | 33 | 48.00 | 10.4 |
| | 4 | 52 | 54 | 52 | 47 | 51.25 | 3.0 |
| | Column Mean | 44.25 | 53.0 | 48.75 | 41.0 | | |
| | Column Standard Deviation | 7.9 | 6.9 | 3.9 | 7.6 | | |
| Quiet | 1 | 47 | 63 | 59 | 55 | 56.00 | 6.8 |
| | 2 | 56 | 66 | 68 | 61 | 62.75 | 5.4 |
| | 3 | 59 | 72 | 69 | 58 | 64.50 | 7.0 |
| | 4 | 58 | 71 | 66 | 55 | 62.50 | 7.3 |
| | Column Mean | 55.0 | 68.0 | 65.5 | 57.25 | | |
| | Column Standard Deviation | 5.5 | 4.2 | 4.5 | 2.9 | | |
| Noise | 1 | 28 | 36 | 24 | 20 | 27.00 | 6.8 |
| | 2 | 34 | 47 | 31 | 30 | 35.50 | 7.9 |
| | 3 | 33 | 44 | 37 | 22 | 34.00 | 9.2 |
| | 4 | 37 | 50 | 44 | 29 | 40.00 | 9.1 |
| | Column Mean | 33.0 | 44.25 | 34.0 | 25.25 | | |
| | Column Standard Deviation | 3.7 | 6.0 | 8.5 | 5.0 | | |




KEY

| | |
|---|---|
| $A_1$ | Non-compression, no peak clipping |
| $A_2$ | Non-compression, with peak clipping |
| $A_3$ | Compression, 3:1 |
| $A_4$ | Compression, 3:1,3:1 |
| $A_5$ | Compression, 1.5:1,3:1 |

Fig. 25. Percent correct scores obtained in both quiet and in noise for each of two frequency responses, and three Amplitude Modes for the first factorial design, and five Amplitude Modes for the second factorial design. Scores are averaged over subject and replication. The values shown are also given in tabular form in Appendix J, Table 30.

frequency response).

In noise, the maximum score for LDL frequency response was obtained with single-channel compression, 3:1 (37%), with the condition of non-compression, no peak clipping being only one percentage point poorer (36%). For flat frequency response, non-compression, no peak clipping elicited the maximum score (33%). The poorest score with LDL frequency response was obtained under the conditions of non-compression, with peak clipping, and two-channel compression, 1.5:1, 3:1 (31% for both conditions). For flat frequency response, both non-compression, with peak clipping and single-channel compression, 3:1 yielded equally poor scores (27%).

## The Effect of Replication

There was a significant interaction between Replication and Frequency Response in the first factorial analysis, and between Replication and Listening Mode in the second factorial analysis. The results of these two interactions are combined and depicted in Figure 26. These results are also given in tabular form in Appendix J, Table 31. On the first replication, the scores obtained for flat frequency response and LDL frequency response were very close, with a difference of only 1% in quiet and 2% in noise. The data then showed that there was no time-order effect for flat frequency response in quiet or in noise. For LDL frequency response, a time-order effect existed in quiet, and an even greater one in noise. That is, the greatest

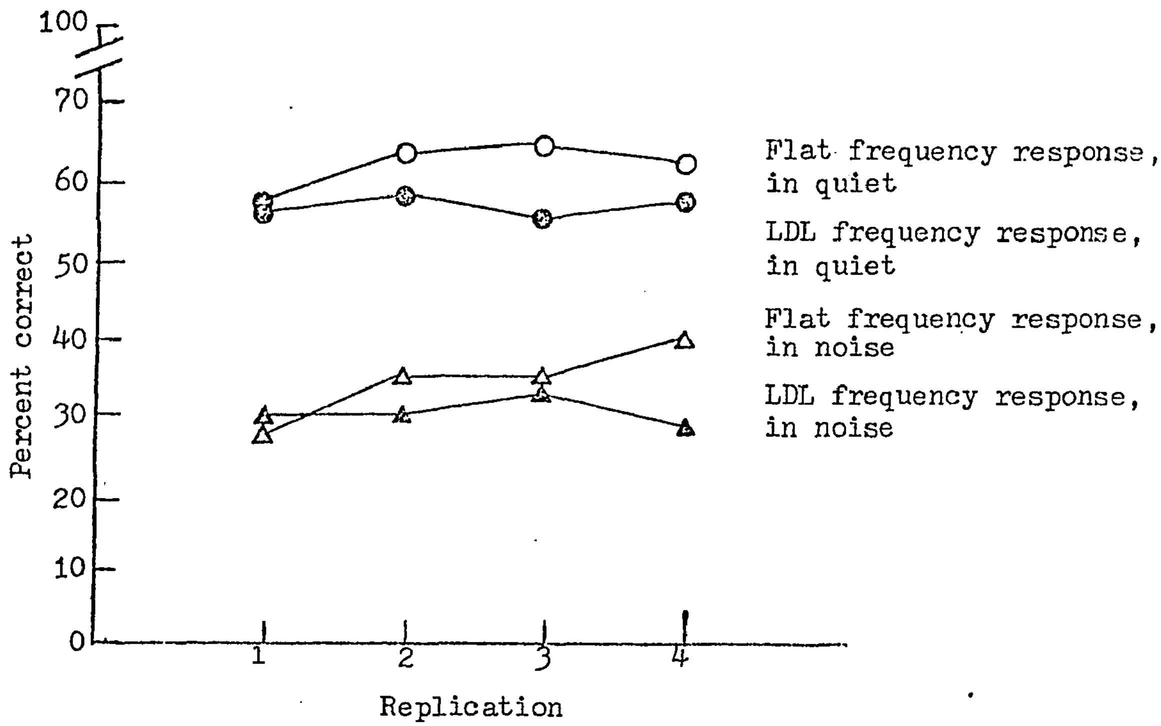



Fig. 26. Percent correct scores obtained in both quiet and in noise for each of two frequency responses and four replications. Scores are averaged over subjects and Amplitude Mode (three Amplitude Modes for flat frequency response, and five Amplitude Modes for LDL frequency response). The values shown are also given in tabular form in Appendix J, Table 31.

increment in score was always seen between the first and the second replications, with the magnitude of this increment being 8% in both quiet and in noise. Then, in quiet, there was no further increase in score, but in noise there was an additional increase of 5% to the final (fourth) replication.

## Individual Differences Among Subjects

There were several significant two- and three-way interactions involving subjects. Some of these interactions, involving Listening Mode, Amplitude Mode, Frequency Response, and Replication, are shown in Figures 27 through 30. The results are also given in tabular form in Appendix J, Tables 32 through 35, respectively. These data showed significant individual differences with respect to the joint effects of Amplitude Mode and Listening Mode. In quiet, for flat frequency response, non-compression, with peak clipping yielded the poorest scores for all subjects. With LDL frequency response, in quiet, this was true for only two subjects, with the other two scoring poorest in the two-channel compression condition, 3:1, 3:1.

In noise, there was variation again: for flat frequency response, two of the four subject scored worst with peak clipping, and the other two with one-channel compression, 3:1. For LDL frequency response, only one subject scored poorest in the non-compression, with peak clipping condition, with the other three yielding their poorest scores with two-channel compression. One must, however,

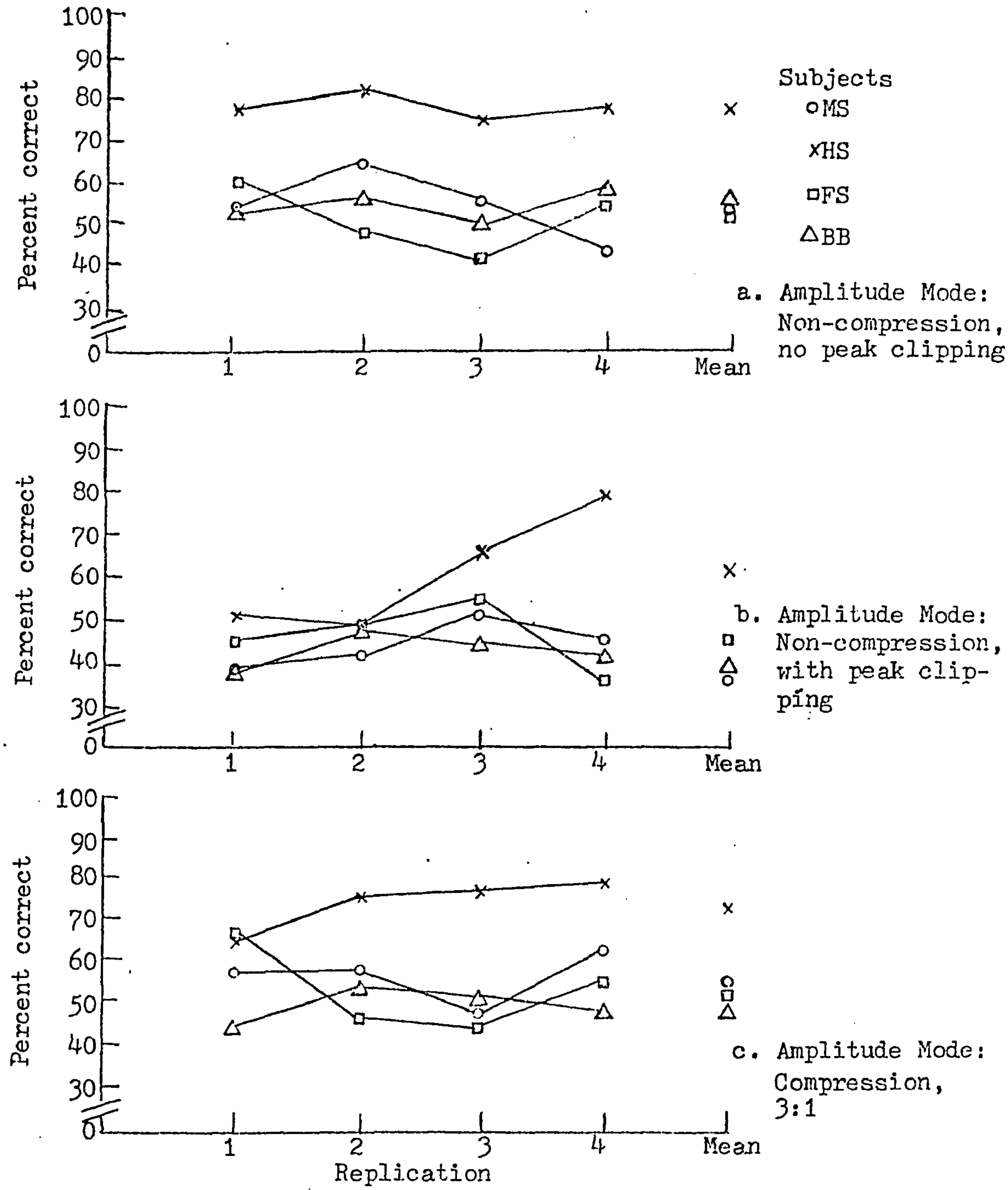



Fig. 27. Percent correct scores obtained in quiet with flat frequency response as a function of Amplitude Mode, replication, and subject. Mean scores are also shown. The values shown are also given in tabular form in Appendix J, Table 32.

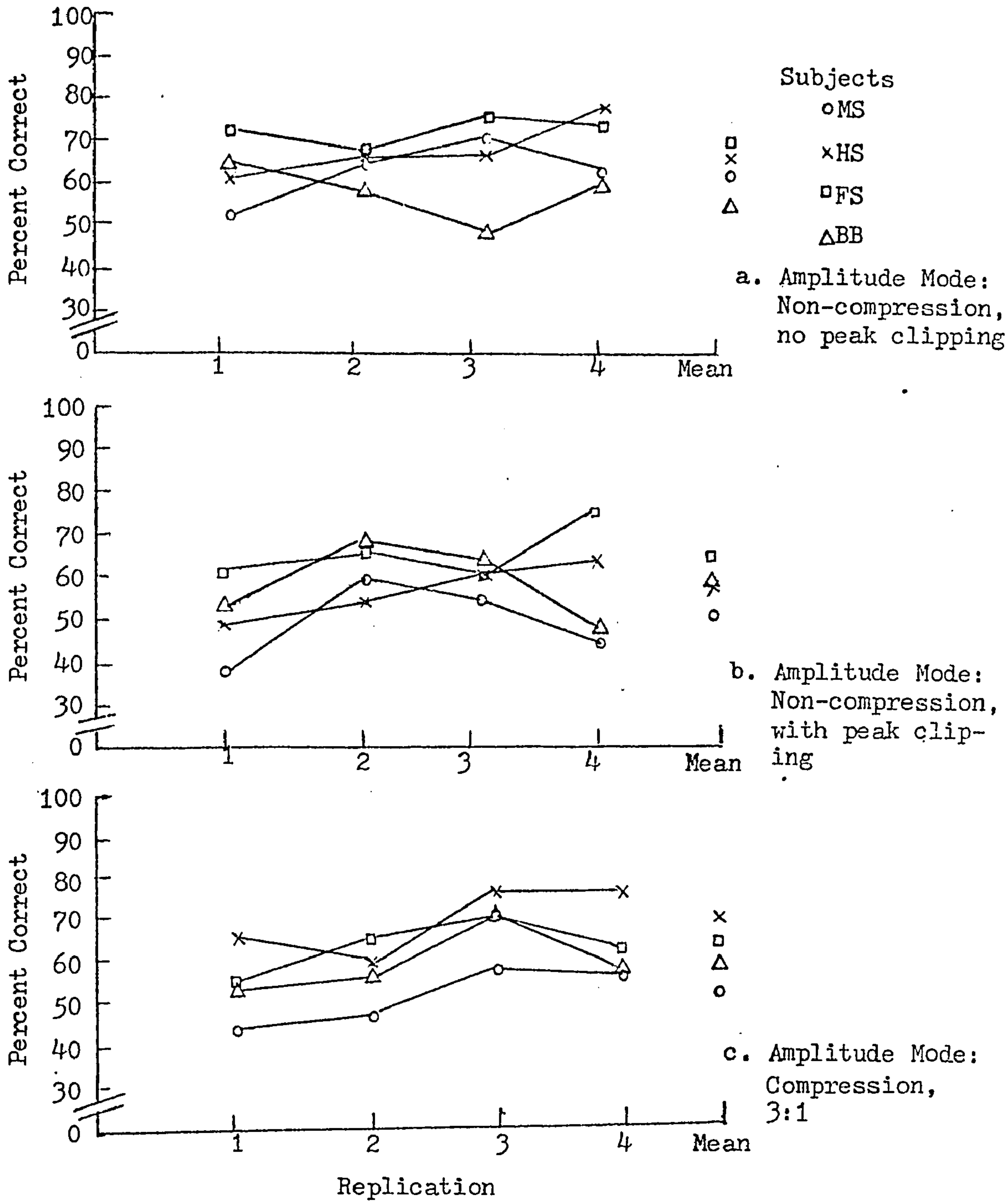
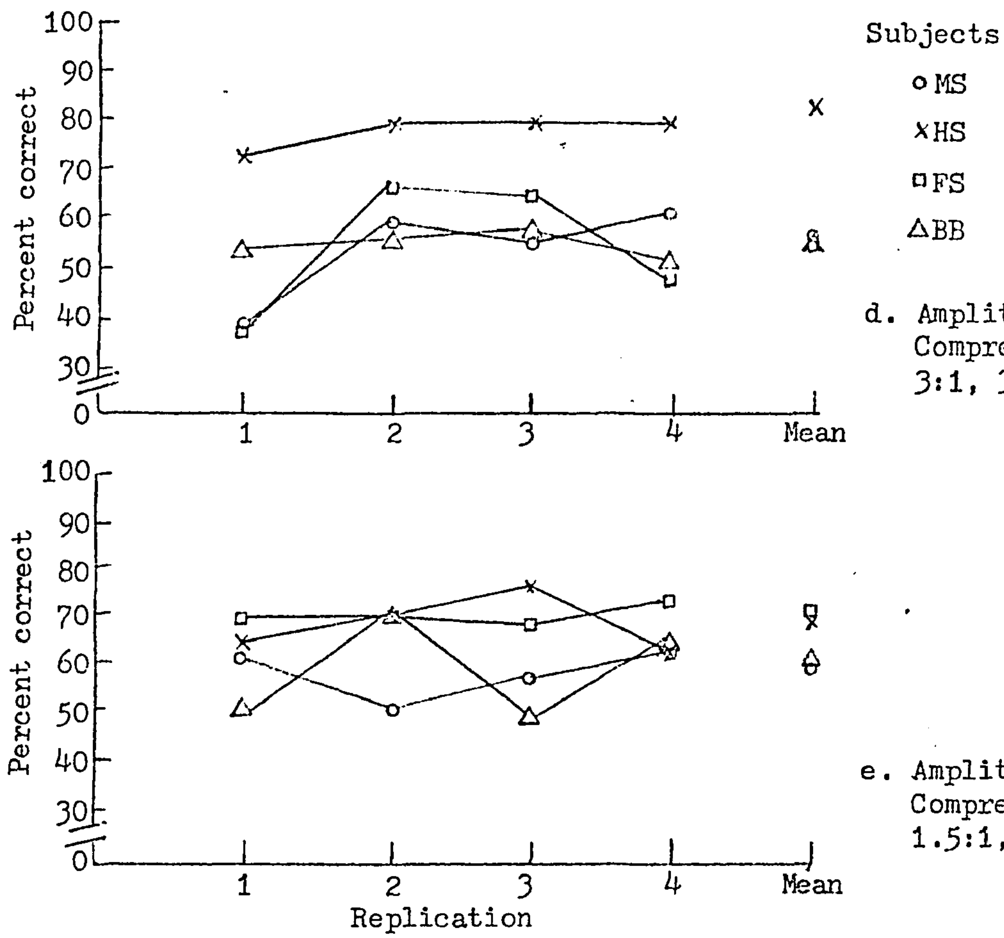



Fig. 28. Percent correct scores obtained in quiet with LDL frequency response as a function of Amplitude Mode, replication, and subject. Mean scores are also shown. The values shown are also given in tabular form in Appendix J, Table 33.

Fig. 28--continued




d. Amplitude Mode: Compression, 3:1, 3:1

e. Amplitude Mode: Compression, 1.5:1, 3:1

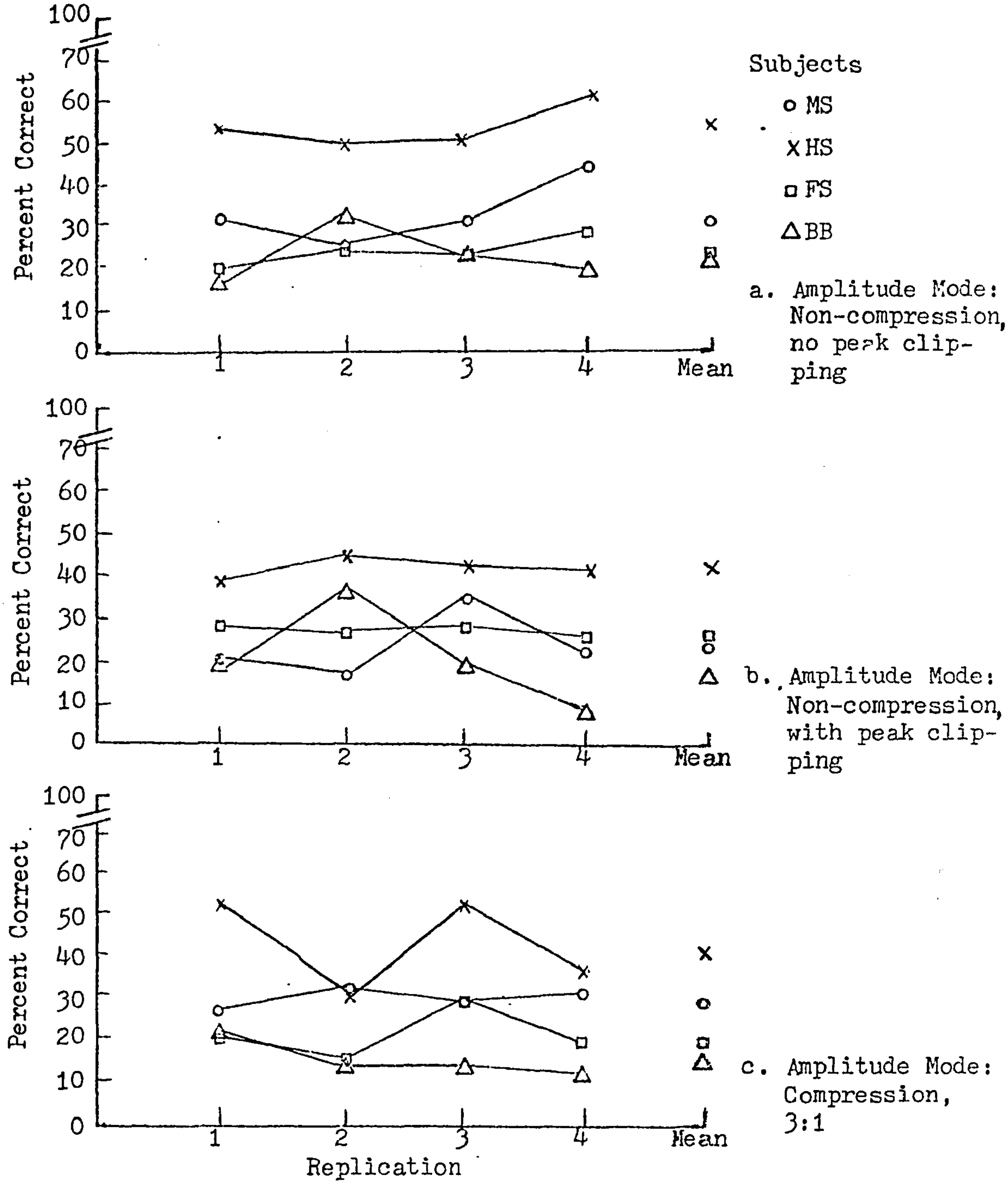



Fig. 29. Percent correct scores obtained in noise with flat frequency response as a function of Amplitude Mode, replication, and subject. Mean scores are also shown. The values shown are also given in tabular form in Appendix J, Table 34.

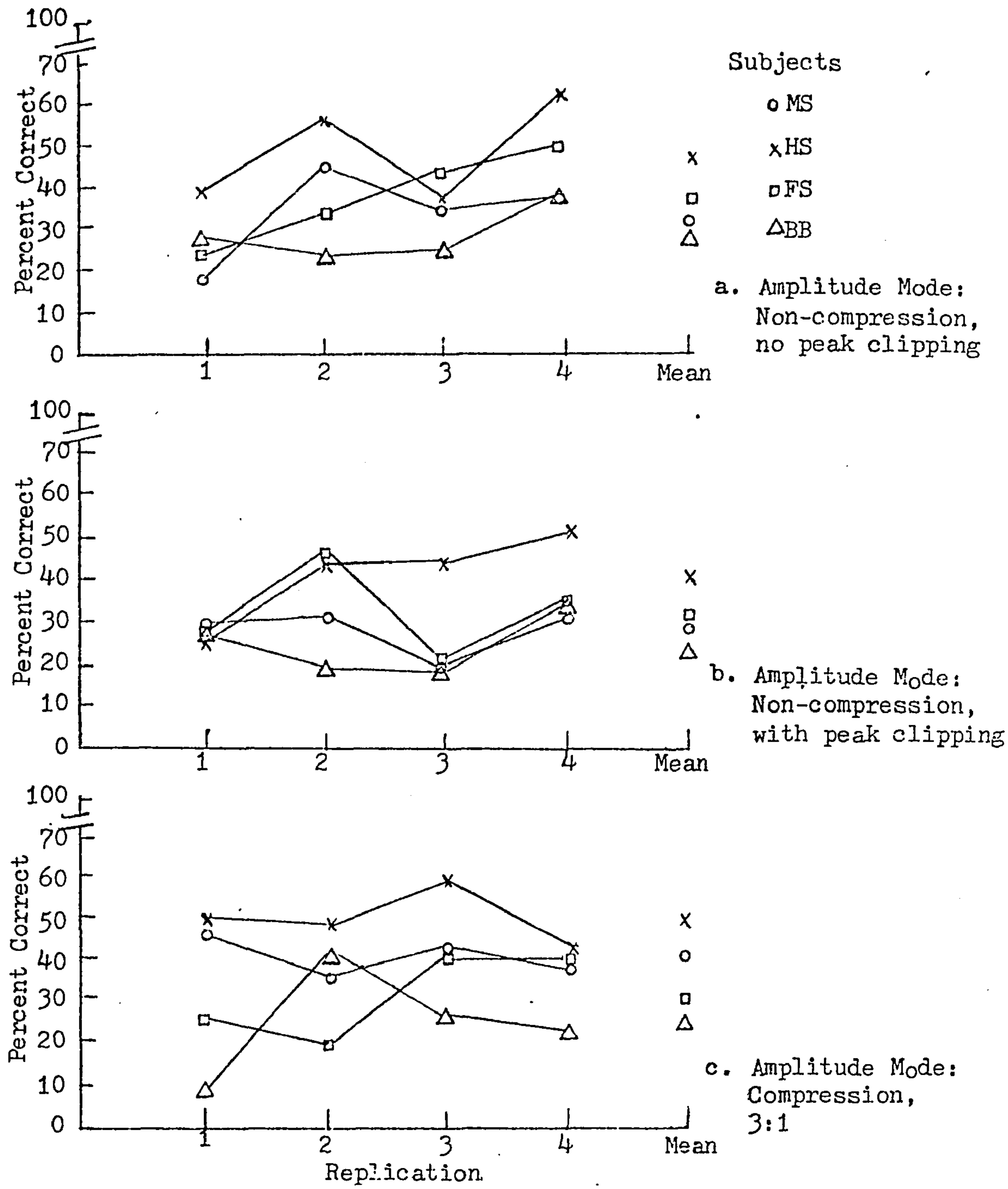
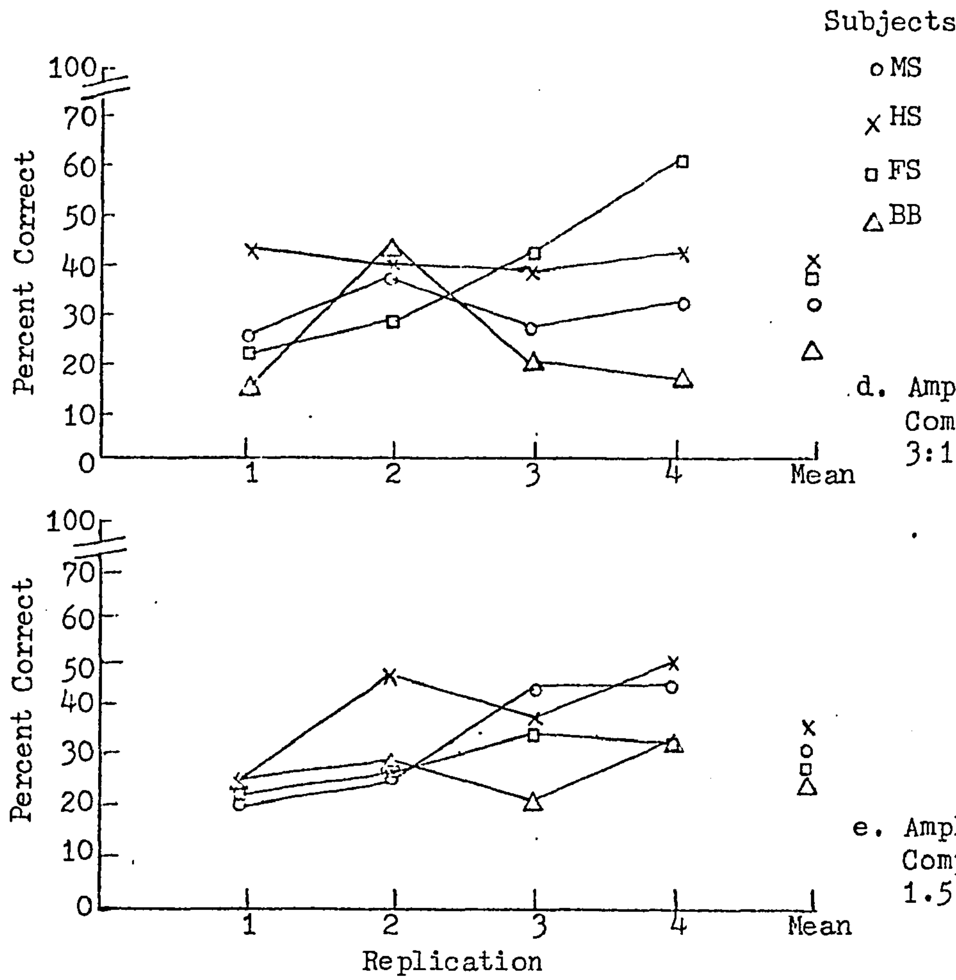



Fig. 30. Percent correct scores obtained in noise with LDL frequency response as a function of Amplitude Mode, replication, and subject. Mean scores are also shown. The values shown are also given in tabular form in Appendix J, Table 35.

Fig. 30--continued




d. Amplitude Mode: Compression, 3:1, 3:1

e. Amplitude Mode: Compression, 1.5:1, 3:1

take into account the following: since the design used was actually two overlapping factorial designs, it was not completely balanced for all factors. That is, there were no two-channel compression conditions with flat frequency response. For flat frequency response there were therefore three Amplitude Modes, whereas for LDL frequency response there were five. Thus, in looking at which Amplitude Mode yielded poorest (or best) scores, one is not looking at a completely balanced design. The decision regarding those conditions that were to be left out was based on earlier data obtained in the pilot study, and were those that were not expected to yield among the highest scores.

The individual differences also showed up for the conditions that yielded the best scores. For example, FS's best score in noise with LDL frequency response was with the Amplitude Mode of compression, 3:1, 3:1. In quiet, also with LDL frequency response, the reverse was true: the Amplitude Mode of compression, 3:1, 3:1 yielded her poorest scores. However, although there were individual exceptions, performance with Amplitude Mode of non-compression, no peak clipping resulted in best performance consistently more often for more subject than did any other single Amplitude Mode.

In summary, although significant individual differences and interactions were evident, the trend appeared to be that non-compression amplification, without peak clipping yielded the best scores for most subjects, whether in quiet or in noise, and whether with flat frequency response or LDL frequency response. And, again under most conditions for most subjects, non-compression amplification, with peak clipping yielded poorer scores more often than did any other Amplitude Mode.

The data depicted in Figures 27 through 30 also provide information on time-order effects. For example, in quiet with LDL frequency response, the subject starting out with the poorest scores for most conditions on the first replication (MS) also showed the greatest overall improvement from first to fourth replication, averaging 11% over the five Amplitude Modes. In noise, however, this pattern did not emerge. That is, the subject starting out with the poorest scores (BB) did not show the greatest increment in score from first to fourth replication. Averaged over Subjects, time-order effect was not significant for flat frequency response, whether in quiet or in noise. For LDL frequency response there was an average increment in score of 6% in quiet and of 13% in noise, from first to last replication. However, even though a significant time-order effect was evident for the latter condition, there were also great individual fluctuations between replications, with scores not always rising consistently.

In comparing flat frequency response to LDL frequency response, all subjects but one (HS) improved with the latter frequency shaping. For all subjects over all conditions, the improvement averaged 4% in quiet and 6% in noise. The greatest increment was apparent with the Amplitude Mode that generally yielded the poorest score, non-compression amplification, with peak clipping, averaging 12%. Subject HS, whose dynamic range was the smallest of the four subjects (range of SRT to LDL was 20 dB), did worse with LDL frequency response as compared to flat frequency response, by an average of 7%. However, although his overall score was poorer with LDL frequency response, he showed a time-order effect that was greater for this frequency shaping than for the flat frequency response. For flat frequency response, HS's scores improved, on the average, by 6% from first to last replication, but for LDL frequency response, they improved by 15%.

In summary, there were significant increments in score from the first to the fourth replication. These increments varied on an individual basis, and for different conditions. In addition, LDL frequency response yielded significantly higher scores than did flat frequency response, the magnitude of the difference being greater in noise than in quiet.

## Analysis by Subtest

Table 29 shows scores, by subject, for the four subtests of the NST, as a function of Frequency Response, these three variables having interacted significantly. The analysis by subtest revealed that identification was

Table 29. Percent correct scores for each subtest of the Nonsense Syllable Test. Data are shown for each subject, as well as the mean scores across subjects. Scores for flat frequency response are averaged over three Amplitude Modes, and scores for LDL frequency response are averaged over five Amplitude Modes. All scores are averaged over Listening Mode.

| SUBTEST | | | Frequency | SUBJECT | | | | |
|---|---|---|---|---|---|---|---|---|
| Number | | Description | Response | MS | HS | FS | BB | Mean |
| 1 | /uv/ | voiced consonant following /u/ | Flat | 18.7 | 52.3 | 19.9 | 30.9 | 29.7 |
| | | | LDL | 24.0 | 44.8 | 28.6 | 33.1 | 32.4 |
| 2 | /uv̄/ | voiceless consonant following /u/ | Flat | 53.4 | 68.6 | 56.6 | 41.2 | 55.1 |
| | | | LDL | 58.6 | 64.7 | 72.8 | 51.7 | 62.1 |
| 3 | /iv̄/ | voiceless consonant following /i/ | Flat | 51.5 | 74.6 | 44.0 | 34.5 | 51.4 |
| | | | LDL | 51.0 | 61.4 | 53.2 | 44.5 | 52.5 |
| 4 | /iv/ | voiced consonant following /i/ | Flat | 42.0 | 42.9 | 30.6 | 27.8 | 35.7 |
| | | | LDL | 42.7 | 54.0 | 43.0 | 33.2 | 43.1 |

superior for voiceless consonants than for voiced, by approximately 20%. When /u/ or /i/ preceded a voiced consonant, subjects scored approximately 8% better, on the average, for /i/ (subtest 4) than for /u/ (subtest 1). When the vowel preceded a voiceless consonant, subjects scored about 7% better, on the average, for /u/ (subtest 2) than for /i/ (subtest 3). The effect of frequency shaping differed for various subjects. By subtest, it was least for MS, who showed no differential effect between flat frequency response and LDL frequency response for subtests containing /i/, and an improvement of about 5% with LDL frequency response for subtests containing /u/. On the other hand, with LDL frequency response, FS showed improvements of 8-16% for subtests containing /i/, and of 9-12% for subtests containing /u/.

There were several additional significant two- and three-way interactions involving Subtests and other variables. One significant four-way interaction involved Subtest, Amplitude Mode, Listening Mode, and Subject. Some of these interactions are depicted in Figures 31 through 34, which show percent correct scores obtained for each subtest. Results are shown for both quiet and noise, by subject, as a function of Amplitude Mode and Frequency Response. These values are also given in tabular form in Appendix J, Tables 36 through 39, respectively.

One of the findings that emerged was that in quiet, with flat frequency response, under the condition of non-

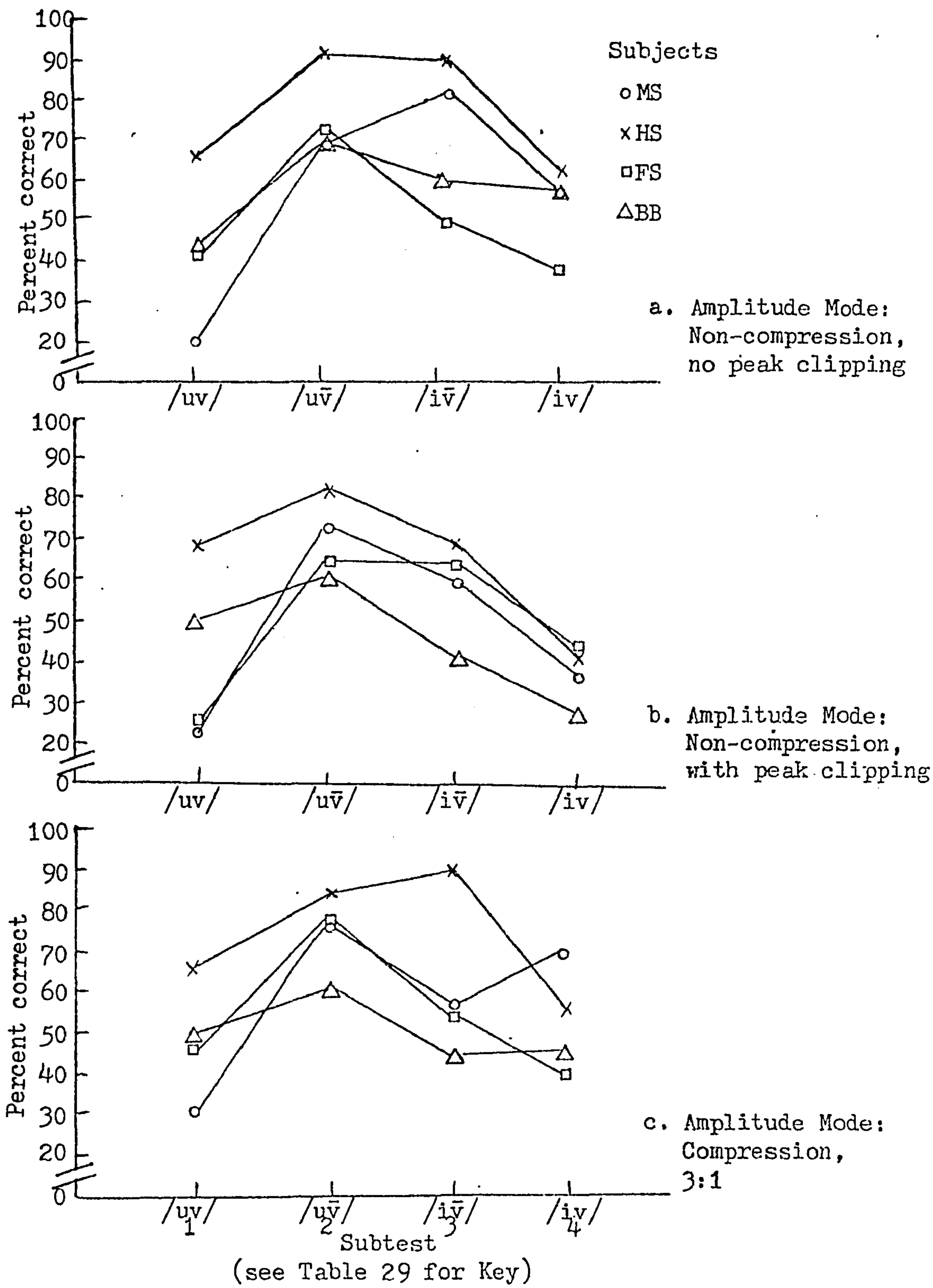



Fig. 31. Percent correct scores obtained in quiet with flat frequency response, as a function of Amplitude Mode, subject, and subtest. The values shown are also given in tabular form in Appendix J, Table 36.

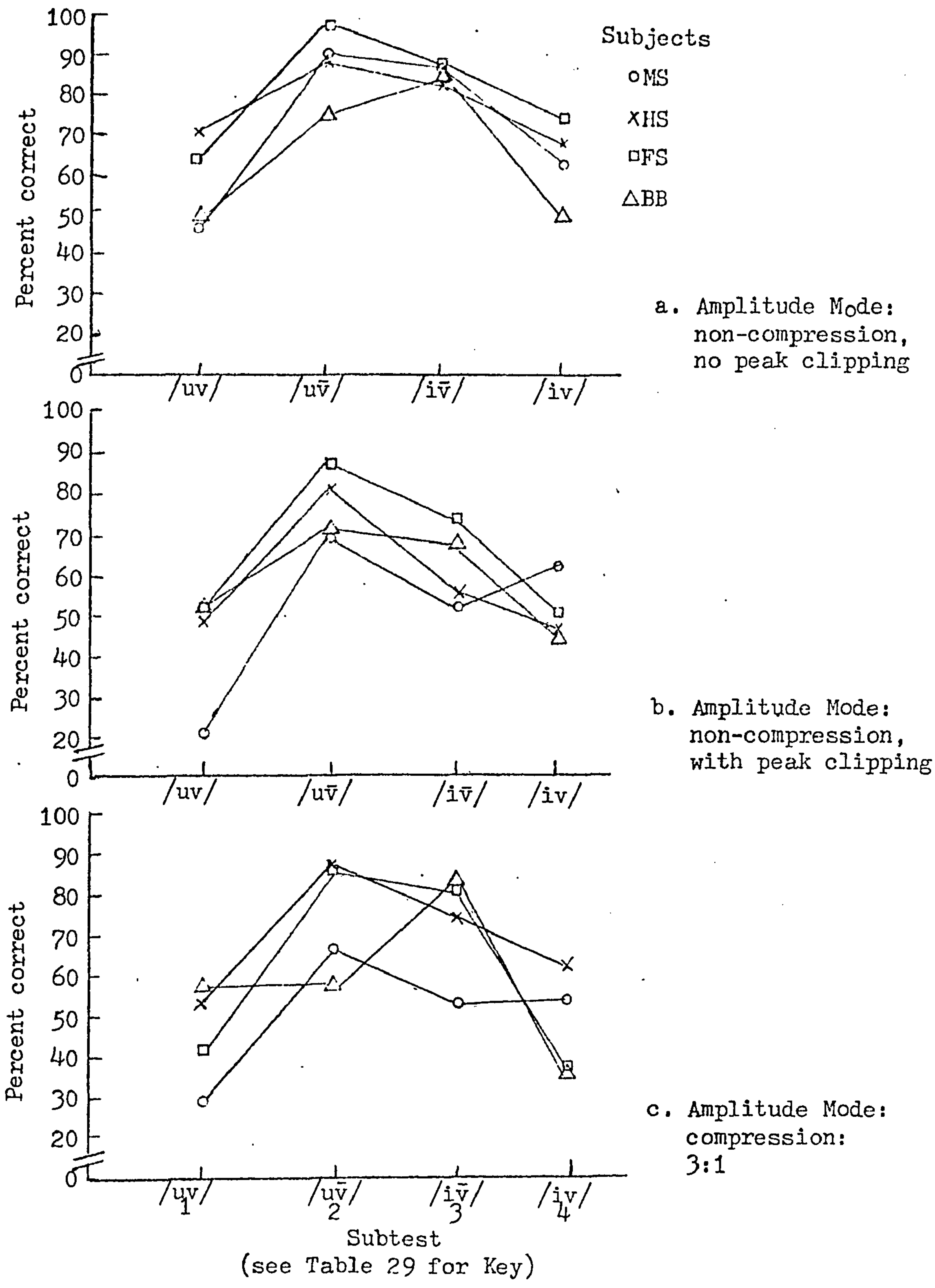
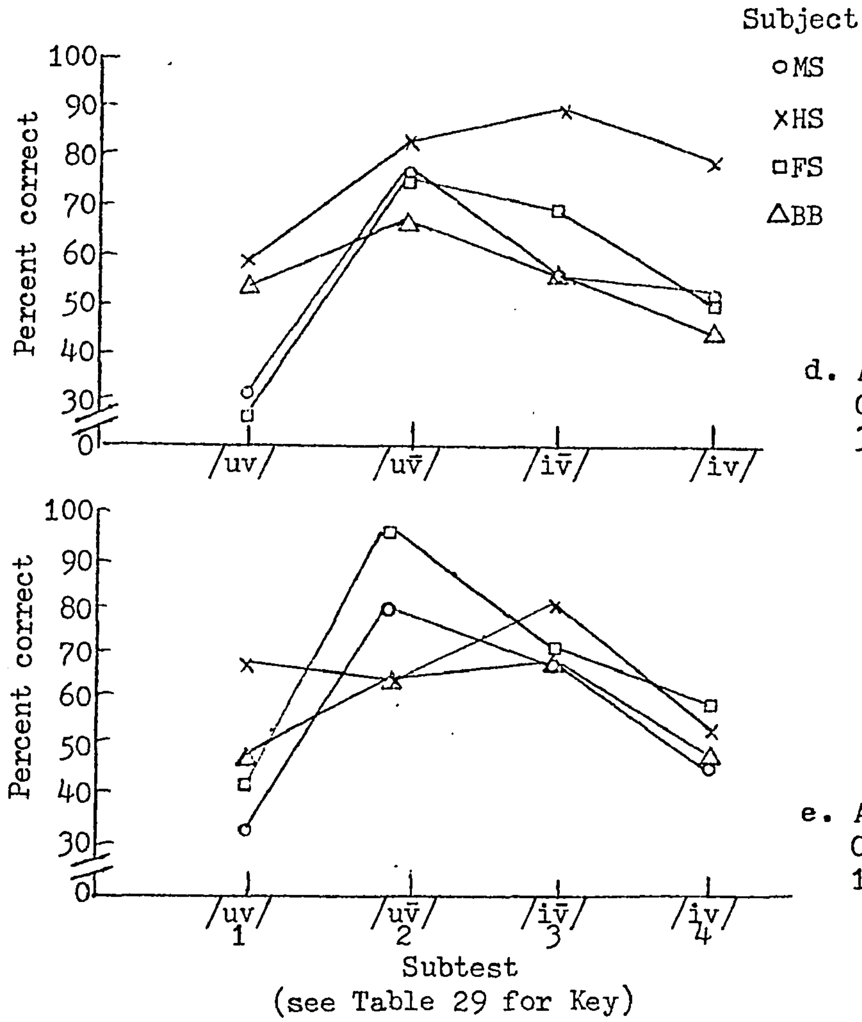



Fig. 32. Percent correct scores obtained in quiet with LDL frequency response as a function of Amplitude Mode, subject, and subtest. The values shown are also given in tabular form in Appendix J, Table 37.

Fig. 32--continued




d. Amplitude Mode: Compression, 3:1, 3:1

e. Amplitude Mode: Compression, 1.5:1, 3:1

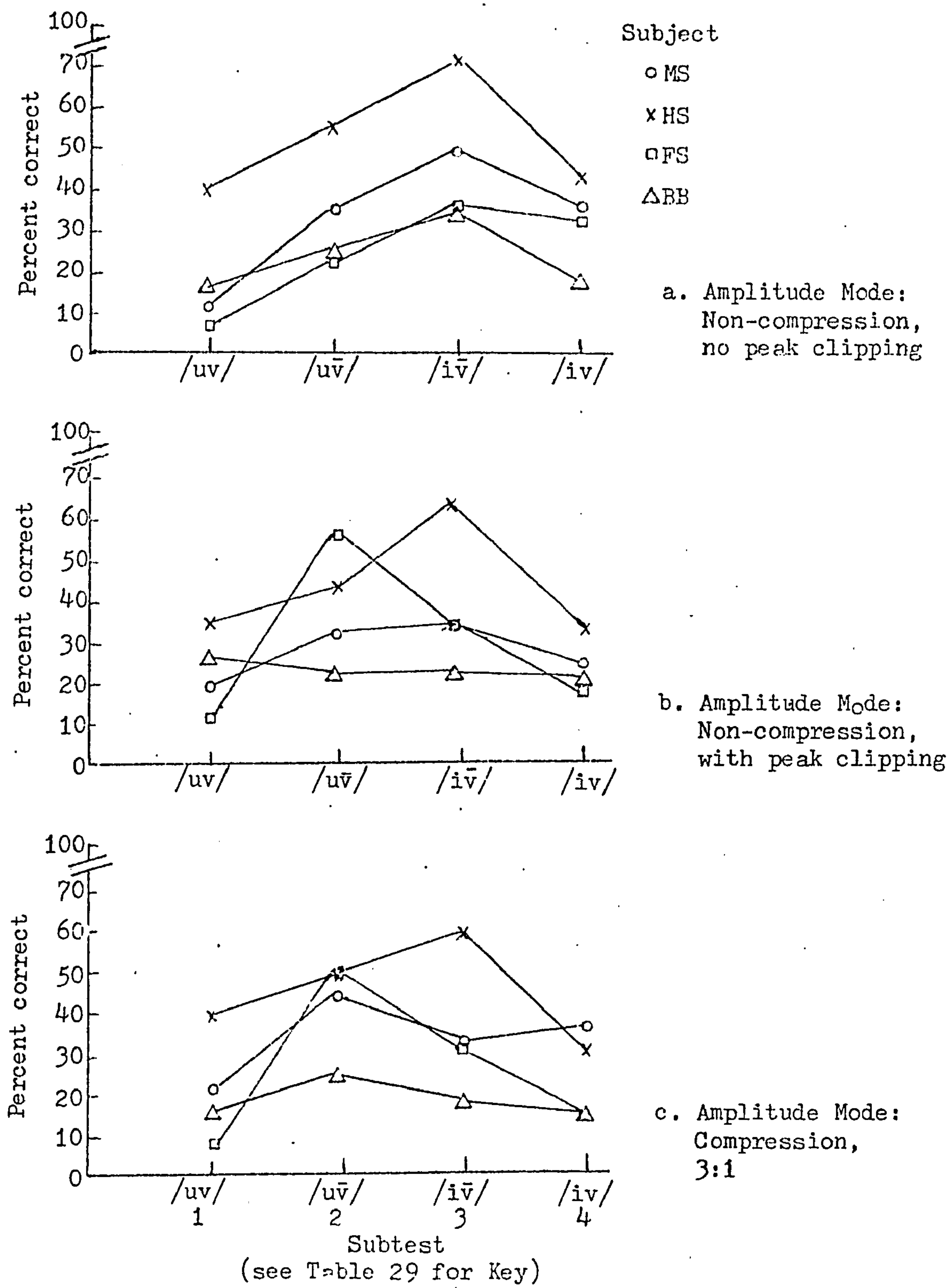



Fig. 33. Percent correct scores obtained in noise with flat frequency response as a function of Amplitude Mode, subject, and subtest. The values shown are also given in tabular form in Appendix J, Table 38.

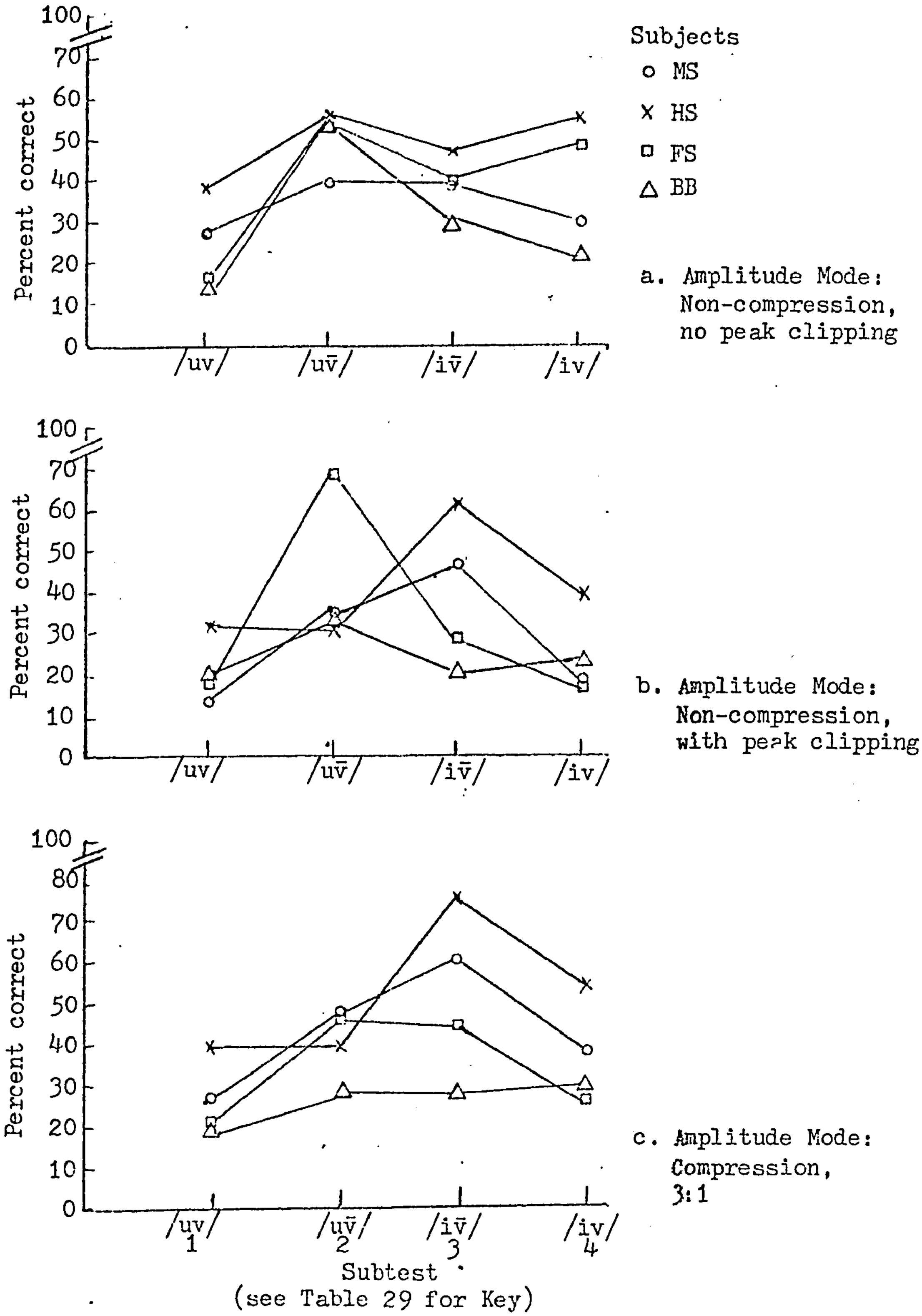
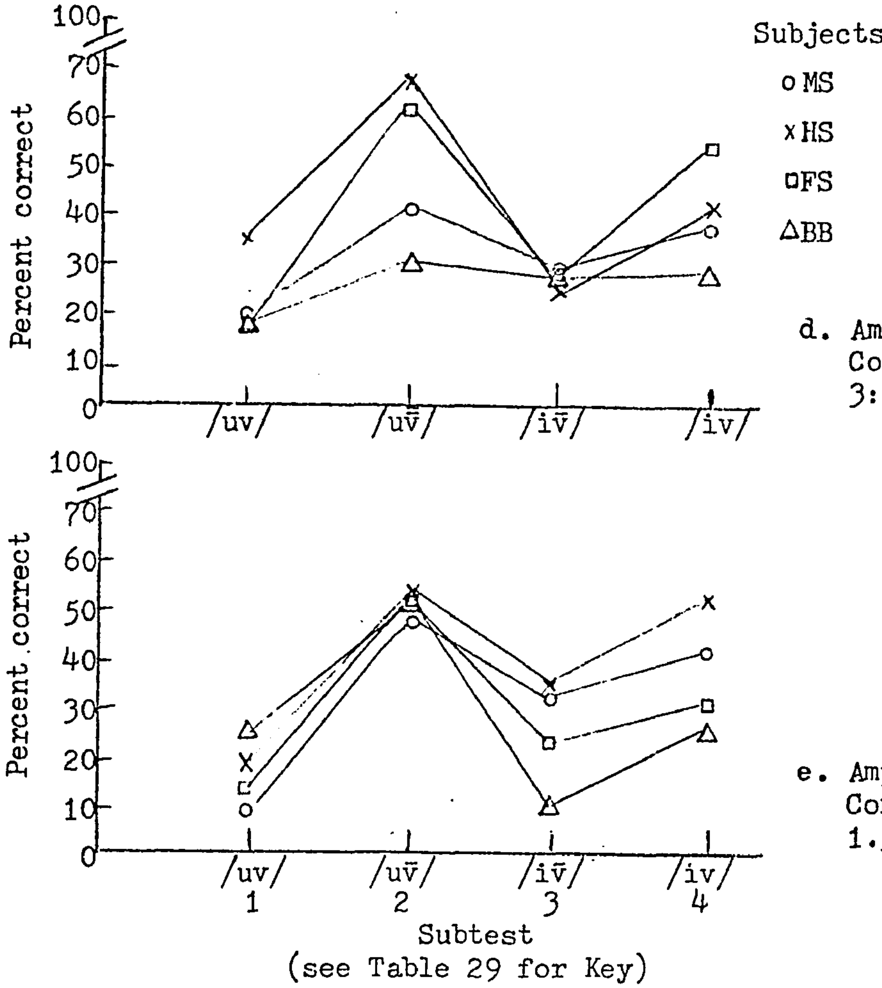



Fig. 34. Percent correct scores obtained in noise with LDL frequency response as a function of Amplitude Mode, subject, and subtest. The values shown are also given in tabular form in Appendix J, Table 39.

Fig. 34--continued




d. Amplitude Mode: Compressicn, 3:1, 3:1

e. Amplitude Mode: Compression, 1.5:1, 3:1

compression, no peak clipping, subjects scored best, on the average, on /u$\bar{v}$/ (voiceless consonant following /u/), and second best on /i$\bar{v}$/ (voiceless consonant following /i/). In noise, this pattern reversed itself, with all subjects achieving their best scores on /i$\bar{v}$/, with /u$\bar{v}$/ ranking second best.

The same pattern that applied to flat frequency response in quiet applied to LDL frequency response in quiet. In noise, with LDL frequency response, on the average, subjects achieved their best scores on /u$\bar{v}$/, as they did in quiet, with /i$\bar{v}$/ and /iv/ (voiced consonant following /i/) ranking second best for different subjects, respectively.

In quiet, the average difference in scores, for all Amplitude Modes, between /u$\bar{v}$/ and /i$\bar{v}$/ was 9%, with scores higher for the former subtest. In noise, however, there was no overall difference in scores between these two subtests. In quiet, the difference in scores between /uv/ (voiced consonant following /u/) and /iv/ was 5%, with scores better for /uv/. In noise, this difference, still in the same direction, doubled to 10%.

With two-channel compression there was a reversal in the pattern of subjects scoring better on voiceless consonants than on voiced. That is, in comparing scores achieved for /i$\bar{v}$/ and /iv/, all subjects scored better on the latter subtest, involving voiced consonants. This reversal did not apply to the subtests containing the vowel /u/.

In summary, there were significant diferences among subtests. On the average, subjects scored better on subtests containing voiceless consonants than on those containing voiced consonants, and better on subtests containing /u/ than on those containing /i/.

# CHAPTER VII

## DISCUSSION OF THE MAIN EXPERIMENT

The main experiment was designed to investigate the potential value of both single- and two-channel compression amplification in improving intelligibility for the hearing-impaired. Comparison of these modes of amplification was made to a non-compression hearing aid, both with and without peak clipping.

The experimental design consisted of the following: two non-compression systems, one single-channel compression system, and two dual-channel compression systems were evaluated for four subjects with sloping, moderate to moderately severe sensorineural hearing losses. All testing was performed using a modification of the Nonsense Syllable Test (NST) (Levitt and White, 1978), spoken by a female speaker in quiet and in noise, at a signal-to-noise ratio of 10 dB. The frequency responses used were a flat frequency response (0 dB/octave slope) and a frequency response shaped such that the amplified speech spectrum above 250 Hz lay uniformly below each subject's Loudness Discomfort Level (the "LDL frequency response curve"). Because of practical constraints on the subjects' time, the two-channel compression conditions employed only the latter frequency response.

## Implications of the Main Experiment

Several findings were revealed by the data obtained in the main experiment. First, compression amplification, whether single-channel or dual-channel, was not found to improve subjects' speech discrimination. Rather, non-compression amplification, without peak clipping, yielded the best discrimination score, on average, among all modes of amplification. The poorest performance was obtained, on average, for those conditions involving peak clipping. Second, the data showed the importance of individualized fitting of hearing aids over a fixed frequency-gain characteristic. Third, time-order effects were observed, possibly a combination of familiarity with the test material and auditory learning. And fourth, voiceless consonants yielded better discrimination scores, with the exception of two conditions, than did voiced consonants. A discussion of each of these findings follows.

### Amplitude Mode

In contrast to the present study, some recent studies (e.g. Villchur, 1973; Yanick, 1976) have shown that compression amplification improves discrimination over non-compression amplification. However, their non-compression conditions which were used as references may not have included the best frequency-gain characteristics for each of the listeners, resulting in an artificial increase of scores with compression. For example, Villchur's non-compression system did not provide high frequency emphasis on the grounds that high frequency emphasis without compression might result in distress from environmental sounds. However, other studies on non-compression amplification have shown that individualized frequency-selective amplification with high frequency emphasis substantially improves discrimination (e.g. Lippmann, 1978). Thus, Villchur's use of a non-compression system without high frequency emphasis as the reference condition may have resulted in artificially increased scores with compression.

Another example may be drawn from Yanick's studies. In an earlier study, Yanick (1973) compared an individually fitted single-channel compression hearing aid to each subject's own aid. His results demonstrated an advantage of the former over the latter in improving discrimination. In this case, the reference condition, i.e. the subject's own aid, may not have been optimally fitted to the subject. This point was reinforced in a later study when Yanick (1975) compared his own instrument used as a non-compression aid to its use as a compression aid. In this latter study, where the hearing aid fitting was optimized for both conditions, he demonstrated little or no advantage for compression.

Nonetheless, in the present study, the advantage of compression amplification may not have been realized for a number of possible reasons, which are discussed below.

Subject Differences

The individual differences found suggests that there may be subjects for whom compression would provide optimum performance, although none of the subjects in this study showed their best performance with compression. The slope and severity of the hearing loss may play a role in determining which subjects are aided best with a compression system. The dynamic range of Speech Reception Threshold (SRT) to Loudness Discomfort Level (LDL) may also have an effect on which subjects are aided best. However, with reference to the last variable, even HS, who had the narrowest dynamic range of the four subjects (20 dB) did not show any substantial improvement with compression over any of the other subjects. On the other hand, Villchur (1973) tested subjects with narrower dynamic ranges than those used in the present study, and found compression amplification to be superior. As pointed out by Villchur (1979), it may be that the combination of the use of moderately impaired subjects and a single, high presentation level made it possible to place the entire dynamic range of the non-compressed test material into the dynamic range of the hearing of the subjects.

Parameters of Compression Amplification

A second possibility as to why the advantage of compression amplification may not have been realized relates to the parameters of compression amplification. The data obtained in the study showed a large number of statistically significant two-, three-, and even four-way interactions involving Amplitude Mode. This raises two issues: First, since different parameters of compression resulted in different relative performance among subjects, it is conceivable that the optimum compression conditions for a given subject were not covered in the study. It is also possible that this optimum set of conditions is likely to differ among subjects. Second, these interactions raise difficulties in comparing results of this study with other studies. Although the effect of an individual parameter can be studied, caution must be exercised in generalizing to other conditions, or to other studies, in that an interaction effect may have taken place (Nabelek, 1973). Thus, combining mode of amplification with frequency responses other than those considered in the present study might have yielded different optimum conditions. Or, the use of different compression ratios, individualized compression ratios, different threshold of compression, or different number of channels from that utilized in the present study, might have yielded better results for compression amplification. However, with reference to the last variable, Lippmann (1978), despite using 16 channels in his compression system, obtained similar results to those of the present study, which utilized only two channels. In addition, it is reasonable to expect the most noticeable improvement of multi-channel compression over single-channel, if there is one, to show up in the difference between use of one channel and of two channels. This did not occur in the present study.

## Level of Presentation of Test Materials

Use of different stimulus levels might also have shown greater advantage for compression amplification. In the present study, the stimulus level at the subject's ear was estimated to be LDL minus 5 decibels. The results obtained were similar to those of Lippmann (1978), who presented his test materials at Most Comfortable Level (MCL) for most conditions. However, Lippmann also found that when reduced stimulus levels were used, his results were similar to those of Villchur (1973) (who also used a range of levels), i.e. compression amplification showed an advantage over non-compression amplification for low stimulus levels.

## Restriction of Dynamic Range of Test Materials

A characteristic of conventional test materials, such as that used in this study and in other studies (e.g. Caraway and Carhart, 1967), is that the range of variation of the levels is relatively small. The intensity used for the NST was typical of a speaker talking in a steady voice from a fixed distance, with a dynamic range of approximately 25 dB. However, in real life, the dynamic range is much larger, ranging from very low level stimuli, e.g. a voice at a distance; to fairly intense speech, e.g. somebody close by. This is in addition to the normal speech variations occurring in conversation. Thus, the advantage of compression amplification might not have been realized because of the restricted dynamic range of the test materials.

Use of the NST as the Test Material

Related to the above point is the possibility that the optimum Amplitude Mode found for the test material used, the NST, may differ from the optimum Amplitude Mode for another type of test material. In a test situation, the material is fixed not only in terms of intensity, but also in terms of content. In real life, however, there is a great deal of variability in the speech signal reaching the listener. Hearing aid wearers find themselves in many different communication situations. It is therefore possible that the optimum Amplitude Mode for those conditions of hearing aid use tested in the present study may not be what is optimum for other communication situations.

Summary

In summary, the results obtained regarding Amplitude Mode imply that for the type of hearing-impaired listeners used in the present study, and with conventional test materials, compression amplification is not superior to non-compression amplification with individualized frequency shaping. However, it was effective in keeping intensity within a desired range. Further, the alternative approach to limiting intensity, that of peak clipping, was found to cause deterioration of discrimination for the hard of hearing. Compression amplification would thus be the desired form of intensity limiting when such is necessary, as with listeners exhibiting a reduced dynamic range.

Frequency Response

A second major finding involved the relative importance of frequency-gain characteristics. As detailed in the Results section of the pilot study and the main experiment, Frequency Response interacted significantly with Subject, and also with Listening Mode. These findings indicated marked individual differences in the effect of frequency response. The interaction with subject assumed great importance for the main study, since it supported the notion that proper individulized fitting is a prerequisite of assessing the relative merits of both compression and non-compression amplification.

The procedure used in determining the LDL frequency response curve involved use of a calibrated noise tape, as discussed on pages 164 through 168. This tape was specially prepared to provide one-third octave bands of noise at equal levels, when played through the experimental equipment. The LDLs for one-third octave bands of noise were measured for each subject, using the calibrated noise tape. The frequency-gain function derived for each subject was referred to as the LDL frequency response curve, and was used in shaping the speech spectrum for several of the experimental conditions.

Two basic questions emerged from the use of this procedure. First, was the LDL frequency response approach, that is, placing as much of the speech spectrum as possible into the residual auditory area, the best approach? Second, was the specific method used in the present study accurate? A discussion of these two questions follows.

Theoretical Justification of Use of LDL Frequency Response

In response to the first question, the fundamental philosophy of Watson and Knudsen (1940), who were among the first proponents of selective amplification, was to "...fit the frequency-intensity areas occupied by the amplified speech sounds into the diminished and distorted auditory sensation area." Recent research has shown that placing as much of the speech spectrum as possible into the residual auditory area, as advocated by Watson and Knudsen, has led to good speech perception. For example, Barfod (1972) and Lippmann (1978) restored the loudness of the peak 10% levels of speech to their hearing-impaired listeners, resulting in improved discrimination. Their method was based on a similar approach to the LDL approach of the present study. Both their frequency-gain characteristics, and that obtained by Watson and Knudsen from their most comfortable equal loudness contour, may have had similar shapes to the LDL curve used in the present study.

Individualized frequency shaping with the LDL frequency response curve, correcting for the speech spectrum, allowed restoration of as much of the speech spectrum as possible for the hearing-impaired listeners (subject to possible limitations, mentioned below). With other procedures of frequency shaping, as soon as one frequency component reaches the LDL for a particular frequency, the overall intensity to the subject can no longer be raised. This is because a single peak of level at any particular frequency may use up a large part of the limited dynamic range available to a patient. Amplification can then be maximized for only some frequencies, with the amount of amplification at other frequencies being, of necessity, less than maximum.

In summary, an important issue in hearing aid research is whether or not individualized selective amplification provides significant improvements in performance over a non-individualized frequency-gain shaping. The interactions obtained in the main experiment between subjects and frequency response, which were highly significant, support the view that individualized frequency shaping is advantageous. Further, the LDL method, which requires individualized frequency shaping, showed the highest scores for all conditions for three of the four subjects. The fourth subject (HS), who, on average, did not show the LDL response to be the best, nonetheless, showed significant learning effects for the LDL frequency response, with scores being low at the beginning and high at the end. An analysis of the LDL frequency response curve for the four subjects showed an overall high frequency emphasis of between 5 and 10 dB, across subjects, for the frequency range of 1000 to 4000 Hz (referred to measurements in a 6cc coupler). This is a steeper high frequency emphasis than that recommended by the Harvard study (Davis, et al., 1947). Nonetheless, it is not known how much better the LDL procedure is compared to a non-individualized frequency response with high frequency emphasis.

Method Used for LDL Frequency Response

The approach using the LDL frequency response curve appeared to have advantages in hearing aid selection, as discussed above. Issues arose, however, regarding the accuracy of the method used in obtaining the LDL frequency response. Among the issues were the following:

The shaping of the LDL frequency response curve was based on the use of LDLs derived for one-third octave bands of noise. These bands were chosen for practical convenience. It is more likely, however, that the ear sums loudness in terms of critical bands. It would, therefore, seem to be more natural to have worked with the critical band, but it is not known exactly what the critical bands are for Loudness Discomfort Levels. Nonetheless, if there is any deviation between the one-third octave bands and the critical bands, it is likely to be consistent and systematic.

Another relevant issue was whether the LDLs obtained for one-third octave bands represented the LDLs for the same third octave bands for speech. In practice, the LDL for speech for the same one-third octave band may have differed, since the temporal pattern of noise does not fluctuate as much as that of speech. Nonetheless, the LDLs obtained for each one-third octave band of noise represented, to a first approximation, the LDLs for a broad band stimulus, namely speech, by one-third octave bands.

Another question that arose from the method used was, if every band was at LDL, how far was the sum of the bands above LDL, if at all? That is, when the contiguous bands were presented simultaneously, each one at LDL, was the resultant complex pattern at or above LDL? An informal subjective trial on a normal-hearing listener (Levitt, 1979) has shown the total Loudness Discomfort Level of the summed bands to exceed the LDLs of the individual bands by several (about 5) decibels. In any case, the procedure used involved obtaining the LDL frequency response curve, and then adjusting the total gain of the summed LDLs, as discussed further below.

One part of the methodology involved an adjustment for the speech spectrum (see page 170). The values of the spectrum that were used in the adjustment were obtained from the data of Dunn and White (1940). Those band levels which were exceeded 10% of the time by the speech peaks were used in deriving the spectrum. Peak speech values, rather than the RMS values, were used for this adjustment. This was based on the reasoning that a person's judgment of his tolerance for discomfort is based on the peaks of speech, rather than on the long-term spectrum of speech. The 10% level was chosen arbitrarily for practical reasons, although the speech peaks perceived by the auditory system may correspond to a different percentage level. However, this is not a crucial point, since Dunn and White's curves were essentially parallel for different percentage levels.

One last point related to the methodology used involved the lower cutoff frequency. Because of the possibility of upward spread of masking, that is, excessive low frequency amplification interfering with reception of the high frequencies, a low frequency rolloff at about 250 Hz was empirically applied. This correction was based on data obtained in the Harvard study (Davis, et al., 1947).

Level of Presentation of Test Materials

A difficult question to answer in any evaluation of speech discrimination with various hearing aids is, with any given frequency response, at what level will the individual's speech discrimination be at its maximum? As is well known, stimulus intensity has a marked effect on intelligibility. In this experiment, wide-band LDL minus 5 dB was used, since this was consistent with the argument of maximizing the amount of speech in the residual hearing area. Traditionally, the maximum intelligiblity has been at a predetermined level of approximately 35 decibels above the Speech Reception Threshold. However, some researchers have found that the range of sensation levels at which maximum discrimination is obtained, e.g. PB-max, varies from one subject to the next. Thus, it has been hypothesized that if there is any single sensation level at which maximum discrimination can be obtained, it might be the Most Comfortable Level. However, some studies which have investigated the relationship between the sensation level selected as most comfortable for loudness and intelligibility, and the sensation level at which the maximum speech discrimination is obtained, have failed to establish MCL as the level of maximum discrimination (Posner and Ventry, 1977). These results indicated that subjects did not choose comfort levels at which they could discriminate the best. In light of these findings, i.e. that subjects had significantly poorer discrimination at their comfort levels, even when subjective intelligibility was used as the criterion for selecting that level, it was decided to conduct experimental conditions at an intensity higher than MCL. In addition to the findings mentioned, Dubno and Hochberg (1977) found that when an adaptive procedure was used, hearing-impaired listeners with sensorineural hearing losses scored highest on speech discrimination testing when the intensity of test administration was close to LDL.

It is, of course, recognized that when a person wears a hearing aid, he is likely to set the aid's volume at a level that he considers to be the most comfortable for his understanding of speech. However, many hearing aid wearers report that MCL is not fixed at any one level, but covers a range. Consequently, they readjust the hearing aid gain to meet various communication needs. This often leads to the listener using levels higher than the level determined clinically as being the one sensation level representing MCL. Since there is a range of possible MCL values, test-retest variability may be high, and might result in increased between-session variability. In the present study, although LDL did show variability for some subjects, it never amounted to more than 5 dB between sessions. This gave an additional advantage to the use of LDL, less a constant, as the level of test administration.

## Summary

The magnitude of difference between flat frequency response and LDL frequency response, although statistically significant, averaged only 4% in quiet and 6% in noise, with the latter frequency response receiving the higher scores. But for the condition of non-compression, with peak clipping, the amount of improvement with the LDL frequency response curve was 12%. This condition was the most difficult one for the subjects, yielding the poorest overall score. The implication was that as listening conditions became more difficult, the advantage of an individualized strategy for hearing aid fitting became increasingly apparent.

In summary, the practical significance of an individualized strategy to hearing aid fitting is two-fold. First, use of a fixed frequency-gain characteristic did not appear to yield the best discrimination score for hearing-impaired listeners, as originally proposed by the Harvard study (Davis, et al., 1947). Second, use of an individualized strategy for hearing aid selection, such as the LDL frequency response curve, did significantly improve discrimination. Although the amount of improvement may not appear large under conditions of quiet, as listening conditions became more difficult, the amount of improvement became greater. The implication is that some form of individualized frequency-gain shaping is desirable. It should be noted that this improvement was obtained in comparison with a flat frequency response (as measured in a 6cc coupler). It is not known how much better the LDL procedure is compared to a non-individualized frequency response with high frequency emphasis. Whereas the LDL frequency-gain characteristic may not necessarily be the optimum choice for all subjects under all conditions, it proved to be an advantageous form of frequency shaping, and can be obtained relatively easily in practice.

Time-Order Effect

A third finding in the main experiment involved a time-order effect, which was found to be significant under certain experimental conditions. This effect was not apparent for flat frequency response, but was apparent for LDL frequency response. With reference to the latter condition, the effect existed in quiet (8% increment in score from first replication to last replication), and to an even greater extent in noise (13% increment in score from first replication to last replication). Not only did the magnitude of this effect differ for quiet and noise, but the pattern of the effect also differed. In quiet, the entire effect was apparent between the first and second replication (8% increment). In noise, there was an equally large increment in score between the first two replications as there was in quiet. Then, after a plateau, there was an additional increment between the third and fourth replications (an additional 5%).

Examination of the data showed that there was some evidence of familiarization with the test procedure. The average improvement was greatest for the first few trials. There was also some evidence of an improvement in discrimination towards the end of the test, an effect that could be attributed to auditory learning. These effects, taken individually, did not reach statistical significance. In combination, however, the time-order effect was significant.

The implication of this finding is that auditory training, in combination with hearing aid selection, may result in improved performance with a hearing aid. The time-order effect found was greater for that frequency response with which the subject was not familiar with listening, i.e. LDL frequency response. And, it was even greater when this listening situation was made more difficult by the addition of noise. Although the present study was not designed for the purpose of confirming this, the practical significance of these results suggests that hearing-impaired listeners might benefit from auditory training, especially in difficult listening situations. This may be especially true for compression amplification, which may restore cues to the listener that he has not heard for a long time.

Pattern of Phoneme Errors

A fourth finding in the main experiment involved the pattern of phoneme errors. Analysis by subtest revealed that identification was superior for unvoiced consonants than for voiced consonants, both in quiet and in noise. This raised the question of whether the voiceless consonants were more or less detectable than the voiced. On the average, the relative levels of voiced sounds produced in various contexts is greater than for voiceless sounds (Fletcher, 1953). This is, however, a complex issue, since the voiceless consonants appear to be easier to detect in certain contexts. The finding of the main experiment was consistent with data obtained in the Wearable Master Hearing Aid (WMHA) study (Levitt and White, 1978) for syllables involving use of the same vowels as in the present study, i.e. /i/ and /u/. (With the /a/ vowel, the pattern in the WMHA study was reversed, i.e. identification for voiced consonants was superior to that of unvoiced.) These findings appear to be characteristic of this particular recording of the Nonsense Syllable Test.

The data obtained revealed the above finding concerning /i/ and /u/ to be consistent in all test situations, except for the two dual-channel compression conditions, with LDL frequency response, in noise. Under these conditions, the pattern reversed with voiced consonants superior to unvoiced consonants, but only when following the vowel /i/. This result may have been attributed to the effect of the adjacent vowel. As illustrated in Levitt (1978a), the spectrograms of some consonants may be modified by the vowels that precede or follow them. The first formant ($F_1$) of /u/ and /i/ are 300 Hz and 370 Hz, respectively, for female speakers. $F_1$ would therefore not have been differentially affected in the two-channel conditions, since the energy of $F_1$ for both vowels passed through the low frequency band, i.e. 250 Hz to 1500 Hz. The second formant ($F_2$) is 950 Hz for /u/, and therefore still passed through the low frequency band. However, $F_2$ is 2800 Hz for /i/, and passed through the high frequency band, i.e. 1500 Hz to 6000 Hz. Because of a possible difference in the effect on $F_2$ for /u/ and /i/ for the two-channel compression conditions, the effect on the following consonants appears to have differed. The result was that the voiced consonants were superior to the unvoiced when following /i/, but not when following /u/. It is not apparent, however, why this differential effect was found for only one specific set of conditions of two-channel compression amplification.

## Conclusions

In conclusion, the present study has shown that an individually fitted non-compression system offers an advantage over a one- or two-channel compression system. That is, with frequency shaping individually adjusted to the subjects, better performance was achieved with non-compression amplification than with compression. This was found to be true for subjects with mildly downward sloping, moderate to moderately severe sensorineural hearing losses, tested with a modification of the Nonsense Syllable Test. These results might, however, have differed under other conditions, with different test materials of greater dynamic range, or for subjects with different slope or severity of hearing loss. The results do not imply that compression amplification is not advantageous over peak clipping for intensity limiting. In addition, because speech has such wide variations in level (a condition that was not included in the present study), it is possible that, on an average long-term basis, compression amplification may actually be better for the listener than non-compression amplification.

# APPENDICES

APPENDIX A

Measured Input-Output Functions for Various Compression Ratios for the Compression Amplifier

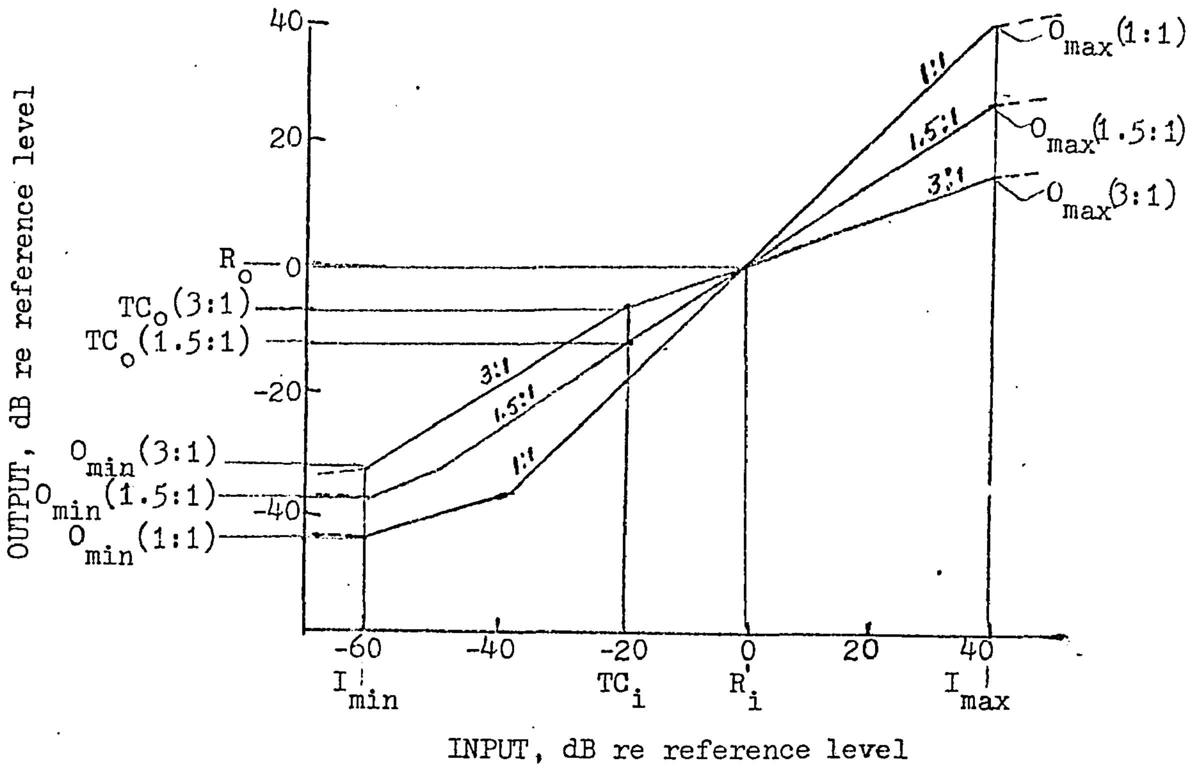



Fig.35. Measured input-output functions for various compression ratios of the compression amplifier. Numbers on functions identify the respective compression ratios. Shown is the change in the dynamic range of the compressor with a change in the compression threshold to -20 dB re reference level. Explanations of abbreviations used are found in the key, on the following page.

Appendix A, Fig. 35--continued

KEY

| | |
|---|---|
| $I_{min}$ | Minimum input level detectable above internal noise. |
| $O_{min}$ | Minimum output level detectable above internal noise. Varies as a function of compression ratio. The numbers in parentheses identify the compression ratio. |
| $TC_i$ | Threshold of compression, at input. |
| $TC_o$ | Threshold of compression, at output. Varies as a function of compression ratio. The numbers in parentheses identify the compression ratio. |
| $R_i$ | Reference level, at input. Equivalent to 100 mv. |
| $R_o$ | Reference level, at output. Equivalent to 100 mv. |
| $I_{max}$ | Maximum input level prior to overload of input circuit. For practical purposes, there is no further increase in output with increase in input above this level. |
| $O_{max}$ | Maximum output level prior to overload of input circuit. Varies as a function of compression ratio. The numbers in parentheses identify the compression ratio. |

Appendix A--continued

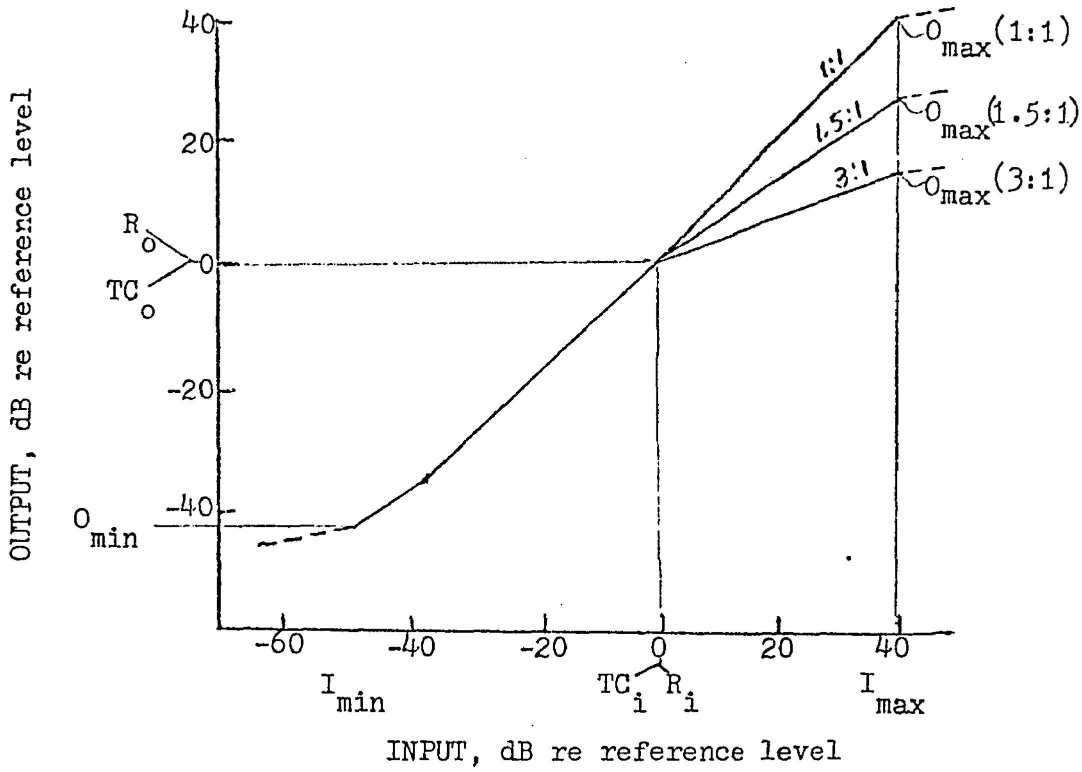



Fig.36. Measured input-output functions for various compression ratios for the compression amplifier. Numbers on functions identify the respective compression ratios. Shown is the change in the dynamic range of the compressor with a change in the compression threshold to 0 dB re reference level. Explanation of abbreviations found in the figure are in the key, on the following page.

Appendix A, Fig. 36--continued

KEY

| | |
|---|---|
| $I_{min}$ | Minimum input level detectable above internal noise. |
| $O_{min}$ | Minimum output level detectable above internal noise. |
| $TC_i$ | Threshold of compression, at input. |
| $TC_o$ | Threshold of compression, at output. |
| $R_i$ | Reference level, at input. Equivalent to 100 mv. |
| $R_o$ | Reference level, at output. Equivalent to 100 mv. |
| $I_{max}$ | Maximum input level prior to overload of input circuit. For practical purposes, there is no further increase in output with increase in input above this level. |
| $O_{max}$ | Maximum output level prior to overload of input circuit. Varies as a function of compression ratio. The numbers in parentheses identify the compression ratio. |

# APPENDIX B

## Frequency Responses of Compression Amplifier for Various Compression Ratios .

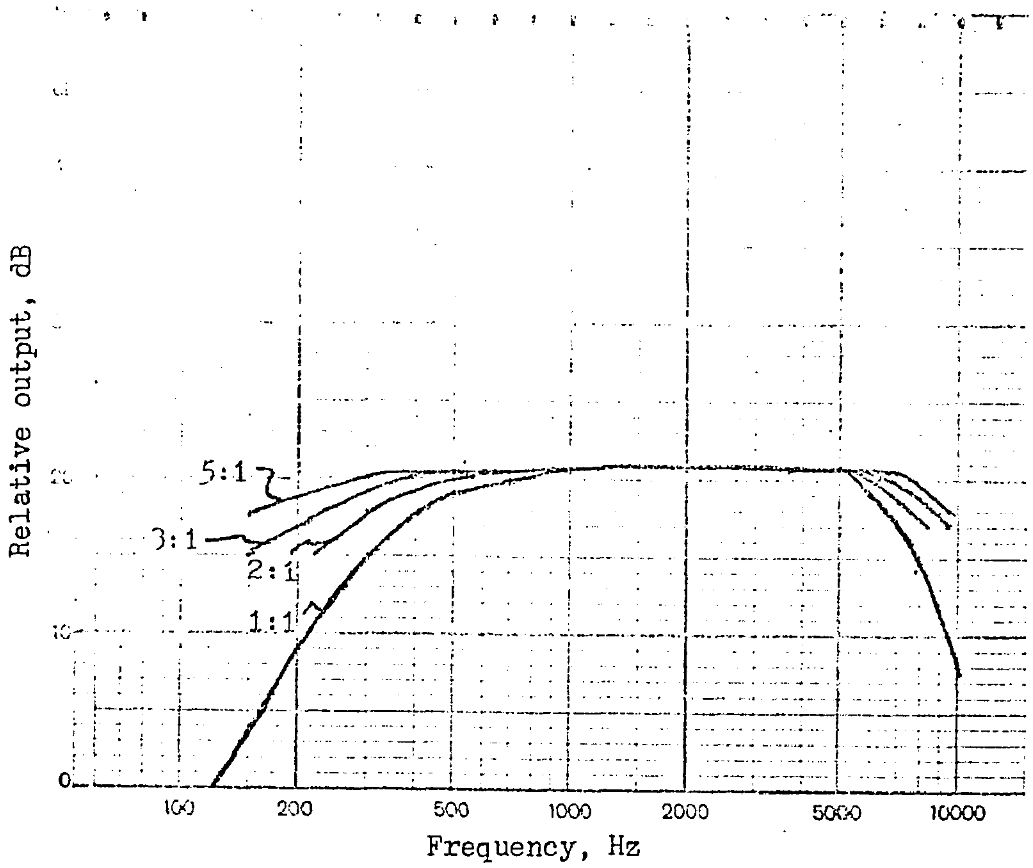



Fig. 37. Frequency responses of compression amplifier for various compression ratios. Input intensity was at the reference level, 100 mv. Frequency range was 250 Hz through 6000 Hz. Number on each curve identifies the respective compression ratio. Since the input is at reference level, there is no change in output intensity, despite changes in compression ratio.

Appendix B--continued

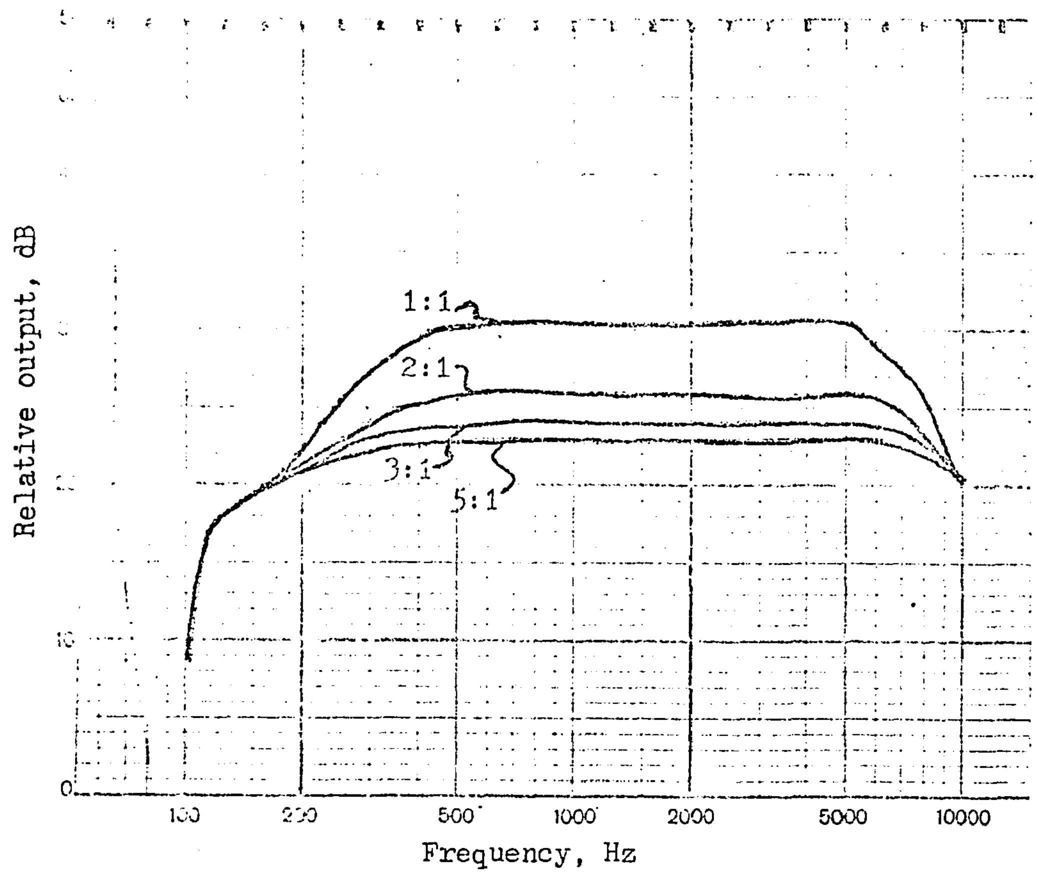



Fig. 38. Frequency responses of compression amplifier for various compression ratios. Input intensity was +10 dB re the reference level. Frequency range was 250 Hz through 6000 Hz. The number on each curve identifies the respective compression ratio. Note that the greater the compression ratio, the less is the relative amplification.

Appendix B--continued

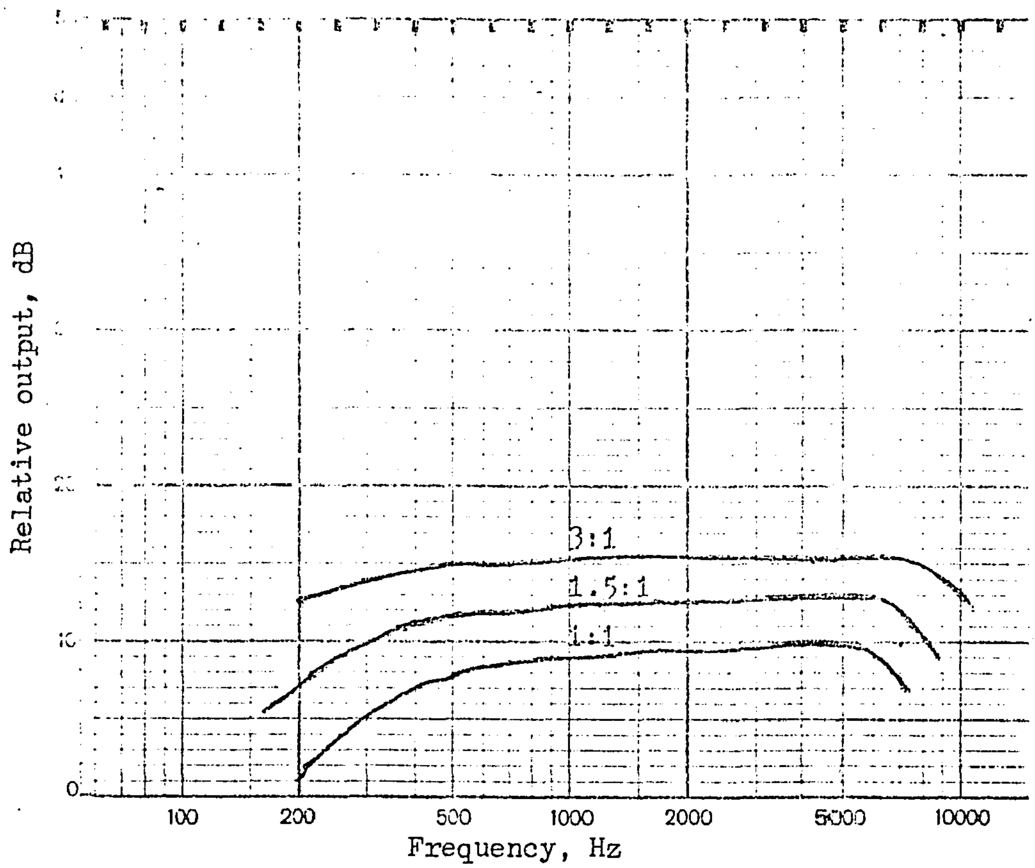



Fig. 39. Frequency responses of compression amplifier for various compression ratios. Input intensity was -10 dB re the reference level of 100 mv. Frequency range was 250 Hz through 6000 Hz. The number on each curve identifies the respective compression ratio. Note that the greater the compression ratio, the greater is the relative amplification.

# APPENDIX C

## Frequency Response of the Compression Amplifier with 6 dB/Octave Slope

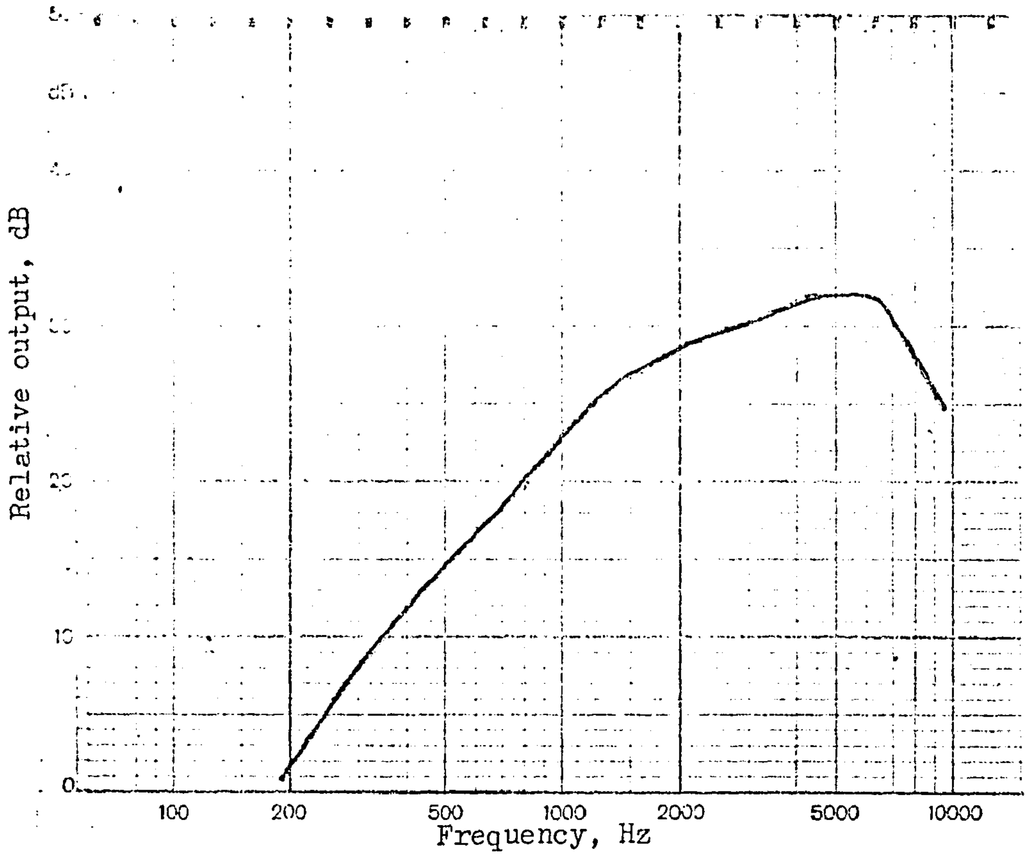



Fig. 40. Frequency response of the compression amplifier at a compression ratio of 1:1, and a slope of 6 dB/octave. Frequency range of channel 1 was 250 Hz to 1500 Hz, and of channel 2, 1500 Hz to 6000 Hz.

APPENDIX D

Real Time Spectral Analysis of White Noise with the Compression Amplifier

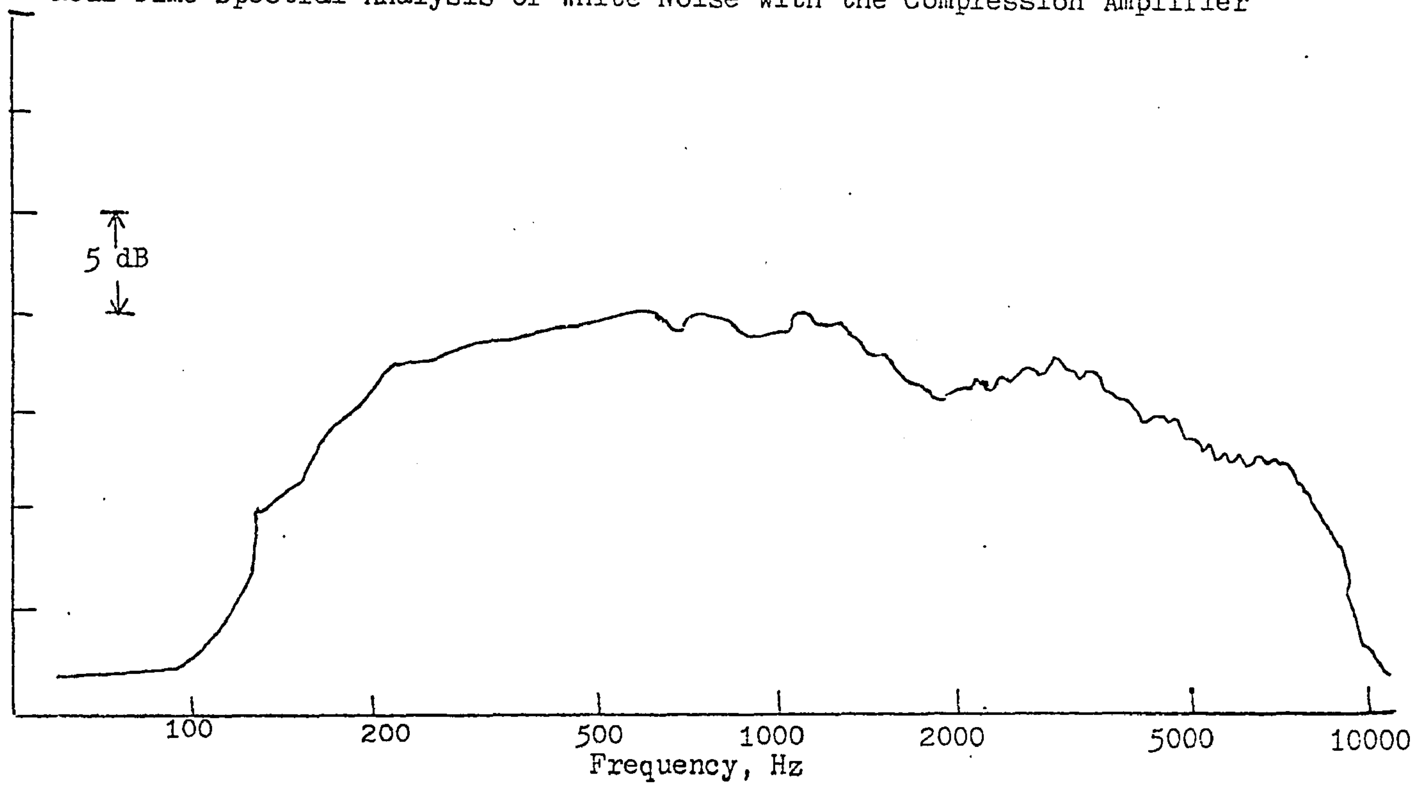



Fig. 41. Real time spectrum analysis of white noise at -10 dB re the reference level of 100 mv. The frequency response was 0 dB/octave, and compression ratio was 1:1. Frequency range was 125 Hz through 6000 Hz.

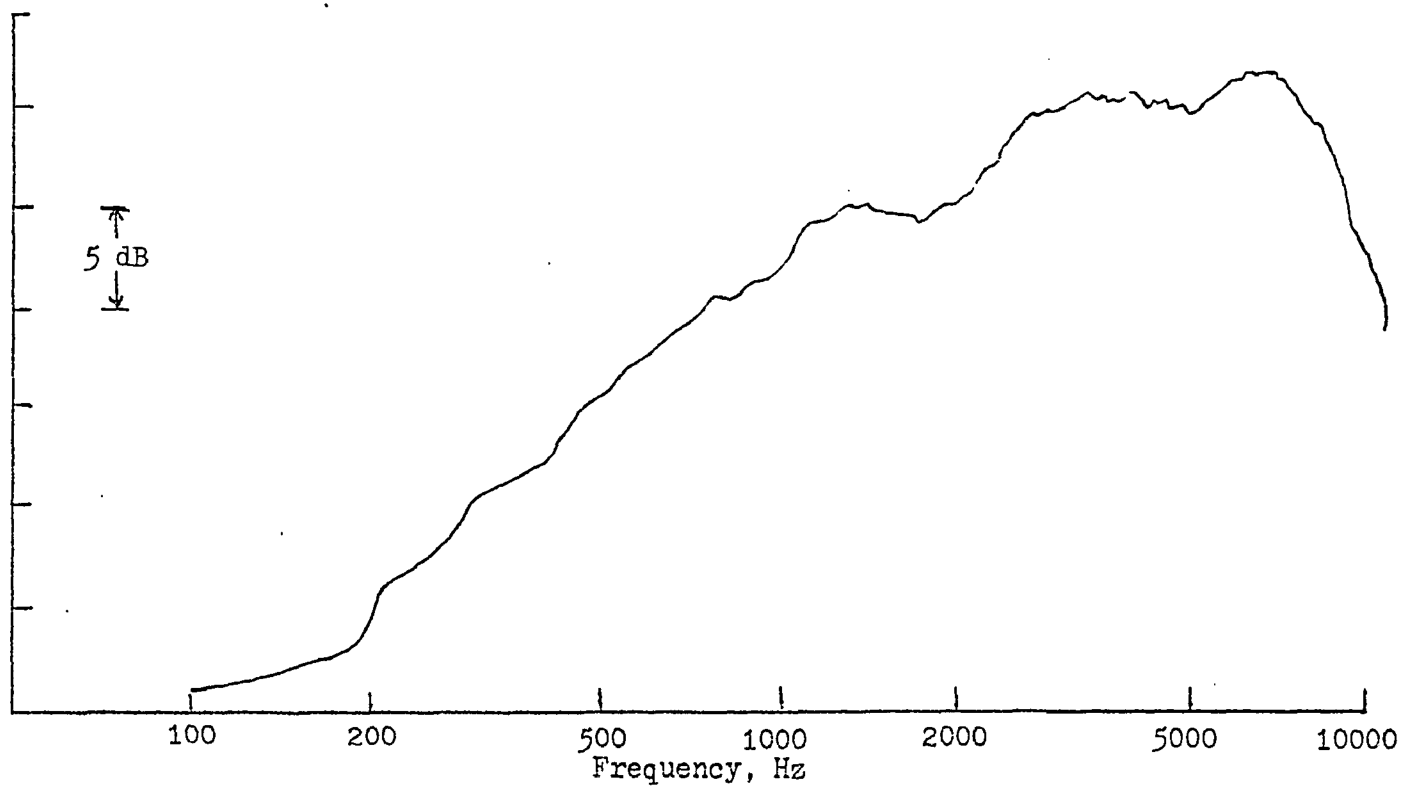



Fig. 42. Real time spectrum analysis of white noise at -10 dB re the reference level of 100 mv. The frequency response was 6 dB/octave, and compression ratio was 1:1. Frequency range was 125 Hz through 6000 Hz.

# APPENDIX E

## Measured Loudness Discomfort Levels for Subjects of the Pilot Study

Loudness Discomfort Levels (LDLs) and relative LDLs for speech, by subject, measured under six conditions (2 frequency responses x 3 bandwidths). The first part of the table shows the values for broad band condition ($L_b$), with values given in dB HTL. The second part of the table shows the values for each of two channels ($L_1$ and $L_2$, respectively), with values given relative to channel 1 ($L_1$).

| | FREQUENCY RESPONSE | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|
| | 0 dB/octave | | | | | 6 dB/octave | | | | |
| | Subject | | | | | Subject | | | | |
| BANDWIDTH | RG | RS | KH | LS | RM | RG | RS | KH | LS | RM |
| Broad band ($L_b$), 125 - 6000 Hz, dB HTL | 95 | 95 | 105 | 110 | 100 | 95 | 100 | 100 | 110 | 100 |
| Channel 1 ($L_1$), 125 - 1000 Hz | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 0 |
| Channel 2 ($L_2$), 1000 - 6000 Hz, dB re $L_1$ | 5 | 10 | 10 | 15 | 15 | -10 | 0 | -10 | 0 | 10 |

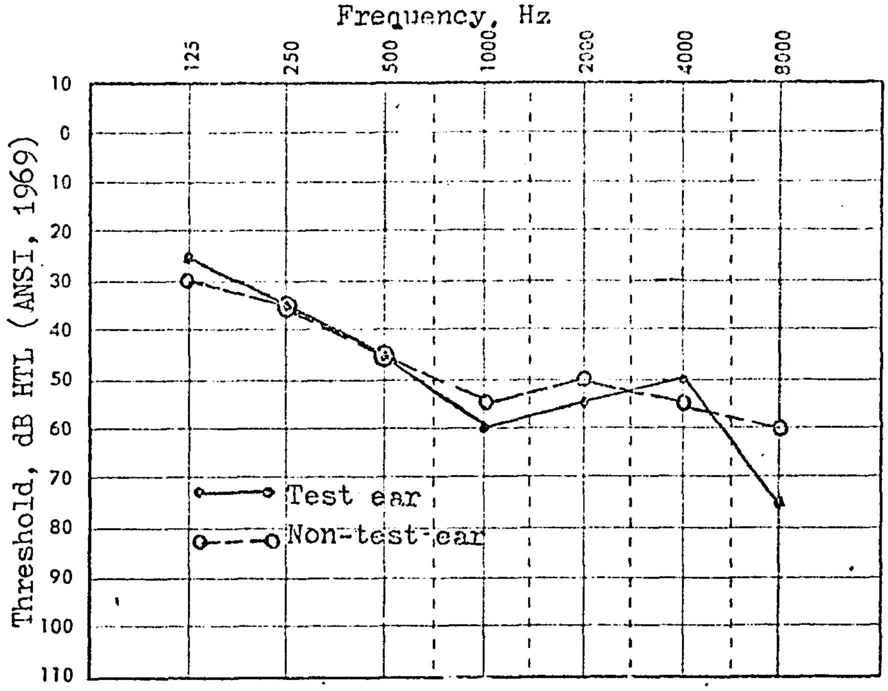
APPENDIX F

Response Form for the Nonsense Syllable Test, Subtest One

| | | | | | | |
|---|---|---|---|---|---|---|
| OF | OP | OSH | OT | OTH | OK | OS |
| OF | OP | OSH | OT | OTH | OK | OS |
| OF | OP | OSH | OT | OTH | OK | OS |
| OF | OP | OSH | OT | OTH | OK | OS |
| OF | OP | OSH | OT | OTH | OK | OS |
| OF | OP | OSH | OT | OTH | OK | OS |
| OF | OP | OSH | OT | OTH | OK | OS |
| OF | OP | OSH | OT | OTH | OK | OS |

# APPENDIX G

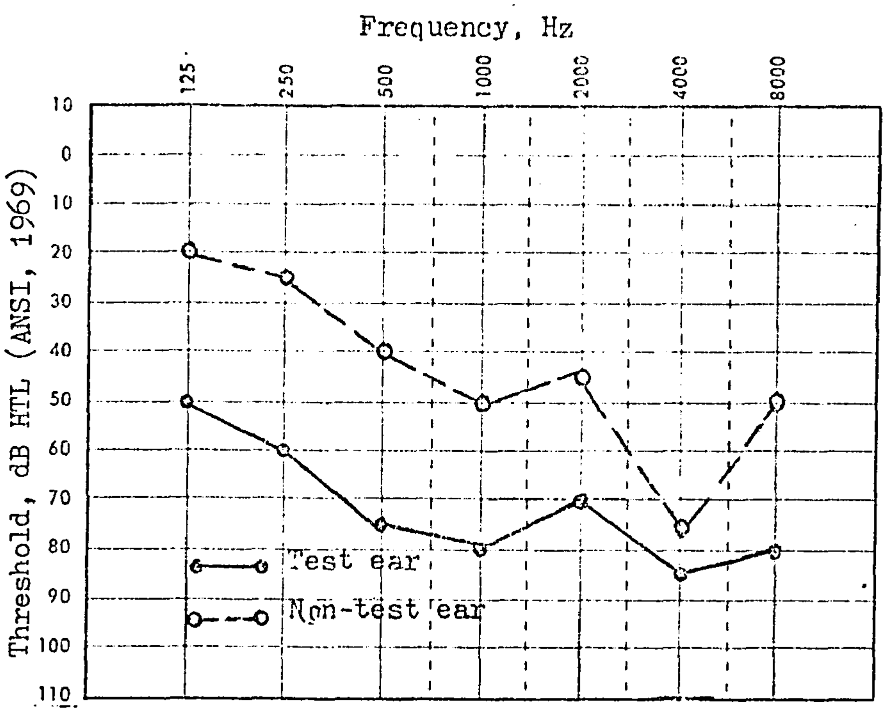
## Audiograms Obtained for Each of the Four Subjects of the Main Experiment




| | R. | L. |
|---|---|---|
| S.R.T. | 60 | 60 |
| DISCRIMINATION @ 90 dB | 68% | — |
| DISCRIMINATION @ 85 dB | — | 70% |
| T.D. (LDL) Speech | 105 | 100 |

Fig. 43. Pure tone thresholds and speech audiometry results for subject MS, male, age 61. Left ear was the test ear.

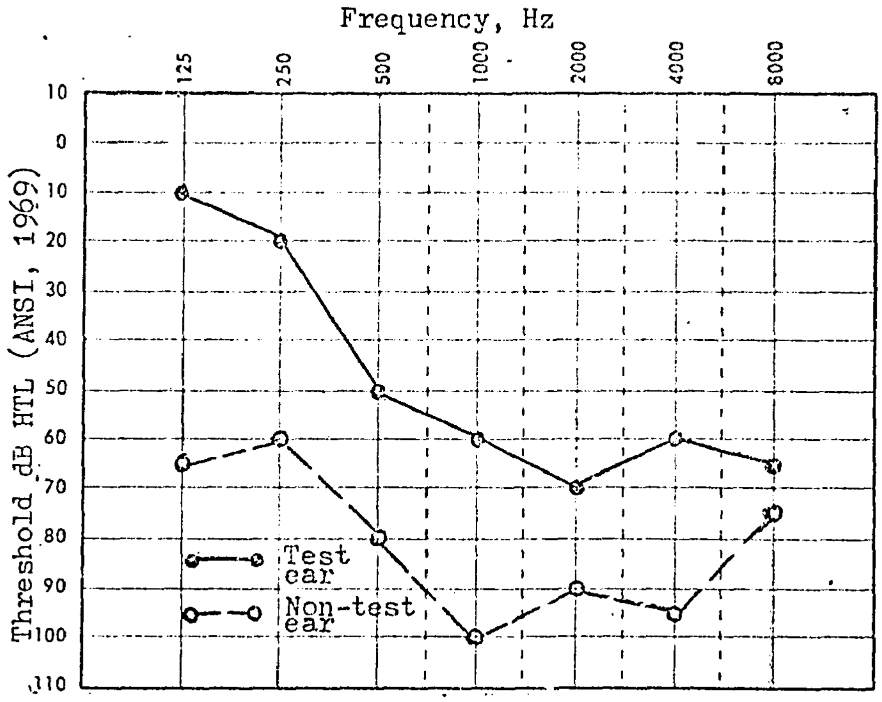
Appendix G--continued




| | R. | L. |
|---|---|---|
| S.R.T. | 80 | 45 |
| DISCRIMINATION @ 105 dB | 70% | — |
| DISCRIMINATION @ 80 dB | — | 76% |
| T.D. (LDL) Speech | 100 | 90 |

Fig. 44. Pure tone thresholds and speech audiometry results for subject HS, male, age 45. Test ear was the right ear.

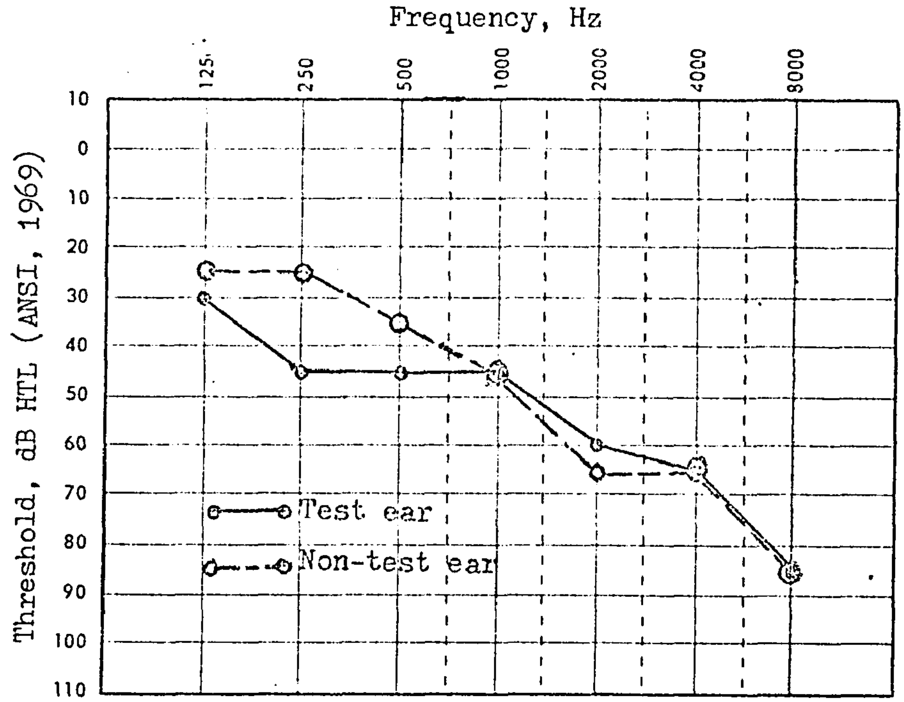
Appendix G--continued




| | | R. | L. |
|---|---|---|---|
| S.R.T. | | 55 | 85 |
| DISCRIMINATION @ | 80 dB | 66% | — |
| DISCRIMINATION @ | 100 dB | — | 6[illegible] |
| T.D.(LDL) Speech | | 95 | 105+ |

Fig. 45. Pure tone thresholds and speech audiometry results for subject FS, female, age 59 years. Test ear was the right ear.

Appendix G--continued

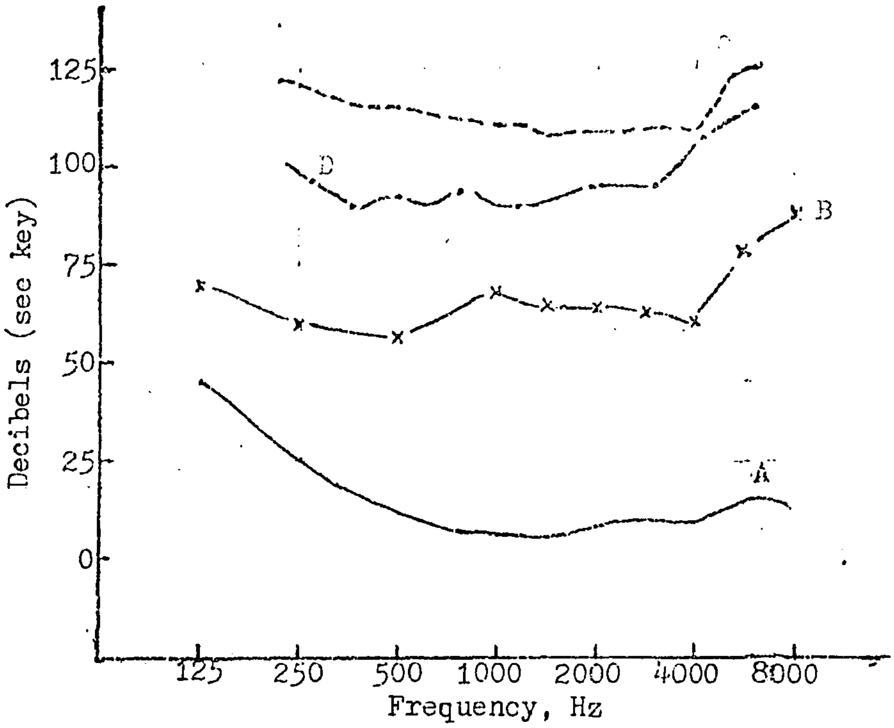



| | R. | L. |
|---|---|---|
| S.R.T. | 45 | 40 |
| DISCRIMINATION @ 80 dB | 56% | — |
| DISCRIMINATION @ 75 dB | — | 36% |
| T.D. (LDL) Speech | 97 | 90 |

Fig. 46. Pure tone thresholds and speech audiometry results for subject BB, female, age 69. Right ear was the test ear.

APPENDIX H

Pure Tone Thresholds, and LDL Frequency Response Curves, the Latter Both With and Without Correction for the Speech Spectrum, for each of the Four Subjects of the Main Experiment




KEY

A —— 1969 ANSI reference thresholds, dB SPL

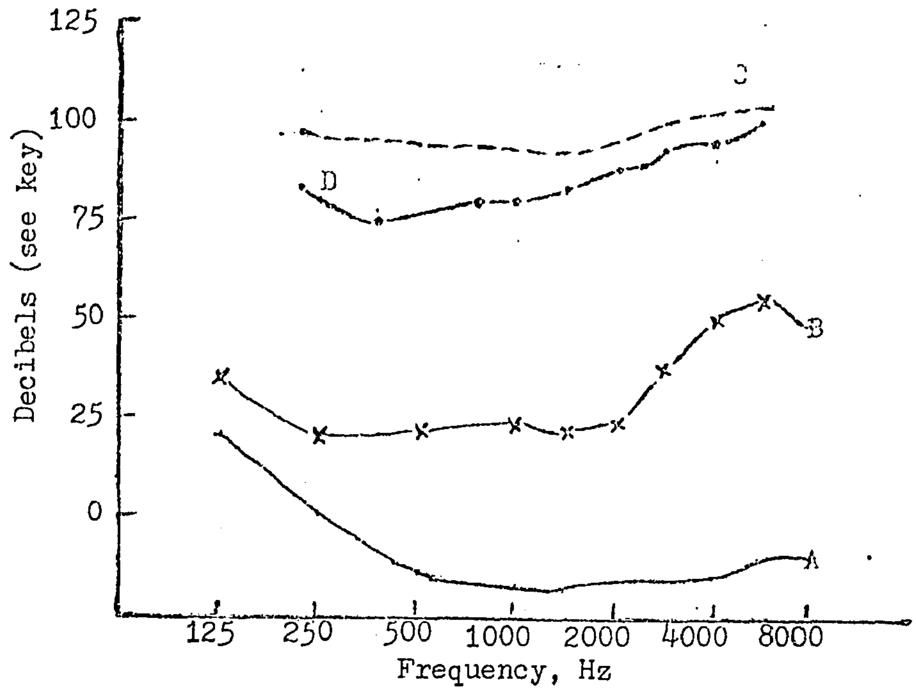
B x—x Thresholds for pure tones, dB SPL

C ---- LDL frequency response curve, dB SPL

D •—• Frequency-gain characteristic to put speech signal at LDL at all frequencies, i.e. curve C, corrected for speech spectrum. Values are given as relative dB, with overall gain determined by the subject's LDL for speech. This curve is the LDL frequency response curve for speech.

Fig. 47. Normal pure tone thresholds (ANSI, 1969); and pure tone thresholds, LDL frequency response curve, and speech spectrum at LDL, for subject MS.

Appendix H--continued




KEY

A —— 1969 ANSI reference thresholds, dB SPL

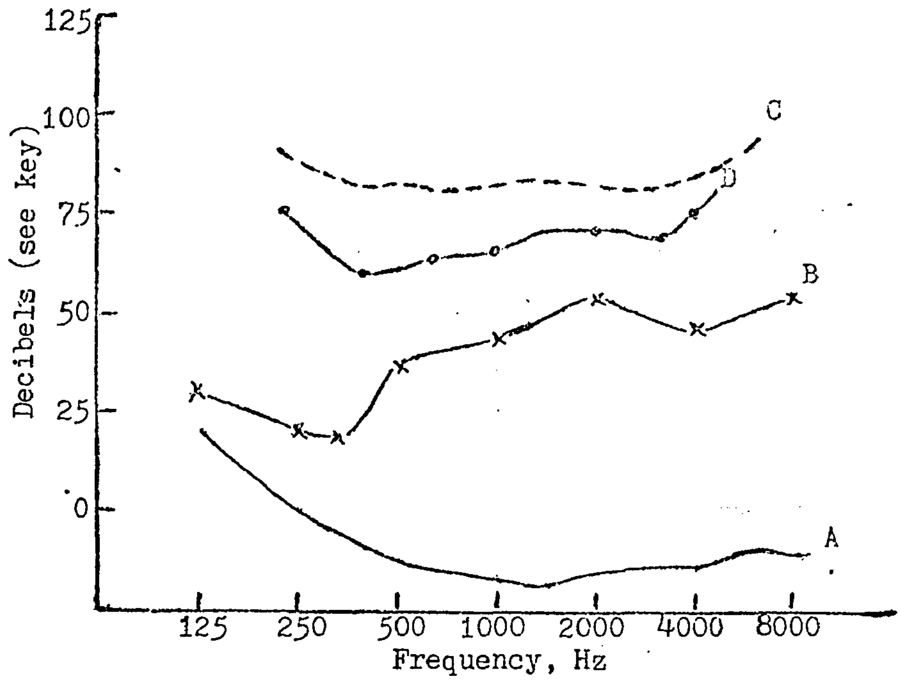
B x—x Thresholds for pure tones, dB SPL

C ---- LDL frequency response curve, dB SPL

D •—• Frequency-gain characteristic to put speech signal at LDL at all frequencies, i.e. curve C, corrected for speech spectrum. Values are given as relative dB, with overall gain determined by the subject's LDL for speech. This curve is the LDL frequency response curve for speech.

Fig. 48. Normal pure tone thresholds (ANSI, 1969); and pure tone thresholds, LDL frequency response curve, and speech spectrum at LDL, for subject HS.

Appendix H-continued




KEY

A—— 1969 ANSI reference thresholds, dB SPL

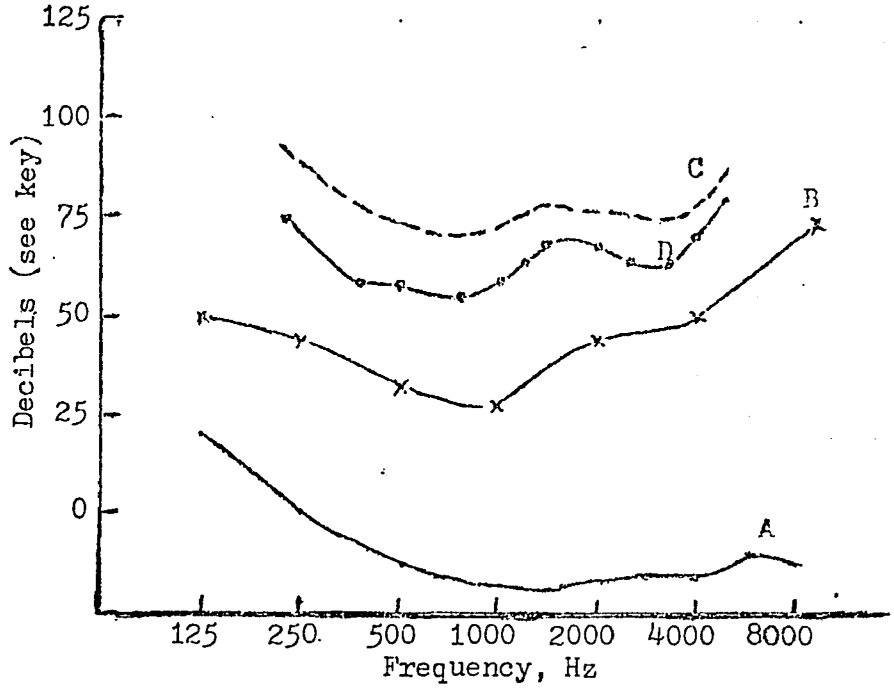
B×—× Thresholds for pure tones, dB SPL

C---- LDL frequency response curve, dB SPL

D•—• Frequency-gain characteristic to put speech signal at LDL at all frequencies, i.e. curve C, with correction for speech spectrum. Values are given as relative dB, with overall gain determined by the subject's wide-band LDL for speech. This curve is the LDL frequency response curve for speech.

Fig. 49. Normal pure tone thresholds (ANSI, 1969); and pure tone thresholds, LDL frequency response curve, and speech spectrum at LDL, for subject FS.

Appendix H--continued




KEY

A —— 1969 ANSI reference thresholds, dB SPL

B x—x Thresholds for pure tones, dB SPL

C ---- LDL frequency response curve, dB SPL

D •—• Frequency-gain characteristic to put speech signal at LDL at all frequencies, i.e. curve C, with correction for speech spectrum. Values are given as relative dB, with overall gain determined by the subject's wide-band LDL for speech. This curve is the LDL frequency response curve for speech.

Fig. 50. Normal pure tone thresholds (ANSI, 1969); and pure tone thresholds, LDL frequency response curve, and speech spectrum at LDL, for subject BB.

# APPENDIX I

## Measured Loudness Discomfort Levels for Subjects of the Main Experiment

Loudness Discomfort Levels (LDLs) and relative LDLs for speech, by subject, measured under six conditions (2 frequency responses x 3 bandwidths). The first part of the table shows the values for broad band condition ($L_b$), with values given in dB HTL. The second part of the table shows the values for each of two channels ($L_1$ and $L_2$, respectively), with values given relative to channel 1 ($L_1$).

| BANDWIDTH | FREQUENCY RESPONSE: 0 dB/octave, Subject: MS | HS | FS | BB | FREQUENCY RESPONSE: LDL frequency response, Subject: MS | HS | FS | BB |
|---|---|---|---|---|---|---|---|---|
| Broad band ($L_b$), 250 - 6000 Hz, dB HTL | 105 | 110 | 85 | 100 | 105 | 100 | 90 | 100 |
| Channel 1 ($L_1$), 250 - 1500 Hz | 0 | 0 | 0 | 0 | 0 | 0 | 0 | 0 |
| Channel 2 ($L_2$), 1500 - 6000 Hz, dB re $L_1$ | 10 | 5 | 0 | 0 | 5 | 10 | 10 | 10 |

# APPENDIX J

## Results of Figures 25 Through 34 in Tabular Form (Results of the Main Experiment)

Table 30. Percent correct scores obtained in both quiet and in noise for each of two frequency responses and three Amplitude Modes for the first factorial design, and one frequency response and five Amplitude Modes for the second factorial design. Scores are averaged over subjects and replications. The values given correspond to those shown in Fig. 25.

| AMPLITUDE MODE | LISTENING MODE | FREQUENCY RESPONSE Flat | FREQUENCY RESPONSE LDL |
|---|---|---|---|
| Non-compression, no peak clipping | Quiet | 61 | 66 |
| | S/N = 10 dB | 33 | 36 |
| Non-compression, with peak clipping | Quiet | 50 | 57 |
| | S/N = 10 dB | 27 | 31 |
| Compression, 3:1 | Quiet | 60 | 61 |
| | S/N = 10 dB | 27 | 37 |
| Compression, 3:1, 3:1 | Quiet | | 61 |
| | S/N = 10 dB | | 33 |
| Compression, 1.5:1, 3:1 | Quiet | | 63 |
| | S/N = 10 dB | | 31 |

Appendix J--continued

Table 31. Percent correct scores obtained in both quiet and in noise, for each of two frequency responses and four replications. Scores are averaged over subject and Amplitude Mode (three Amplitude Modes for flat frequency response, and five Amplitude Modes for LDL frequency response). Mean scores are also given. The values given correspond to those shown in Fig. 26.

| LISTENING MODE | FREQUENCY RESPONSE | Replication 1 | 2 | 3 | 4 | Mean |
|---|---|---|---|---|---|---|
| Quiet | Flat | 55 | 58 | 56 | 58 | 57 |
| | LDL | 56 | 63 | 65 | 63 | 62 |
| S/N = 10 dB | Flat | 29 | 29 | 31 | 28 | 29 |
| | LDL | 27 | 35 | 33 | 40 | 34 |

Appendix J--continued

Table 32. Percent correct scores obtained in quiet with flat frequency response, as a function of Amplitude Mode, replication, and subject. Mean scores are also shown. The values given correspond to those shown in Fig. 27.

| AMPLITUDE MODE | SUBJECT | Replication 1 | 2 | 3 | 4 | Mean |
|---|---|---|---|---|---|---|
| Non-compression, no peak clipping | MS | 55 | 65 | 58 | 43 | 55 |
| | HS | 77 | 83 | 74 | 78 | 78 |
| | FS | 62 | 49 | 43 | 57 | 53 |
| | BB | 54 | 57 | 52 | 62 | 56 |
| Non-compression, with peak clipping | MS | 39 | 42 | 53 | 45 | 45 |
| | HS | 53 | 51 | 67 | 80 | 63 |
| | FS | 45 | 51 | 56 | 36 | 47 |
| | BB | 38 | 50 | 44 | 40 | 43 |
| Compression, 3:1 | MS | 58 | 59 | 49 | 66 | 58 |
| | HS | 66 | 77 | 78 | 80 | 76 |
| | FS | 68 | 48 | 47 | 57 | 55 |
| | BB | 46 | 55 | 51 | 49 | 50 |

Appendix J--continued

Table 33. Percent correct scores obtained inquiet with LDL frequency response, as a function of Amplitude Mode, replication, and subject. Mean scores are also shown. The values given correspond to those shown in Fig. 28.

| AMPLITUDE MODE | SUBJECT | Replication 1 | 2 | 3 | 4 | Mean |
|---|---|---|---|---|---|---|
| Non-compression, no peak clipping | MS | 52 | 65 | 72 | 65 | 64 |
| | HS | 62 | 66 | 68 | 79 | 69 |
| | FS | 73 | 67 | 77 | 76 | 73 |
| | BB | 65 | 57 | 49 | 61 | 58 |
| Non-compression, with peak clipping | MS | 37 | 60 | 54 | 44 | 49 |
| | HS | 48 | 54 | 60 | 63 | 56 |
| | FS | 61 | 64 | 60 | 75 | 65 |
| | BB | 53 | 67 | 62 | 47 | 57 |
| Compression, 3:1 | MS | 43 | 47 | 59 | 58 | 52 |
| | HS | 66 | 60 | 77 | 77 | 70 |
| | FS | 54 | 65 | 70 | 62 | 63 |
| | BB | 53 | 55 | 70 | 58 | 59 |
| Compression, 3:1, 3:1 | MS | 40 | 60 | 55 | 60 | 54 |
| | HS | 75 | 80 | 80 | 75 | 78 |
| | FS | 38 | 69 | 67 | 44 | 55 |
| | BB | 54 | 56 | 58 | 47 | 54 |
| Compression, 1.5:1, 3:1 | MS | 61 | 50 | 57 | 61 | 57 |
| | HS | 63 | 69 | 75 | 61 | 67 |
| | FS | 69 | 69 | 67 | 72 | 69 |
| | BB | 50 | 69 | 48 | 63 | 58 |

Appendix J--continued

Table 34. Percent correct scores obtained in noise with flat frequency response as a function of Amplitude Mode, replication and subject. Mean scores are also shown. The values given correspond to those shown in Fig. 29.

| AMPLITUDE MODE | SUBJECT | Replication 1 | 2 | 3 | 4 | Mean |
|---|---|---|---|---|---|---|
| Non-compression, no peak clipping | MS | 32 | 26 | 30 | 45 | 33 |
| | HS | 54 | 51 | 52 | 62 | 54 |
| | FS | 22 | 25 | 23 | 29 | 25 |
| | BB | 18 | 33 | 23 | 20 | 23 |
| Non-compression, with peak clipping | MS | 24 | 16 | 36 | 20 | 24 |
| | HS | 38 | 46 | 43 | 41 | 42 |
| | FS | 26 | 25 | 27 | 24 | 25 |
| | BB | 19 | 35 | 18 | 7 | 17 |
| Compression, 3:1 | MS | 28 | 33 | 31 | 34 | 32 |
| | HS | 54 | 31 | 51 | 37 | 43 |
| | FS | 21 | 16 | 31 | 21 | 22 |
| | BB | 23 | 15 | 13 | 12 | 16 |

Appendix J--continued

Table 35. Percent correct scores obtained in noise with LDL frequency response as a function of Amplitude Mode, replication, and subject. Mean scores are also shown. The values given correspond to those shown in Fig. 30.

| AMPLITUDE MODE | SUBJECT | Replication 1 | 2 | 3 | 4 | Mean |
|---|---|---|---|---|---|---|
| Non-compression, no peak clipping | MS | 19 | 44 | 33 | 37 | 32 |
| | HS | 40 | 55 | 36 | 62 | 48 |
| | FS | 24 | 34 | 43 | 49 | 37 |
| | BB | 27 | 23 | 24 | 37 | 27 |
| Non-compression, with peak clipping | MS | 31 | 31 | 18 | 33 | 28 |
| | HS | 25 | 43 | 45 | 53 | 41 |
| | FS | 28 | 46 | 22 | 37 | 33 |
| | BB | 27 | 18 | 17 | 36 | 24 |
| Compression, 3:1 | MS | 46 | 36 | 44 | 39 | 41 |
| | HS | 50 | 50 | 61 | 45 | 52 |
| | FS | 27 | 21 | 42 | 42 | 33 |
| | BB | 8 | 42 | 28 | 24 | 25 |
| Compression, 3:1, 3:1 | MS | 26 | 37 | 27 | 32 | 31 |
| | HS | 42 | 39 | 38 | 42 | 41 |
| | FS | 22 | 28 | 42 | 60 | 37 |
| | BB | 15 | 41 | 21 | 17 | 23 |
| Compression, 1.5:1, 3:1 | MS | 19 | 24 | 43 | 43 | 31 |
| | HS | 23 | 46 | 38 | 47 | 38 |
| | FS | 20 | 26 | 36 | 32 | 28 |
| | BB | 23 | 28 | 20 | 32 | 25 |

Appendix J--continued

Table 36. Percent correct scores obtained in quiet with flat frequency response, as a function of Amplitude Mode, subject and subtest. The values given correspond to those shown in Fig. 31.

| AMPLITUDE MODE | SUBJECT | Subtest /uv/ | /uv̄/ | /iv̄/ | /iv/ |
|---|---|---|---|---|---|
| Non-compression, no peak clipping | MS | 18.3 | 67.3 | 79.5 | 56.3 |
| | HS | 65.8 | 91.6 | 89.1 | 59.6 |
| | FS | 40.6 | 71.7 | 50.1 | 48.4 |
| | BB | 43.3 | 66.7 | 58.5 | 56.3 |
| Non-compression, with peak clipping | MS | 18.9 | 70.9 | 58.5 | 34.0 |
| | HS | 67.6 | 79.0 | 67.5 | 36.4 |
| | FS | 23.7 | 62.6 | 62.7 | 40.6 |
| | BB | 48.5 | 58.5 | 39.4 | 26.5 |
| Compression, 3:1 | MS | 30.5 | 73.6 | 58.5 | 68.9 |
| | HS | 66.0 | 84.5 | 90.1 | 56.3 |
| | FS | 45.3 | 77.4 | 55.1 | 40.6 |
| | BB | 48.3 | 60.7 | 45.8 | 46.8 |

Appendix J--continued

Table 37. Percent correct scores obtained in quiet with LDL frequency response, as a function of Amplitude Mode, subject and subtest. The values given correspond to those shown in Fig. 32.

| AMPLITUDE MODE | SUBJECT | Subtest /uv/ | /uv̄/ | /iv̄/ | /iv/ |
|---|---|---|---|---|---|
| Non-compression, no peak clipping | MS | 39.0 | 80.8 | 77.8 | 53.2 |
| | HS | 62.7 | 79.0 | 73.2 | 59.5 |
| | FS | 54.7 | 89.1 | 79.5 | 65.8 |
| | BB | 43.5 | 67.3 | 75.5 | 43.7 |
| Non-compression, with peak clipping | MS | 20.0 | 67.3 | 50.0 | 59.6 |
| | HS | 48.3 | 79.5 | 52.0 | 43.7 |
| | FS | 52.1 | 85.9 | 71.2 | 48.4 |
| | BB | 51.6 | 69.4 | 65.2 | 41.7 |
| Compression, 3:1 | MS | 28.9 | 67.3 | 54.3 | 56.3 |
| | HS | 53.2 | 86.8 | 75.4 | 62.5 |
| | FS | 42.2 | 85.9 | 80.4 | 37.3 |
| | BB | 56.3 | 58.5 | 82.8 | 36.4 |
| Compression, 3:1, 3:1 | MS | 31.1 | 77.2 | 56.4 | 50.0 |
| | HS | 59.5 | 83.3 | 86.8 | 78.4 |
| | FS | 27.1 | 75.4 | 67.8 | 48.5 |
| | BB | 52.9 | 64.8 | 56.3 | 42.2 |
| Compression, 1.5:1, 3:1 | MS | 34.1 | 79.0 | 66.8 | 46.8 |
| | HS | 67.7 | 65.3 | 79.5 | 54.7 |
| | FS | 43.5 | 93.8 | 71.2 | 59.4 |
| | BB | 50.0 | 64.8 | 67.3 | 48.4 |

Appendix J--continued

Table 38. Percent correct scores obtained in noise with flat frequency response, as a function of Amplitude Mode, subject and subtest. The values given correspond to those shown in Fig. 33.

| AMPLITUDE MODE | SUBJECT | Subtest /uv/ | /uv̄/ | /iv̄/ | /iv/ |
|---|---|---|---|---|---|
| Non-compression, no peak clipping | MS | 11.6 | 36.8 | 50.0 | 37.5 |
| | HS | 42.1 | 57.1 | 73.8 | 45.3 |
| | FS | 7.6 | 24.4 | 37.4 | 34.2 |
| | BB | 16.4 | 26.2 | 35.2 | 16.6 |
| Non-compression, with peak clipping | MS | 14.6 | 29.1 | 30.6 | 21.6 |
| | HS | 32.7 | 41.6 | 62.8 | 31.1 |
| | FS | 7.6 | 54.5 | 31.2 | 14.9 |
| | BB | 22.1 | 17.2 | 18.4 | 16.8 |
| Compression, 3:1 | MS | 20.0 | 41.5 | 30.8 | 35.4 |
| | HS | 39.1 | 47.4 | 57.0 | 29.6 |
| | FS | 7.6 | 47.8 | 28.7 | 11.6 |
| | BB | 13.4 | 22.8 | 15.0 | 11.7 |

Appendix J--continued

Table 39. Percent correct scores obtained in noise with LDL frequency response, as a function of Amplitude Mode, subject, and subtest. The values given correspond to those shown in Fig. 34.

| AMPLITUDE MODE | SUBJECT | Subtest /uv/ | /u$\bar{v}$/ | /i$\bar{v}$/ | /iv/ |
|---|---|---|---|---|---|
| Non-compression, no peak clipping | MS | 25.8 | 38.9 | 37.8 | 29.4 |
| | HS | 37.2 | 54.3 | 45.8 | 56.5 |
| | FS | 13.9 | 52.2 | 38.7 | 48.4 |
| | BB | 13.0 | 52.2 | 28.5 | 20.2 |
| Non-compression, with peak clipping | MS | 11.6 | 37.1 | 47.8 | 20.3 |
| | HS | 32.3 | 30.8 | 63.2 | 38.8 |
| | FS | 17.3 | 71.3 | 29.8 | 18.2 |
| | BB | 18.5 | 34.9 | 19.6 | 24.8 |
| Compression, 3:1 | MS | 27.5 | 45.7 | 58.7 | 34.3 |
| | HS | 38.6 | 39.4 | 75.5 | 51.8 |
| | FS | 18.9 | 45.8 | 43.1 | 25.0 |
| | BB | 18.2 | 26.4 | 26.8 | 27.8 |
| Compression, 3:1, 3:1 | MS | 20.0 | 39.3 | 26.8 | 37.3 |
| | HS | 34.9 | 67.2 | 21.0 | 40.5 |
| | FS | 14.6 | 61.1 | 26.0 | 52.2 |
| | BB | 15.3 | 29.2 | 22.6 | 24.8 |
| Compression, 1.5:1, 3:1 | MS | 8.8 | 47.7 | 32.8 | 42.1 |
| | HS | 16.9 | 52.1 | 34.8 | 51.6 |
| | FS | 14.0 | 50.7 | 21.8 | 30.8 |
| | BB | 23.4 | 50.0 | 8.4 | 25.7 |

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