THE EFFECT OF AMPLITUDE COMPRESSION
ON THE INTELLIGIBILITY OF SPEECH
FOR PERSONS WITH SENSORINEURAL HEARING LOSS

by

RICHARD PAUL LIPPMANN

B.S., Polytechnic Institute of Brooklyn
(1970)

S.M., Massachusetts Institute of Technology
(1973)

E.E., Massachusetts Institute of Technology
(1974)

SUBMITTED IN PARTIAL FULFILLMENT
OF THE REQUIREMENTS FOR THE
DEGREE OF

DOCTOR OF PHILOSOPHY

at the

MASSACHUSETTS INSTITUTE OF TECHNOLOGY

February, 1978

ARCHIVES
MASS INST TECH
MAY 15 1978
LIBRARIES

Signature redacted

Signature of Author ....
Department of Electrical Engineering and
Computer Science, February 1, 1978

Signature redacted

Certified by ....
Thesis Supervisor

Accepted by ....
Chairman, Departmental Committee on Graduate Students

THE EFFECT OF AMPLITUDE COMPRESSION

ON THE INTELLIGIBILITY OF SPEECH

FOR PERSONS WITH SENSORINEURAL HEARING LOSS

by

Richard Paul Lippmann

Three types of amplitude compression have been suggested for use in hearing aids for persons with sensorineural hearing loss: limiting, automatic volume control and syllabic compression. Recent research has focused on multiple-channel syllabic compression which modifies the short-term level variations of speech segments in order to match the dynamic range of speech to the residual dynamic range of impaired listeners. This research has been inconclusive, however. The goal of the present thesis was to further explore the usefulness of multiple-channel syllablic compression.

Experiments were conducted on 5 listeners with sensorineural impairments and reduced dynamic ranges using 16 channel, computer controlled, amplitude compression systems. Each subject was tested with two compression systems and, for reference purposes, 4 linear systems. One of the compression systems was adjusted to restore normal equal loudness contours; the other employed reduced high-frequency emphasis and reduced compression ratios. The 4 linear systems differed only in their frequency-gain characteristic (orthotelephonic plus 3 characteristics with high-frequency emphasis that were expected to produce better results than orthotelephonic). The 6 systems were compared on each of the 5 subjects using nonsense CVC monosyllables and nonsense sentences spoken by male and female talkers and presented in quiet/anechoic and noisy/reverberant environments. In addition, systems were compared using standard word and sentence tests. All tests were performed using circumaural earphones that were calibrated relative to free field. All materials were presented at the most comfortable level for each listener in the main experiments and reduced levels were investigated in supplementary experiments. The results of the main experiments were: (1) the linear systems with high-frequency emphasis performed substantially better than the orthotelephonic system; and (2) neither of the two compression systems performed substantially better than the best linear system. Also, an analysis of consonant confusions made in quiet indicated that most errors for both the linear and compression systems were caused by errors in perception of the feature place. In considering these results, it should be noted that (a) the long-term level of the speech material was held constant before processing and there was generally minimal item-to-item level variation in this material, (b) the subjects suffered from only moderate losses, (c) there was minimal training with compressed materials, and (d) only two compression systems were tested. The results concerning the comparison of compression to linear amplification are consistent with those of other recent studies if one takes into account differences in the linear systems to which compression systems were compared. In the supplementary experiments it was found that compression performed better than linear amplification when the speech level before processing was reduced. This also occurred in the main experiments for the one speech test with a large item-to-item level variation. The advantage of compression under these conditions was greatest for listeners with more severe low-frequency losses.

THESIS SUPERVISOR: Louis D. Braida
TITLE: Professor, Department of Electrical Engineering

ACKNOWLEDGEMENTS

This thesis marks the end of my seventh year at M.I.T. and the end of a long experimental program in which many people took part. Throughout this work L. Braida and N. Durlach have provided continual guidance, review, and advice and seen to it that I received technical and financial support. Their recent and past efforts in reviewing this thesis have been heroic and they have provided valuable detailed reviews of all published forms of this work. E. Villchur has also been involved in this work from the beginning and his research was, in fact, one of the sparks that initiated this work. He provided much encouragement and detailed advice on many aspects of this research and he taught me much during the time he was working in our lab. K. Stevens has also provided advice concerning aspects of this work related to speech and speech perception. The consultants on the hearing-aid grant (R. Bilger, A. Boothroyd, D. Klatt, H. Levitt, D. Ling, J. Miller, R. Morrison, J. Pickett, M. Schultz, D. Ward, and O. Welsh) also provided advice.

Many graduate and undergraduate students have helped on various aspects of this research. C. Reed, M. Clements, and Y. Perlmutter helped out in extremely tedious recording and splicing sessions. P. Peterson, M. Doherty, and S. Rubin helped out in experiments performed to validate the speech tests used. M. Bush, D. Christopher, P. Herman, and B. Hicks were talkers "without significant regional characteristics", as was C. Lippmann. Mickey Dupree helped in dubbing and measuring the levels of the speech materials. J. Boddie, D. Cuddy, M. Krasner, J. Homan, P. Herman, and M. Davis all contributed to the design of the 16-channel speech processor. P. Herman and M. Davis were also both very helpful in solving technical problems. D. Mook provided a program for analysis of consonant confusions, a sounding board for statistical questions, and tea. Many others, including J. Berliner, S. Colburn, B. Rabinowitz, M. Picheny, B. Hicks, and R. Domnitz, provided encouragement and help.

Ionia Lewis has been invaluable throughout my stay at M.I.T. She typed this thesis into the computer and has typed talks and other papers which flowed from this work and has been extremely helpful and encouraging.

In the past few months, many people have helped finish this document. First, I would like to thank all the people from the First United Presbyterian Church of Cambridge for their prayers and also for their help. Many a night there were 2-4 people at terminals editing in TECO, proofing text, and feeding the computer-controlled typewriter. I would especially like to thank Bill Long, who was here yesterday and the day before till midnight, and Sue Wagerman who edited some of the more atrocious sections of this thesis, proofed, and was fun to work with. I would also like to thank Steve Bates, Carol Bishop, Elaine Sullivan, Gordan Weekly and Eliza Van Buskirk who were from church and helped out at the last minute. A number of people from M.I.T. also helped edit. L. Braida took responsibility for updating the review of amplitude compression. Steve DeGennaro helped edit till all hours to get a copy of this thesis ready for my defense and Mike Picheny helped edit many sections. They both also provided needed encouragement. Denny Freeman helped by answering questions about the computer and getting the system up whenever it crashed. Ken Krieg helped the last two nights during the wee hours to proof and type out text.

I would also like to thank my subjects, ED, IK, FM, PR, ES, HS for coming in and putting up with long hours of tedious work. I would especially like to thank HS for his time, encouragement, and advice on gardening.

The contribution made in my life and on this thesis by Carolyn Lippmann is immeasurable. She gave keen advice on many aspects of this work, provided financial support and bore our two children during the period in which this thesis was in the process of becoming a thesis.

Organizational Note

It has become a tradition of the Auditory Perception Group at M.I.T. to write doctoral theses in a number of parts which are in a format (albeit not a length) intended to be compatible with papers submitted to the Journal of the Acoustical Society of America. In keeping with this tradition this thesis is written in three parts. The first part is a review of previous research on amplitude compression and is part of a monograph to be published by the Journal of Speech and Hearing Research. The second part describes the major experimental work of the thesis. The third part describes the free-field calibration of the circumaural earphones used in this thesis and will be submitted to the Journal of the Acoustical Society of America.

Table of Contents

Title.. 1

Abstract... 2

Acknowledgements... 4

Organizational Note...................................... 6

Table of Contents.. 7

PART 1

PREVIOUS RESEARCH ON AMPLITUDE COMPRESSION............. 14

A. Preliminary Remarks................................... 14

B. Characteristics of Compression Amplifiers........... 17

1. General Characteristics........................... 17

2. Characteristics of Limiting....................... 24

3. Characteristics of Automatic Volume Control..... 28

4. Characteristics of Syllabic Compression......... 30

C. Review of Syllabic Compression...................... 34

1. General Comments.................................. 34

2. Detailed Reviews.................................. 38

D. Concluding Remarks................................... 63

Figure Captions.. 66

Figures.. 67

Tables.. 75

References.. 78

PART 2

A STUDY OF MULTICHANNEL AMPLITUDE COMPRESSION
AND LINEAR AMPLIFICATION
FOR PERSONS WITH SENSORINEURAL HEARING LOSS............. 91

Introduction.. 91

1. Subjects... 95

1.1 Clinical Measurements................................. 95

1.2 Laboratory Measurements............................... 96

1.2a Thresholds... 97

1.2b Loudness Discomfort Levels........................... 98

1.2c Loudness Balances.................................... 98

1.2d Average Equal Loudness Contours..................... 99

2. Experiment I... 100

2.1 Procedures.. 101

2.2 Systems... 105

2.2a Linear Systems - Design.............................. 105

2.2b Compression Systems - Design......................... 106

2.2c Implementation....................................... 108

2.2d System Characteristics............................... 109

2.2d.1 Output 10% Levels.................. 109

2.2d.2 Normalized Functional Gain......... 111

2.2d.3 Compression Curves................. 112

2.2d.4 Most Comfortable Levels............ 115

2.3 Speech Tests................................ 116

2.3a Materials............................. 116

2.3b Recording Procedures
and Acoustic Characteristics.............. 117

2.3c Responses............................. 120

2.3d Scoring............................... 122

2.4 Results..................................... 122

2.4a Variability of Scores
and of Quality Ratings.................... 122

2.4b Relationship Between Word or Syllable
and Phoneme Scores........................ 124

2.4c Overall Comparisons Between Conditions... 125

2.4c.1 Phoneme Scores..................... 125

2.4c.2 Quality Ratings.................... 127

2.4c.3 Consonant and Vowel Scores......... 128

2.4d Comparisons Between Systems............. 129

2.4d.1 Phoneme Scores..................... 129

2.4d.1a Linear Systems.................. 130

2.4d.1b Compression Systems............. 131

2.4d.1c Linear Amplification
versus Compression........................ 132

2.4d.2 Quality Ratings.................... 134

2.4d.3 Consonant Confusions.............. 136

3. Experiment II.. 139

3.1 Speech Tests.. 140

3.2 Methods and Procedures........................ 141

3.3 Results and Discussion........................ 142

3.3a Most Comfortable Levels.................. 142

3.3b Speech Tests................................ 142

3.3b.1 Overall Level of Scores............ 143

3.3b.2 ORTHO versus OMCL.................. 143

3.3b.3 COMCL versus OMCL.................. 144

3.3b.4 Word Scores.......................... 145

3.3b.5 Quality Judgements.................. 145

4. Experiment III...................................... 146

4.1 Methods and Procedures........................ 146

4.2 Results and Discussion........................ 147

5. Discussion.. 149

5.1 Summary of Present Study...................... 149

5.2 Comparison to Previous Research on Linear Amplification...................... 154

5.3 Comparison to Previous Research on Multiple Channel Syllabic Compression...... 156

5.4 Overview - Linear Amplification.............. 161

5.5 Overview - Multiple Channel Syllabic Compression.......................... 163

Footnotes.................................... 168

Figure Captions.............................. 175

Figures...................................... 179

Tables....................................... 199

References................................... 210

Appendices

1. Subject Histories.......................... 218

2. Pre-Experimental Clinical Results............ 223

3. Threshold and Loudness Discomfort Level Measurements........................ 230

4a. Procedures for Monaural Loudness Balances and Alternate Binaural Loudness Balances...... 232

4b. Individual Monaural Loudness Balances and Alternate Binaural Loudness Balances.......... 236

5a. Methods Used to Determine Equal Loudness Contours and Relate these Contours to Normal Contours.. 244

5b. Individual Phon Equal Loudness Contours....... 246

6. 10% Cumulative Levels of Speech Materials..... 252

7. Dunn and White 10% Speech Levels............. 255

8. Most Comfortable Presentation Levels chosen for Speech in Experiment I.................. 256

9. 10% Output Levels in Experiment I for Individual Subjects...................... 261

10. Normalized Functional Gain of Systems in

Experiment I for Individual Subjects.......... 266
11. Compression Ratios for the EL and COMCL Systems in Experiment I.............................. 271
12. Individual Compression Curves of the COMCL and EL Systems in Experiment I................... 272
13. Percent Phoneme Correct Scores and Quality Judgements in Experiment I............ 283
14. Consonant Confusion Matrices................. 297
15. Most Comfortable Presentation Levels Chosen in Experiment II...................... 305
16. Word and Phoneme Scores and Quality Judgements in Experiment II........... 306
17. Phoneme Scores and Most Comfortable Levels in Experiment III............................ 309

PART 3

FREE-FIELD CALIBRATION OF A NEW CIRCUMAURAL EARPHONE... 311

Introduction.. 312
I. Subjects and Equipment....................... 314
II. Method.. 315

III. Results and Discussion........................ 318

A. Probe Microphone and

Threshold Measurements...................... 318

B. Free-Field Calibration...................... 319

Footnotes.. 324

Table.. 325

Figure Captions.................................... 326

Figures.. 328

References... 332

Part 1

PREVIOUS RESEARCH ON AMPLITUDE COMPRESSION

A. PRELIMINARY REMARKS

There are a number of indications that listeners with sensorineural impairments require some form of nonlinear amplitude processing as well as linear amplification. Many of these listeners exhibit increased detection thresholds without correspondingly increased "saturation" or "discomfort" thresholds. Thus, when amplification is used to overcome loss of absolute sensitivity, additional amplitude processing is required to prevent intense sounds from causing pain or discomfort. Also, for some listeners with sensorineural impairments, speech intelligibility rises to a maximum and then falls as intensity is increased. Thus, to achieve best speech perception, the overall gain of the amplifying system used by such listeners must be controlled to assure that the proper output level is maintained as the input level varies. Furthermore, many listeners with sensorineural impairments exhibit abnormal loudness functions. Linear amplification can, in general, restore only one equal-loudness contour to normal for such listeners. To correct the entire set of equal-loudness contours, some form of frequency-dependent nonlinear amplitude processing is required. Finally, past research on linear amplification suggests that the shape of the frequency-gain characteristic must be varied as a function of input level to ensure best speech reception. Clearly, nonlinear processing is required to achieve this variation.

There are three types of nonlinear amplitude processing, with distinct level-temporal characteristics, that have been suggested in order to overcome the above limitations of linear amplification. One type, limiting, is used to protect the ear from high or painful peak sound levels. Limiters should not affect sounds below a critical level. They must act very rapidly in order to deal with intense sound bursts having rapid onsets. A second type, automatic volume control (AVC), is used to keep the long-term average presentation level (measured over time intervals corresponding to phrases or sentences) near that corresponding to maximum intelligibility. Since the action of automatic volume control should be slow compared to the sound variations that occur within syllables, it can be regarded as a slowly varying linear amplifier. A final type of amplitude processing, which we shall call "syllabic compression", is concerned with altering the short-term intensity relations among speech elements to improve intelligibility. Syllabic compressors must act rapidly enough to respond to level variations associated with transitions between speech sounds. The level characteristics of syllabic compressors are typically chosen to match the range of speech amplitudes to the residual dynamic range of impaired listeners.

Considerable confusion presently exists in the technical and commercial literature concerned with the use of nonlinear amplitude processing in aids for the impaired. Often, attempts are made to design a single processor, with one fixed level-temporal characteristic, to perform two or three of the above functions. In general, this is not possible. For example, the slow gain variation required for automatic volume control is inappropriate for limiting or syllabic compression. Also, the level characteristics of automatic volume control, which are keyed to the average intensity of speech, may conflict with those of a limiter, which are based on peak levels, and with those of syllabic compression, which are based on matching intensity ranges. It is, however, possible to incorporate these three separate functions in a single aid, provided they are arranged in cascade as shown in Fig. 1.

Our review of research on amplitude compression for hearing aids focuses on syllabic compression, the effects of which are least understood and constitute a focal point of current hearing aid research. Because the three types of processing are so often confused, however, we include brief sections concerned with the properties of limiters and automatic volume controls. Also, we preface this review with a discussion of the general characteristics of the amplifiers typically included in compression systems.

Our discussion is divided into three sections: (B) Characteristics of Compression Amplifiers, (C) Detailed Review of Syllabic Compression, and (D) Concluding Remarks.

B. CHARACTERISTICS OF COMPRESSION AMPLIFIERS

1. General Characteristics

The purpose of a compression amplifier is to reduce the dynamic range of a class of input signals. A diagram of the type of system that we will refer to as a "compressor" or "compression amplifier" is presented in Fig. 2. In this system the signal is amplified by a low-distortion variable-gain linear amplifier. The gain of the amplifier is controlled by a level detector which senses the input or output signals, or both. Because amplifier gain depends on signal level, the characteristics of the compression amplifier differ from those of a conventional linear amplifier. In particular, although a compression amplifier and a linear amplifier both transform a given sinusoidal input into a sinusoidal output (of the same frequency), the relation of the output amplitude to the input amplitude is different in the two cases. More specifically, if the input amplitude of a sinusoid of frequency f is X, the output amplitude for a linear system is of the form G(f)X [where G(f) is the frequency-gain characteristic], whereas the output amplitude for a compressor is of the form A(f,X). In other words, the output of a compressor differs from that of a linear system in that the output amplitude depends nonlinearly on the input amplitude. Ideally, the output of a compressor is free of distortion components and the static gain characteristics of a compressor are described by the function A(f,X). A plot of the output level versus the input level (typically, in dB vs dB) at one frequency describes A(f,X) at that frequency and is called a compression curve (see Fig. 3). The compression ratio (CR) is the inverse slope of the compression curve (small-range dB change in input divided by the resulting change in output); the compression threshold is the input level at which the compression ratio becomes greater than one; the compression range is the range of input levels above the compression threshold over which CR > 1 and the output is essentially undistorted; and expansion denotes operation in a region on the compression curve where CR < 1. Finally, it should be noted that the phrase "frequency-gain characteristic" has meaning for a compression system only when the input level [X(f)] is specified, and that the frequency-gain characteristic for one input level can be derived from the characteristic for another level by the use of the compression curves. Also, the reference level used to define the frequency-gain characteristic is often chosen (sometimes implicitly) to be the maximum level that one expects to encounter in the use of the system. When a given characteristic is achieved by adding filtering at the output of the compressor, the phrase "post-compression equalization" is often used.

Static characteristics describe only part of the behavior of a compressor. When the input level of a compressor varies rapidly, as occurs with speech, the dynamic characteristics of a compressor become important. These characteristics are typically described by the attack time, TA, and the release time, TR, which roughly describe the output envelope after a "step" change in the input envelope (see Fig. 4). There are number of definitions of TA and TR. (e.g., Carter, 1964; Blesser, 1969; ANSI, 1976). The following are similar to those expressed in ANSI (1976) except the input level change is 20 dB instead of 25 dB. The attack time TA is the time required for the output of a compressor to come within 2 dB of the level specified by the compression curve after the input level increases by at least 20 dB to a level above the compression threshold. The release time TR is the time required for the output of a compressor to come within 2 dB of the level specified by the compression curve after the input level decreases at least 20 dB from a level that was above the compression threshold.

These definitions are illustrated in Fig. 4. Unfortunately, the time constants TA and TR do not completely describe the dynamic behavior of a compressor. This behavior also depends on the amount and rate of input level change, the input level, the input spectrum, the compression ratio, and the recent history of input level changes. The attack and release time do, however, roughly describe the dynamic behavior of simple compression amplifiers.

There are two fundamental limitations on attack and release times. First, both TA and TR must be much smaller than the time between successive input level changes that are meant to be controlled, so that the static characteristics apply. Second, either TA or TR must be longer than 2-5 periods of the lowest frequency amplified by the system so that compressor action responds to the envelope rather than the instantaneous waveform. If both TA and TR are short, low-frequency components will be distorted (Carter, 1964; Blesser, 1969; Noble and Bird, 1969). Often TA is deliberately chosen to be very short (TA < 1 msec) to prevent excessive overshoot in the output envelope (which might cause discomfort or annoyance) when the input increases rapidly. When this is the case, both of the above limitations can usually be satisfied simultaneously for speech signals by setting TR greater than or equal to 20 msec (e.g., Edgardh, 1952; Johansson, 1973; Villchur, 1973). The upper limit on the choice of TR depends, of course, on the intended function of the compressor.

Amplitude processing directed toward the same goals as compression amplification (and often imprecisely referred to as compression amplification in the hearing-aid literature) is sometimes accomplished using a "nonlinear distortion system." Such a system includes instantaneous nonlinear distortion together with pre- and post-distortion filtering. It differs from the compression amplification system shown in Fig. 2 in that the response to a sinusoid is generally not a sinusoid (even in the steady state). The nonlinear distortion typically involves signed square rooting, cube rooting, or, as shown in Fig. 5, symmetric peak clipping.

The operation of a distortion system is determined by the specific nonlinearity chosen and the pre- and post-distortion filtering used. A compression curve can be plotted for a distortion system (total rms output, or output at input frequency, versus input), but must be interpreted with care because the output signal may be extremely distorted. In multiband distortion systems, out-of-band harmonic and intermodulation distortion components can effectively be eliminated by pre- and post-distortion filtering. However, in-band harmonic and intermodulation components are not eliminated unless the bands are very narrow. The terms attack and release time do not apply to a distortion system.

There is one form of degradation that necessarily accompanies both compression amplification and processing by nonlinear distortion systems: an increase in background noise during periods when the signal is weak or absent. The only sure technique for reducing this degradation, the effects of which are not yet well understood, is to provide a high input signal-to-noise ratio by (e.g., by appropriate microphone selection and placement). In addition, the effect of the noise should always be minimized by such basic design considerations as selecting a compression ratio that is no higher than needed, selecting a compression threshold that is no lower than needed, and selecting a compression curve with a sharp "knee" at the compression threshold. Multichannel compressors provide considerable flexibility in these choices because compression characteristics in each channel can be adjusted to take advantage of spectral variations in signal and noise levels. A number of investigators have also suggested more elaborate techniques, such as modifying the compressor to provide expansion below the compression threshold (Blesser, 1969; Villchur, 1973) or reducing the overall gain when the input is noise alone (Parker, 1953; Hellwarth and Jones, 1967).

Two purely practical problems are encountered when constructing compression hearing aids. First, elements that perform the required functions must be obtained. In the past, the unavailability of such elements has limited the class of compressors that could be constructed economically to those with a high compression ratio (CR 4). In the last 10 years, however, the introduction of high quality integrated circuits (full-wave rectifiers, rms detectors, log and antilog converters, and multipliers) has overcome this limitation. Second, the compressor must be incorporated in a low-cost portable hearing aid without compromising the desired characteristics. A number of unnecessary distortions in commercial aids (Krebs, 1972; Nabelek, 1973, 1975; Burnett and Bassin, 1976; Burnett and Schweitzer, 1977) reflect the failure to solve this problem adequately. These distortions include: "thump", a low-frequency transient that accompanies gain changes and can cause extreme distortion (Hathaway, 1950); harmonic distortion associated with nonlinearity in the gain-controlling element; distortion at low frequencies caused by improper filtering of the gain-control signal; "holes" in the output caused by excessive gain reduction after an increase in input (Blesser and Kent, 1968); excessive overshoot caused by an attack time that is too long; and excessive noise during quiet periods caused by improper adjustment of the compression threshold or compression ratio, or by noise introduced by the gain-controlling device. Many of these unnecessary distortions can be particularly detrimental to speech perception because they tend to occur during the level changes that typically accompany transitions between vowels and consonants. Although theories and electronic devices exist that permit these distortions to be reduced or eliminated (Carter, 1964; Dolby, 1967; Blesser and Kent, 1968; Blesser, 1969; Burwen, 1971; Blackmer, 1972), they have been applied to hearing-aid construction only recently. Most commercial compression aids available today (e.g., Viet, 1973) appear to be designed to use a minimum number of components rather than to eliminate distortions.

2. Characteristics of Limiting

The purpose of limiting is to prevent the output sound level from becoming dangerous or uncomfortable (e.g., Davis et at., 1947; McCandless, 1973). The importance of controlling level has been underscored by Wallenfels (1967) who stressed that aids which produced uncomfortable sounds were simply not used by impaired listeners. Output sound levels can be controlled by either a peak clipper or a compressor-limiter. A peak clipper distorts the instantaneous waveform, never allowing the instantaneous output to exceed a critical level, as shown in Fig. 5. A compressor-limiter rapidly lowers the gain when the input exceeds a critical level and holds the gain down until the input has decreased sufficiently. A compression curve for a compressor-limiter with a maximum output level of about 100 dB SPL is shown in Fig. 6. The linear gain of the limiter in Fig. 6 is assumed to be unity for signals of amplitude smaller than 100 dB SPL. In practice, the output level at the compression threshold would be adjusted on an individual basis to be somewhat less than the discomfort level. Nominal characteristics of a compressor-limiter are given in the first row of Table 1. Limiters with these characteristics are commonly used in broadcast and recording studios (Shorter et al., 1967; Blesser, 1969; Noble and Bird, 1969). An extensive set of comparisons between clipping and compression techniques for protection against intense sounds is available in the literature, some of which is reviewed below. The characteristics of the compression-limiters included in this review are given in rows 2-5 of Table 1.

Davis et al. (1947), as part of the "Harvard Master Hearing Aid" study, compared clipping to compression-limiting for impaired listeners (sensorineural, conductive, and mixed) by measuring recognition scores for monosyllabic words imbedded in a fixed carrier sentence. With clipping, scores generally decline after the input level is increased beyond the point at which peak levels exceed the clipping threshold, even when highpass filtering is used before clipping. With compression-limiting no such decline is observed; the scores remain essentially constant when the input level is raised above the compression threshold. Hudgins et al. (1948) demonstrated that a wearable hearing aid with compression-limiting could be built and that at high input levels this aid performed better than two commercial aids that simply saturated and thus clipped.

Silverman and Harrison (1951) described a compression-limiter that was used in group hearing aids at schools for the deaf. The limiter freed teachers from worrying about talking too loudly and removed the fear of sudden acoustic shock from the children. A. W. de Vos (1969) commented favorably on a compression-limiter in a group aid that had been used in a school for the deaf since 1954.

Since 1951 a number of researchers have studied the advantages of compression-limiting in commercial hearing aids (e.g., Fournier, 1951; Pestalozza, 1953; Portmann and Portmann, 1961; Bizaquet and Viet, 1968). Recently, Blegvad (1974) compared a behind-the-ear aid with compression-limiting to ordinary behind-the-ear aids. Of 42 patients with sensorineural losses who used both types of aids for a period of two months, only 13 preferred the aid with compression-limiting. Similar results were obtained by Ludvigsen and Nielsen (1975) for severely impaired adults and by Brink et al. (1975) for profoundly impaired children.

The above recent research reported in the clinical literature has demonstrated no significant advantages for the compression-limiting available in commercial hearing aids. Some of the inadequacies of these aids are illustrated by measurements made at the National Bureau of Standards and reported by Burnett and Bassin (1976) and Burnett and Schweitzer (1977). The characteristics of 81 compression hearing aids were measured using standardized procedures (ANSI, 1976). With volume control on maximum, some of the aids had compression thresholds that were as low as 55 dB SPL in free field. These thresolds are unnecessarily low for most persons with hearing loss and would cause limiting of speech at normal conversational levels, as well as excessive amplification of background noise. Also, some aids produced 112-148 dB SPL output with 80 dB SPL input. Many of these output levels exceed the 100-110 dB SPL limit suggested by McCandless (1973) and by the work of Hood and Poole (1966). For these aids, compression characteristics were probably ineffective in preventing discomfort or pain from intense sounds. In addition, attack times on all aids ranged from 1 to 20 msec and release times from under 10 to more than 500 msec. Very few aids had attack/release times similar to those suggested by Davis et al. (1947) in the Harvard "master aid" study. Finally, some of the aids exhibited unnecessary ringing during output overshoot and undershoot which was similar to that noted by Nabelek (1973). In general, the negative clinical results may be explained by the wide range of characteristics noted in these measurements, by confusion about how these characteristics should be adjusted and the related absence of individual fitting, and by unnecessary distortions introduced by some of the aids.

3. Characteristics of Automatic Volume Control

Automatic volume control (AVC), which is also referred to as automatic gain control (AGC), adjusts the gain as a function of the long-term average speech input level. An AVC compressor acts very slowly and can thus generally be considered to be a linear amplifier in terms of its effect on the short-term level variations of speech. A compression curve for an AVC compressor is presented in Fig. 7. This curve indicates that sounds in the 60-100 dB SPL range would be presented over the 60-65 dB SPL range, and that sounds below 60 dB SPL would be amplified linearly with unity gain. In practice, the gain of an AVC compressor would be adjusted on an individual basis so that signals were presented at levels that led to good speech reception and that were comfortable over the long term. Nominal characteristics of an AVC compressor are given in the first row of Table 2. AVC compressors with these characteristics have long been used in broadcast and recording studios, in portable tape recorders, and as elements of speech processing systems (e.g., Kaiser and Bauer, 1962; Hellwarth and Jones, 1967; Torick et.al., 1968; Blesser, 1969; Bevan, 1973). Also, hearing aids with these characteristics (specifically TR 150 msec) have been designated AVC aids by the Swedish Medical Board (Johansson and Lindblad, 1971). AVC is used in hearing aids to keep the long-term output level near that corresponding to maximum intelligibility while the input level varies (e.g., Johansen, 1973). This type of processing might be particularly useful for persons with sensorineural losses who exhibit highly peaked articulation functions (Davis et.al., 1947; Huizing and Rejentes, 1952). Characteristics of some of the AVC compressors that have been studied for possible use with impaired listeners are given in rows 2 and 3 of Table 2.

According to Poliakoff (1950), the first use of AVC was made in a nonwearable aid in 1936. The need for such level control was indicated by a study which showed that out of 500 patients fitted with hearing aids about 50% had a tolerance range for speech of less than 26 dB. Aspinall (1951) described AVC which, together with peak clipping, was incorporated in the "British Master Hearing Aid" to accommodate the reduced tolerance range of listeners with recruitment. Although he suggested that AVC be evaluated with sentences, no results of such tests were given. AVC in conjunction with compression-limiting was incorporated in a group hearing aid designed for use in schools for the deaf (Silverman, 1949), but no evaluation of the effectiveness of AVC in this aid are available.

Flemming and Rice (1969) studied a compressor whose dynamic characteristics were appropriate for AVC, but whose static characteristics were designed to match the dynamic range of speech to the residual hearing of impaired listeners (and hence would probably be more appropriate for syllabic compression). On the basis of a preliminary investigation, involving both normal and impaired listeners, they reported that their compression system was marginally beneficial, but that there was no relation between the optimum compression ratio and the amount of recruitment.

Although there are presently a number of commercial hearing aids that incorporate AVC (e.g., Burnett and Schweitzer, 1977), no study has yet demonstrated clear cut advantages of AVC for impaired listeners.

4. Characteristics of Syllabic Compression

As noted above, we use the term syllabic compression to refer to nonlinear amplitude processing designed to increase speech intelligibility by altering the short-term intensity relations among speech elements. The characteristics of syllabic compressors are often chosen to match the dynamic range of speech to the residual dynamic range of impaired listeners in an attempt to compensate for abnormal loudness function. In general, such processing would tend to restore audibility to low-level speech elements without allowing high-level elements to become abnormally loud. Even if normal loudness relations were restored by syllabic compression, however, normal hearing would not be restored. Not only do sensorineural impairments often result in hearing anomalies other than those associated with loudness (e.g., abnormally poor frequency resolution, increased spread of masking, tinnitus, etc.), but amplitude compression, by its very nature, degrades a listener's effective intensity resolution. All of these factors must be taken into account in the design of syllabic compression systems.

Some insight into the effects of syllabic compression is available in studies which relate speech intelligibility to the consonant-to-vowel ratio (CVR), the ratio of acoustical power in a consonant to that in an adjacent vowel (Fairbanks and Miron, 1957; House et al., 1965; Williams et al., 1966; Salmon, 1970; Hecker, 1974). For syllables in which the consonant is /s/, CVR's ranging from -18 to -9 dB are typically observed, depending on talker and vocal effort. The range of CVR's for other consonants is smaller, but the variation with talker and vocal effort is similar to that for /s/. All studies in which talker or vocal effort were varied have shown that the average CVR is significantly correlated with intelligibility for words presented at equivalent peak levels in a background of additive noise to normal listeners. For example, variation in vocal effort which causes the CVR for /s/ to increase by 6-9 dB is accompanied by an increase of as much as 17 points in the Modified Rhyme Test score (Williams et al., 1966). Similar results for the consonants /sh, dz, t, tch, f, r, w, 1/ but not /b/, were obtained by Hecker (1974) who used computer processing to vary the CVR.

A single-channel compression amplifier with appropriate characteristics would achieve increases in the CVR that are roughly similar to those observed in the above studies. Such compression has been shown to improve the intelligibility of speech presented in noise to normals (e.g., Kretsinger and Young, 1960), and this suggests that impaired listeners may benefit from such processing, since additive noise simulates many characteristics of recruitment (Stevens, 1966; Richards, 1973).

Theoretically, the benefits of single-channel syllabic compression are likely to be quite limited, however, because the compression curve cannot be varied as a function of frequency and thus compressor action cannot reflect variations in hearing loss with frequency, or changes in the spectral characteristics of the input signal. For example, a single-channel system would not appear to be well matched to listeners with sloping audiograms who exhibit severe recruitment at high frequencies, but normal loudness function at low frequencies. Similarly, a single-channel syllabic compressor is likely to amplify the frication noise of /z/ or /v/ insufficiently because voicing energy controls compressor action for these sounds. To overcome these limitations, a number of investigators have proposed using multichannel syllabic compression systems in which the channels process separate frequency bands and the compression characteristics can thus be made frequency dependent. A simple multichannel system in which each channel compresses a distinct band of frequencies could be used to compensate for reduced dynamic range and recruitment. A more elaborate multichannel system, in which the compressor action for a given band of frequencies is partially controlled by signal components in other bands, might also be capable of reducing the spread of masking in frequency and time.

Little is presently known about the appropriate choice of static characteristics for a syllabic compressor. Theoretical arguments based on the properties of speech, the hearing loss, and the environment are still largely in formative stages. By contrast, certain of the dynamic characteristics of syllabic compressors are strongly constrained if the compressor is both to control the level of short speech elements and to introduce minimal distortions in the processed materials. As discussed in Sec. B-1, the attack time should be less than roughly 1 msec and the release time as short as possible, namely 20 msec. Not surprisingly, these choices are similar to those suggested more than 30 years ago for use in the speech spectrograph (Dudley and Gruenz, 1946; Steinberg and French, 1946). The same constraints could be applied to multichannel systems, although in such systems the attack time can sometimes be lengthened and the release time correspondingly shortened because filtering reduces rapid onset and offset rates.

C. REVIEW OF SYLLABIC COMPRESSION

1. General Comments

In general, studies of syllabic compression have focused on the problem of improving the intelligibility of words or syllables for persons with sensorineural impairments. Very little work has been concerned either with the "quality" of compressed speech or with an analysis of the perceptual confusions that occur with compression. Despite the effort that has been expended, it is still unclear whether syllabic compression is useful in improving speech reception for impaired listeners. In addition to the problems that have been discussed in connection with research on linear amplification (e.g., specification of functional gain, inadequacies of speech tests, limited characterization of impairments, etc.), research on syllabic compression has generally been limited by three major problems.

First, syllabic compression systems are usually much more complex than linear amplification systems. As indicated above, the operation of even the simplest compression amplifier depends on the settings of such parameters as attack and release times, compression ratio, compression threshold, compression range, and frequency-gain characteristic. In addition, internal distortion and noise are likely to play a prominent role in compression systems. Because of this complexity (which, of course, increases as the number of channels increases), and because standardization of compression parameters has only recently begun to receive attention, the problem of adequately specifying the systems studied is a difficult one. In most research on compression, the systems considered have not been described adequately. In addition, the complexity of these systems has made it difficult to perform experiments in which the parameters of the system are varied systematically and in which interaction effects are studied.

Second, most studies that have compared syllabic compression to linear amplification have failed to give adequate consideration to the choice of frequency gain characteristic of the linear system. Not only have relevant acoustic effects been ignored in specifying these characteristics, but the characteristics have not been well matched to the listeners used. In most cases, the characteristic has been flat, or nominally flat, independent of the properties of the hearing loss (e.g., the shape of the audiogram). As indicated in Braida et al. (1978), a flat characteristic is far from optimum for a large fraction of listeners with sensorineural impairments. On the whole, this failure to choose an appropriate frequency-gain characteristic for the linear system tends to exaggerate the benefits associated with amplitude compression. On the other hand, it should also be noted that in most studies very little consideration has been given to the frequency-gain characteristic of he compression system. To what extent these two deficiencies tend to cancel (so that the reported relative performance of compression and linear amplification is roughly correct) is unknown.

Third, there has been inadequate consideration of the selection and variation of speech levels used in the study of syllabic compression. One aspect of this problem concerns the long-term level of the speech material. This parameter is particularly critical for studies of amplitude compression because the signal transformation produced by compression depends on level. Also the effects of varying the input level differ from those of varying the output level (even when the range of output level variation is the same). In addition, when compression is compared to linear amplification, the effect of varying input level is likely to be stronger for the linear system because the gain in the compression system is greater for low-level signals than for high-level signals. Throughout our review of syllabic compression, we tend to focus on intelligibility scores maximized over level rather than on performance at presumably comparable levels. A second aspect of this level problem concerns the variation of levels that occurs within items of a test presented at a given overall level. Within the context of the discussion of different types of compression presented in Sec. B and the scheme shown in Fig. 1, an appropriate test of syllabic compression would incorporate syllable-to-syllable level variations comparable to those encountered in everyday speech, but would employ a fixed, long-term, overall level (theoretically achieved through the use of AVC prior to compression). Unfortunately most of the tests that have been conducted do not meet this criterion. In particular, the natural intersyllable level variation that occurs in words and phases has often been reduced by artificially preprocessing the test materials (equating levels). In general, it is not possible to estimate the effects of this preprocessing on the intelligibility scores obtained. It seems obvious however, that this preprocessing reduces the measured effectiveness of syllabic compression relative to that of linear amplification.

The characteristics of the systems considered in this review of syllabic compression (called compression in this section) are given in Table 3.

2. Detailed Reviews

Edgardh (1952) was among the first to propose using syllabic compression to increase speech intelligibility for impaired listeners. He suggested using a modified form of compression limiting (high compression ratio, low compression threshold, and short release time) to equalize the levels of vowels and consonants for persons with reduced dynamic ranges. Although he does not appear to have evaluated this type of "extreme limitation" with impaired listeners, he reported that processed speech exhibited a certain sibilance, but no "distortion of speech as to affect adversely its comprehension could be observed - either in male or female voices." He further noted, however, that "each breath taken by the talker was amplified to a loud gasp", as would be expected for a system with a high compression ratio and a low compression threshold. Although the high compression ratio (CR = 7) specified by Edgardh has not been recommended by more recent studies, it is noteworthy that Edgardh's suggestions for attack and release times (1-2 msec and 20 msec, respectively) have often been adopted by later investigators.

Parker (1953) experimented with a modified commercial limiter as a means to reduce "short time fatigue" in impaired ears caused by high level sounds. (Results of Parker's study that relate to the effect of highpass filtering are discussed in Chapter II.) He measured the intelligibility of both linearly amplified (nominally flat gain) and compressed PB-50 word lists (Egan, 1948) at presentation levels varying from 6 to 36 dB above the speech reception threshold for linearly amplified spondees. Of the 10 subjects with sensorineural losses studied (average loss roughly 50, 55, and 60 dB at 0.5, 1 and 2 kHz), only 4 showed an increase in intelligibility with compression (when scores are maximized over level). Averaged over these 4 subjects, the maximum score obtained with compressed speech exceeded the maximum score obtained with linearly amplified speech by roughly 22 points. However, 8 of the 10 subjects (including the above 4) also showed an average improvement of roughly 15 points when a sharp highpass filter at 0.67 kHz was applied to the linearly amplified speech (at the input level producing maximum score for the unfiltered speech). Only 2 of Parker's subjects were benefitted more by compression than by filtering. In interpreting these results it should be noted that the compression system caused increased distortion and noise. It had a high compression ratio and a low compression threshold, "amplified (internal noise during silent periods) to a loudness level that ultimately became almost as high as the speech signal", and had a release time that was short enough to have distorted speech elements with low frequency components. Also, the above results refer only to scores maximized over level; at low levels, most of the subjects achieved substantially higher scores with compression than with linear amplification.

Kretsinger and Young (1960) used a compressor described by Daniel (1957) and demonstrated that persons with normal hearing listening in noise could achieve improved word intelligibility scores with compression. They compressed over input ranges of 10 and 20 dB, "added white noise 3 dB below speech level" to the compressed signals, and presented W-22 lists (Hirsh et al., 1952) at 70 dB SPL to 30 normal hearing subjects. On the average, scores increased from 57% for linear amplification (nominally flat gain) to 85% for 10 dB of compression and 78% for 20 dB of compression. For speech processed by "10 and 20 dB of instantaneous clipping", intelligibility increased to only 63%

and 61%.

Lynn and Carhart (1963) investigated the effect of various attack and release times using a compressor "constructed from hearing-aid components". They utilized 3 groups of listeners (10 per group) with losses that were attributed to otosclerosis, labyrinthine hydrops, and presbycusis. The loss for each group (averaged over the frequencies 0.5, 1, and 2 kHz and over the members of the group) was roughly 35-40 dB. The average audiogram for the otosclerotic group was flat; for the other groups it fell roughly 10 dB per octave above approximately 0.2-0.5 kHz. The investigators first measured the speech reception threshold (SRT) of 30 hearing impaired subjects using processed isolated spondees and found that the SRT decreased by about 10 dB as TA increased from 5 msec to 85 msec, independent of TR. They then presented paired PB-50 words in carrier sentences (e.g., "Please repeat STRIFE BAIT.") 25 dB above the SRT. They found little change in scores for TA/TR ranging from 6/30 to 20/500 msec and generally little improvement in intelligibility for compressed materials. Average discrimination score was constant to within 4 points except for the two extreme values of TA/TR (70/400 and 85/1200 msec) where it decreased by 4 and 12 points. The maximum increase in average score (obtained by first averaging over members of a group and then maximizing over TA/TR) above linear amplification (nominally flat gain) was roughly 9 points (82% vs 73%) for the presbycusis group, 8 points (80% vs 72%) for the hydrops group, and 2 points (93% vs 91%) for the otosclerosis group.

As noted by the authors, the reduction in SRT with increasing TA would be expected for isolated spondees. When TA is large, the beginning of the spondee was amplified with maximum gain to a level that was roughly 45 dB above SRT (because of the initial overshoot pictured in Fig. 4). Under this condition, the measured value of SRT would clearly be lower than for short attack times. Since the PB-50 words were presented 25 dB above the measured SRT, the decrease in scores at long TA/TR times may simply reflect reduced presentation level. (Intelligibility scores for monosyllables often increase 2-4% per dB in the region 10-20 dB above SRT). Also, extreme compression was used and, although distortion was not specified, the compressor used is likely to have been subject to distortions characteristic of commercial aids (e.g., Nabelek, 1973). Finally, the 10 subjects with otosclerosis achieved high word scores with linear amplification (greater than 90% on the average) and, to the extent that their losses were purely conductive, would probably not have had a reduced dynamic range requiring compression.

Caraway (1964) and Caraway and Carhart (1967) used a three-channel instantaneous compressor with square- or cube-rooting. Although the signals in the three bands (0.2-1 kHz, 1-2 kHz, and 2-5 kHz) were processed independently, the same compression ratio (2 or 3) was used for all bands. Compression was compared to linear amplification (nominally flat gain) for CNC words (N.U. Auditory Test #4; Tillman et al., 1963) in a carrier sentence with words equated for peak power after processing and presented at sensation levels in the range 0-24 dB SL. Scores for normal listeners were essentially perfect for all systems at 24 dB SL. Scores for impaired listeners (labryinthine otosclerosis, labryinthine hydrops, and presbycusis; average audiograms similar to those in Lynn and Carhart, 1963) showed only slight improvement over linear amplification. More specifically, the maximum increase in average score (obtained by first averaging over members of a group and then maximizing over presentation level) was less than or equal to 4 points for all patient groups. Also, the advantage of compression at reduced presentation levels was only slightly greater than at higher levels. It should be noted, however, that many of the impaired subjects were able to achieve very high scores with linear amplification: more than a third of the 36 impaired listeners studied achieved scores in the range 90-98% at 24 dB SL, and the average score for all impaired listeners at this level was roughly 80%. For these listeners and for the speech tests used, it would be difficult to demonstrate an improvement in intelligibility with any kind of speech processing. The performance of some of the listeners may also have been limited by certain characteristics of the compression system. For example, the compression ratio was constant over the frequency range of the system rather than matched to the reduction in dynamic range. Also, high levels of harmonic distortion occurred at all input levels and was extreme at low frequencies. Intermodulation distortion, which may have been even larger, was not measured. Similar results for impaired listeners have been reported by Trinder (1972) who used wide-band instantaneous square-, cube-, and Nth-rooting. He found, however, that this processing severely reduced speech intelligibility for normals listening in quiet.

Burchfield (1971) modified the compressor used by Caraway and Carhart to reduce harmonic distortion by filtering. The filtering employed, however, could not eliminate in-band intermodulation distortion. This modified compressor was compared to linear amplification (nominally flat gain) for 36 listeners with sensorineural losses accompanied by recruitment (flat or gently sloping audiograms; loss roughly 50 dB averaged over subjects and frequencies 0.5, 1, and 2 kHz). Intelligibility was measured using CNC words (N.U. Auditory Text #6; Tillman and Carhart, 1966) in a carrier sentence presented at a peak level that was 24 dB above each subject's SRT measured with linear amplification. The average increase in intelligibility obtained with compression over that obtained with linear amplification was roughly 12 points (63% to 74%) for both CR=2 and CR=3.

Vargo (1977) reported on a series of experiments designed to resolve the difference between the findings of Caraway and Carhart (1967) and the more positive findings of Burchfield (1971). However, he used a single-channel compressor whereas the previous investigators had used three-band instantaneous distortion systems. He compared a linear system (nominally flat gain) to the compression system adjusted to have CR values of 2 and 5 using CNC words (Peterson and Lehiste, 1962) equated for peak level after processing. All materials were presented at 10, 20, and 30 dB SL to 12 normal hearing subjects and to 9 subjects with Meniere's disease and relatively flat losses (loss approximately 50 dB averaged over 0.5, 1 and 2 kHz, and over subjects). The differences in mean scores obtained with the 3 systems (linear, CR=2, CR=5) did not exceed 3 points for either group of listeners at any presentation level. This result was interpreted as supporting the results of Caraway and Carhart.

Johansson and Lindblad (1971) and Johansson (1973) suggested the values TA = 2 msec, TR = 20 msec, and CR = 6 under quiet listening conditions for subjects with narrow dynamic ranges. Also, based on experience in Swedish schools for the hard of hearing, they suggested using CR 5 in the presence of classroom noise. In addition, Johansson (1973) reported the results of experiments that investigated the effect of varying the attack and release times in a single-channel compressor on the intelligibility of CVC nonsense syllables. In this test, TR = 200 TA, and TR was varied from 10 to 1000 msec. Ten normal subjects were tested with flat noise added after processing (S/N = 5 dB) and 12 impaired subjects were tested in silence. For both groups, vowel intelligibility was independent of TR, but final-consonant intelligibility fell when TR was greater than roughly 150 msec. For the normals, a release time of 1 sec increased confusions for the weaker voiceless consonants and /m/.

Thomas and Sparks (1971) compared the intelligibility of PB-50 words that were highpass filtered (12 dB per octave at 1.1 kHz) and infinitely clipped with that of speech that was linearly amplified (nominally flat gain) and had the same average rms level (10, 20, 30, and 40 dB SL). They tested 16 impaired listeners with a variety of etiologies and hearing losses (the loss of individual subjects, averaged over 0.5, 1, and 2 kHz, ranged from 15 to 75 dB).

Maximum scores with both types of processing were obtained at the highest levels tested and showed little advantage for the filtered-clipped speech (less than 5 points averaged over subjects). A substantial advantage, however, was obtained at the lower levels.

Ruhrberg and Esser (1973) evaluated the intelligibility of both compressed and linearly anplified (nominally flat gain) monosyllables presented at equivalent peak levels to listeners with cochlear and retrocochlear impairments. For the listeners with moderate cochlear impairments, scores with compression averaged 10 points higher; for severe cochlear impairments, the improvement was 27 points. Listeners with retrocochlear losses achieved smaller improvements: 5 and 17 points, respectively.

Robinson and Huntington (1973) evaluated a modification of the compressor shown in Fig. 2 in which a time delay is introduced between the input and the variable-gain linear amplifier. This modified system is similar to a compression-limiter described by Shorter et al. (1967) and was utilized in an attempt to obtain short (several milliseconds or less) time constants. Although this delay tends to compensate for the time required for the level detector to sense changes in the input signal, and to reduce the overshoot and undershoot in the output envelope (see Fig. 8), it does not (as suggested by the authors) circumvent the limitations on attack and release time discussed in Sec. B-1 of this Chapter. Nevertheless, the basic idea of introducing a delay to permit the compression system effectively to look ahead in time is important and deserves serious consideration. Robinson and Huntington measured intelligibility using monosyllabic word lists in quiet and noise with both normal and hearing impaired subjects. Their results indicated that under some conditions the intelligibility of compressed speech is superior to that of linearly amplified speech. Unfortunately, no detailed results of this study have been made available.

Gregory and Drysdale (1976) employed "high frequency carrier clipping" (HFCC) to achieve amplitude compression with short attack and release times. In this scheme a high frequency carrier is amplitude modulated by speech, clipped, and filtered to remove distortion components. The resulting signal is then heterodyned back to the original audio frequencies. HFCC was compared to a system with instantaneous peak clipping and to a linear system (nominally flat gain). The bandwidths of all systems were restricted to 400-2500 Hz and 16 dB of clipping was used for both the HFCC system and the instantaneous clipper. This 16 dB of clipping was obtained by first measuring the output peak speech level without clipping, increasing the input level by 16 dB, and then adjusting the clipping to return the output to the original level. Words and sentences were presented to children and adults with sensorineural losses in quiet and also to normals in broadband noise (S/N = 20-30 dB). It was found that 7 of the 16 impaired listeners tested with words and 4 of the 7 impaired listeners tested with sentences had scores with HFCC that were more than 10 points higher than scores obtained with linear amplification (roughly 50%). Each of the 12 normals tested with words and 7 of the 8 normals tested with sentences demonstrated increased scores with HFCC. Also, peak clipping was less effective than HFCC.

Yanick (1973), in a study involving 12 listeners with mild to moderate sensorineural losses, compared a custom-fit, wearable, single-channel compression aid having a low compression threshold to each subject's own aid in free field. The gain and compression ratio of the experimental aid was adjusted for each subject on the basis of the SRT and the speech discomfort level. As would be expected, compression significantly increased the range between each subject's SRT and discomfort level for speech. It also increased intelligibility (measured with PB-50 word lists) at input levels of 45 and 70 dB SPL. At the higher level the inrease was dramatic: 91% for compression compared to 39% for the subject's own aid. The author attributed the advantage of compression to overload of the subject's own aid at the high input level. This work demonstrates that compression can be built into a wearable aid without sacrificing characteristics important for speech perception.

Yanick (1975) also compared a single-channel, wearable, compression aid having very short time constants to a second wearable aid with compression-limiting. Six subjects with mild to moderate sensorineural losses having flat or gently sloping audiograms received intelligibility tests (PB-50 words) both in quiet and in the presence of speech babble. When the input to the aid with compression-limiting was adjusted such that the compression threshold was not exceeded and the aid was thus linear, the output levels of the two aids were 85 and 90 dB SPL, respectively. Under this condition, the average score for compression-limiting minus the average score for compression was -2, 4, 14, and 0 points at S/N = 0, 10, 20 dB and in quiet. Thus there was little or no advantage demonstrated for compression.

Nabelek and Robinette (1975) evaluated the effect of distortions that occur during compression overshoot in commercial hearing aids on Modified Rhyme Test scores (Kreul et al., 1968). They tested 7 aids using 10 normals listening in noise and and 10 listeners with sensorineural impairments (loss approximately 50 dB averaged over 0.5, 1, and 2 kHz and over subjects). In general, poorer scores were obtained with the aids that exhibited greater compression related distortion. Also, the two aids with the shortest attack/release times (6/30 and 17/50 msec) always ranked among the top three in terms of MRT scores. As pointed out by the authors, these results are preliminary: only a limited sample of aids was tested and many characteristics of the aids were not equated or varied independently.

One of the most encouraging studies of syllabic compression was made by Villchur (1973) using high quality commercial (DBX) compressors and specially designed and properly calibrated circumaural earphones. He designed his compression system to "restore normal loudness to each acoustical speech element of importance," noting that this might require many frequency bands, each with its own compressor. At least two bands are necessary because "without multiband compression, only the amplitude ratio between successive speech elements can be changed, and not that between elements that occur simultaneously." His system utilized two compression bands with an adjustable crossover frequency and a 1/3-octave filter bank for post-compression frequency equalization. This design permitted independent adjustment of the compression ratio in the two bands as well as the frequency-gain characteristic of the system as a whole.

Villchur fi st tested his system with two normal-hearing subjects and a shaped noise background (to simulate hearing loss) that raised thresholds by about 70 dB at 0.5 kHz, 75 dB at 1 kHz, and 90 dB at 2 kHz. Intelligibility was measured using CVC nonsense syllables imbedded in sentences. Materials were recorded by a female speaker in a slightly reverberant environment. Compression ratios were determined from the reduction in dynamic range caused by the noise and equalization was chosen to place the compressed speech at a level within the reduced dynamic range equivalent to the corresponding level in the normal case. Presentation levels were chosen by each subject "on the basis of maximum clarity consistent with long-range comfort." The results show that scores for final consonants increased from roughly 38% with functionally flat linear amplification to 84% with compression and equalization. The average score was 58% with compression alone and 51% with equalization alone. Thus, although compression alone provided some increase in intelligibility, compression combined with equalization provided a much larger increase.

Villchur also measured intelligibility for the same speech material using 6 subjects with moderate to severe sensorineural hearing losses (average loss roughly 41, 47, and 67 dB at 0.5, 1, and 2 kHz). Initially he calculated the equalization and compression needed to restore normal loudness relationships for speech on the basis of pure tone threshold, discomfort level, and equal-loudness measurements, as well as the characteristics of speech. On the basis of this calculation, he chose the crossover frequency defining the two bands (this frequency ranged from 1.3-2.5 kHz for different subjects), the compression ratios in the two bands, and the post-compression equalization. Then he allowed subjects to vary frequency equalization and compression ratios from their calculated values while listening to continuous speech to achieve maximum intelligibility consistent with long-range comfort. The adjusted values of the compression ratios were usually near the calculated values: the average values of CR in the low and high bands were 2.2 and 3.5 before adjustment, and 2.1 and 2.8 after adjustment, respectively. The compression systems resulting from these procedures were compared to a linear system that had a functionally flat frequency-gain characteristic except for subjectselected low-frequency rolloff. Averaged over subjects, the linear lystem had a functional gain that was roughly flat above 500 Hz, but fell at roughly 15 dB/octave below 500 Hz. Speech testing occurred both in quiet and with a competing voice 10 dB below the test words. The speech materials were presented at the most comfortable level (MCL), and with the speech level at the input to the compressor reduced 10 and 20 dB. The results at MCL in quiet show an improvement in terminal consonant scores ranging from 5 points (40% to 45%) to 40 points (25% to 65%)

with an average of 21 points, and an improvement in initial consonant recognition ranging from 3 points (60% to 63%) to 22 points (44% to 66%) with an average of 10 points. The high intelligibility of vowels obtained with the linear system at MCL was either maintained or slightly increased by compression. Furthermore, "almost all of the reduced-input scores showed increased benefit from processing, and the improvement from processing was usually maintained or increased in the interference tests." Two subjects were also tested with Harvard sentences (IEEE, 1969). With compression, key word scores increased from 28% to 48% for one subject and from 73% to 89% for the other.

Villchur's results suggest that syllabic compression can be of considerable value. Those elements of his study that appear to contribute to these results include the use of multichannel compression combined with frequency equalization; the determination of compression and equalization parameters on the basis of each subject's hearing characteristics and individual adjustments; the use of equipment with low distortion and noise characteristics; and the use of subjects who potentially could derive large benefits from such processing. It should be noted, however, that the advantage of compression demonstrated in his study may have been artificially inflated by comparison to an inferior linear amplification system (i.e., a system with inadequate high-frequency emphasis).

Yanick (1976) performed a series of experiments that were similar in design to those of Villchur (1973) but differed in that a background of cafeteria noise was always present, lower S/N ratios were used, and both one-channel and two-channel systems were tested. Yanick's two-channel system was similar to Villchur's. It had a crossover frequency of 1.5 kHz and was individually fitted by allowing subjects to adjust high- and low-frequency channel compression ratios, the gain of the high- relative to the low-frequency channel, and "treble" emphasis and "bass" rolloff (highpass filtering with a slope of 0, 6, 12, or 18 dB/octave below 1.5 kHz). The resulting system was compared to a linear system that was electrically flat above 1.5 kHz, modified by subject-adjusted bass rolloff similar to that used with the compression system. All adjustments were made while listening to a sentence on a tape loop, and all materials were presented through an insert hearing-aid receiver at the level thought (by the subject) to provide maximum intelligibility. Systems were compared using Harvard sentences that had been recorded by a male talker in a reverberant environment and presented at S/N ratios of 0 and 6 dB. All subjects had moderate to severe sensorineural impairments (losses of 40-70 DB in the "speech frequencies"). One group of 12 subjects had flat audiograms (from 0.5 to 4 kHz); a second group of 12 had sloping audiograms (12-25 dB/octave above 0.75 kHz).

The main experiment compared the linear system to the two-channel compression system. The two-channel compression system had average low- and high-frequency CR values of 2.4 and 2.8 for the flat group and 1.5 and 2.5 for the sloping group, and an average bass rolloff of 6 dB/octave for the flat group and 12 dB/octave for the sloping group. The key word scores showed large gains for the two-channel compression system for both subject groups and both S/N ratios: 37 points (54% - 91%) for the flat group and 28 points (56% - 84%) for the sloping group at S/N = 6 dB; 34 points (38% - 72%) for the flat group and 38 points (29% - 67%) for the sloping group at S/N = 0 dB. Further experiments, performed using 6 subjects from each of the two groups, demonstrated that performance with each subject's own hearing aid was roughly comparable to that with the linear amplification system. They also showed that a single channel compression system with an average CR value of 1.8 performed significantly worse than the two-channel system. Scores, averaged over all subjects, for linear amplification, single-channel compression, and two-channel compression were 61%, 65%, and 91% at S/N = 6 dB and 43%, 30%, and 78% at S/N = 0 dB.

There are two major aspects of Yanick's study that should be noted in interpreting his results. The first concerns the frequency-gain characteristics of the systems studied, Although the subjects were allowed to adjust many characteristics of the equalization provided in the two-channel compression system, they were only allowed to adjust the bass rolloff of the linear system. These procedures, combined with the use of an insert receiver (which tends to reduce the gain at high frequencies), may have artificially inflated the scores for the two channel compression system relative to those for the linear system. The second element of Yanick's study that requires comment concerns the use of low S/N ratios for a noise (cafeteria noise) whose low-freuency components are likely to predominate, Under such conditions the gain of the single-channel system and of the low-frequency channel of the two-channel system may be controlled primarily by the background noise rather than by the speech signal, The single-channel system would then operate as a linear system with roughly constant gain, and the two-channel system would operate as a single-channel compression system with compression confined to the high-frequency channel, This could explain the poor performance of the single-channel system relative to the two-channel system,

Yanick and Drucker (1976) performed a second series of experiments that were similar to the previous experiments of Yanick (1976), except that the systems studied consisted of, in addition to the linear system and the two-channel system used previously, a second two-channel compression system. This second two-channel system differed from the first in that amplitude expansion (CR = 0.7) was included in the low-frequency channel below the compression threshold. The 6 listeners tested had mild to moderate sensorineural impairments (average loss of 25-60 dB at 0.5, 1, and 2 kHz) and sloping audiograms (12 - 25 dB/octave above 0.75 kHz). Key word scores on Harvard sentences for the linear, two-channel, and two-channel expansion systems were 46%, 77%, and 87% at S/N = 6 dB, and 22%, 61%, and 69% at S/N = 0 dB. These results thus appear to support the conjecture (e.g., Villchur, 1973) that expansion can reduce the effects of background noise. The increase in intelligibility with expansion is difficult to understand, however, given the low S/N ratios used in these experiments. More specifically, since the level of the noise in the low-frequency channel was only 0 to 6 dB below the peak speech level, and the compression threshold in this channel was 30 dB below the peak speech level, the level of the noise in the low-frequency channel should rarely have fallen below the compression threshold and expansion should have had little effect. Also, the results obtained using the unmodified two-channel system are not entirely consistent with those of the previous study. For example, a comparison of the scores obtained in the two studies with the unmodified two-channel system by subjects with sloping losses shows that scores were roughly 6 points lower in the current study despite the fact that the subjects in this study had less severe losses. Also, the scores obtained with expansion in the current study were, on the average, only 2 points greater than those obtained without expansion in the previous study.

Barfod (1976) explored the hypothesis that multichannel compression individually fitted to restore normal equal-loudness contours is superior to linear amplification. In a study involving 5 subjects with sensorineural losses, he compared each subject's own aid to a four-band linear system that was "optimally chosen and fitted" to each subject's loss, and to one-, two-, and three-channel compression systems designed to restore normal equal-loudness contours for tones. All subjects had bilateral losses, were hearing aid users, and had normal hearing at low frequencies (losses of 20 dB or less at 0.25 or 0.5 kHz) and significantly impaired hearing at high frequencies (losses greater than 50 dB above 2 kHz). The average audiogram for the 5 subjects showed losses of 22, 36, and 62 dB at 0.5, 1, and 2 kHz. Characteristics of the 4 experimental systems were based on equal-loudness contours that were related to normal equal-loudness contours via the normal low-frequency hearing of the impaired subjects. The optimal linear system (CH0) was determined (in an unexplained manner) on the basis of previous research (see discussion of Barfod, 1972, in Braida et al., 1978). According to our estimate, which attempts to take account of certain acoustic factors ignored by Barfod, the functional gain provided by this system was roughly flat above 750 Hz, but fell at roughly 18 dB/octave below 750 Hz. Each compression system had one linear low-frequency channel plus one (CH1), two (CH2), or three (CH3) high-frequency compression channels. The band limits, compression ratios, and relative gains of the channels were chosen to restore normal equal-loudness contours. Band limits (averaged over subjects) were roughly 0.9, 2.0 and 2.8 kHz for CH3, 0.9 and 2.0 kHz for CH2, and 1.2 kHz for CH1; and the CR values (averaged over subjects)in the various channels (in order of increasing frequency) were 1.0, 2.5, 4.8, and 5.6 for CH3, 1.0, 3.8, and 4.2 for CH2, and 1.0 and 4.7 for CH1. Since post-compression equalization was not used, gain adjustments could only be made between channels. The systems were compared both in quiet and with speech-spectrum noise added before processing at various S/N ratios, using nonsense CVC syllables spoken by a male talker. In all tests the input speech level was fixed at 65 dB SPL, and materials were presented using Beyer DT-48 headphones.

The relative performance of the 5 systems did not vary significantly with S/N ratio or subject (except for one subject whose results are excluded from the following averages). The scores, in terms of per cent phonemes correct averaged over the quiet condition and S/N ratios of -5, 0, 5, 10, and 15 dB, were 54% (own aid), 65% (CH0), 46% (CH1), 46% (CH2), and 64% (CH3). Thus the three-channel compression system, which came closest to restoring normal equal-loudness contours, performed neither better nor worse than the "optimal" linear system; however, both of these systems were much superior to the one- and two-channel compression systems and significantly better than the subject's own aid.

Barfod was the first investigator to study the effect of varying the number of compression bands systematically. However, the gain and compression ratio of each channel were not varied independently but rather co-varied to restore normal equal-loudness contours as well as possible for a given number of channels. Also, as the number of channels was reduced, the release times were increased (to avoid distortion at low-frequencies) and the attack times, which were constrained to equal the release times, were lengthened as well. It is not known whether his results would apply to systems in which these parameters were varied independently.

Barfod's results on compression are at variance with the more positive findings of Villchur (1973). Among the differences between the two studies that may account for this discrepancy are the following. First, Barfod's subjects on the average had sharply sloping losses; Villchur's subjects had more severe but gradually sloping losses. Second, Barfod's linear systems employed greater high-frequency emphasis than Villchur's. Third, Barfod's system was adjusted to restore normal equal-loudness contours for pure tones; Villchur's system included post-compression equalization and his procedures included individual modifications of system parameters. Fourth, Barfod's tests were performed at a fixed input level; Villchur's tests were performed at each subject's most comfortable level and 10 and 20 dB below this level. Fifth, Barfod's system had equal attack/release times that varied from 24 msec at low frequencies to 6 msec at high frequencies; Villchur's system had attack times of less than 1 msec and release times of roughly 20 msec in all channels.

D. CONCLUDING REMARKS

(1) Laboratory studies have demonstrated that compression-limiting preserves speech intelligibility at high levels better than simple peak clipping while providing the same protection from intense sounds. Recent clinical studies utilizing commercial hearing aids, however, have demonstrated no advantage for compression limiting. These negative results appear to be caused primarily by inappropriate compression characteristics in commercial aids and lack of individual fitting. Also, the aids compared have often differed in an uncontrolled manner with respect to a variety of electroacoustic properties other than the method used to limit intense sounds.

(2) Although the need for AVC in hearing aids has been recognized since 1950, there has been little serious research on this topic. There appear to be no technical obstacles to incorporating AVC in wearable hearing aids without compromising other essential features of the aids. Furthermore, the widespread use of AVC in sound recording applications suggests that, at least in certain highly-controlled circumstances, AVC can be used successfully without degrading quality or intelligibility. In order to determine the value of AVC for hearing aids, its effects on speech reception, comfort and annoyance must be studied

in controlled experiments with impaired listeners and in a wide variety of realistic environments. Testing AVC in realistic environments is particularly crucial because it is the function of AVC to control long-term levels, and constructing simulations of environments that preserve parameters relevant to the control of these levels is exceedingly difficult.

(3) Despite the numerous studies of syllabic compression, it is still impossible to state whether syllabic compression is beneficial for persons with sensorineural losses.

Much of the research performed prior to 1970 was hampered by technical limitations that impeded the use of low compression ratios without employing instantaneous distortion devices, by poor design that introduced ancillary distortion and noise, and by the selection of inappropriate time constants. Also, almost all of the systems utilized single-channel compression in which neither the compression ratio nor the gain varied with frequency. In the only study prior to 1970 that employed multichannel compression (which, theoretically, should be superior to single-channel compression) and low compression ratios, the system suffered from considerable distortion and the compression ratio and gain were not varied over frequency.

During the last decade, it has become possible to overcome essentially all the technical limitations that faced earlier investigators. Advances in electronics have made it possible to build low-distortion compressors with compression ratios that are both low and variable. These compressors can now be easily combined to create multichannel compression systems that have low distortion, that are frequency dependent (in both compression ratio and gain) and that employ appropriate time constants. Unfortunately, however, the research conducted during the past decade with these more advanced systems has still not led to a clear picture of the benefits that can be achieved with syllabic compression. The results that have been obtained are in many cases inconsistent, and the reasons for the discrepancies are not yet thoroughly understood. Clearly, further research on syllabic compression is required before it can be recommended for clinical use.

FIGURE CAPTIONS

Figure 1. Black-box diagram of hearing aid that incorporates three types of nonlinear amplitude processing.

Figure 2. Black-box diagram of compression amplifier.

Figure 3. Static compression characteristics.

Figure 4. Dynamic compression characteristics. The constants α and β are arbitrary.

Figure 5. Distortion characteristics of an instantaneous peak clipper. The constants α and β are arbitrary.

Figure 6. Compression curve of a compression-limiter.

Figure 7. Compression curve of an AVC compressor.

Figure 8. Dynamic characteristics of a compressor of the type suggested by Robinson and Huntington (1973) assuming a large compression ratio.

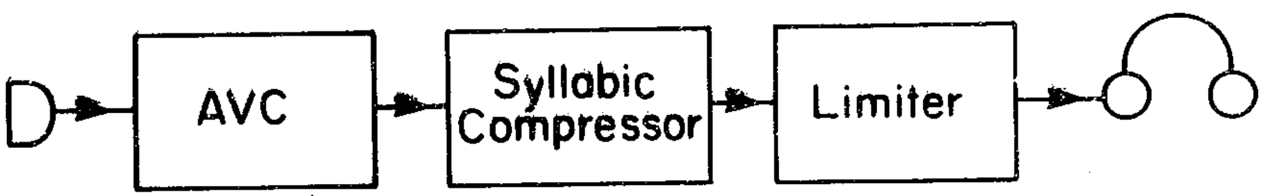


FIGURE 1.

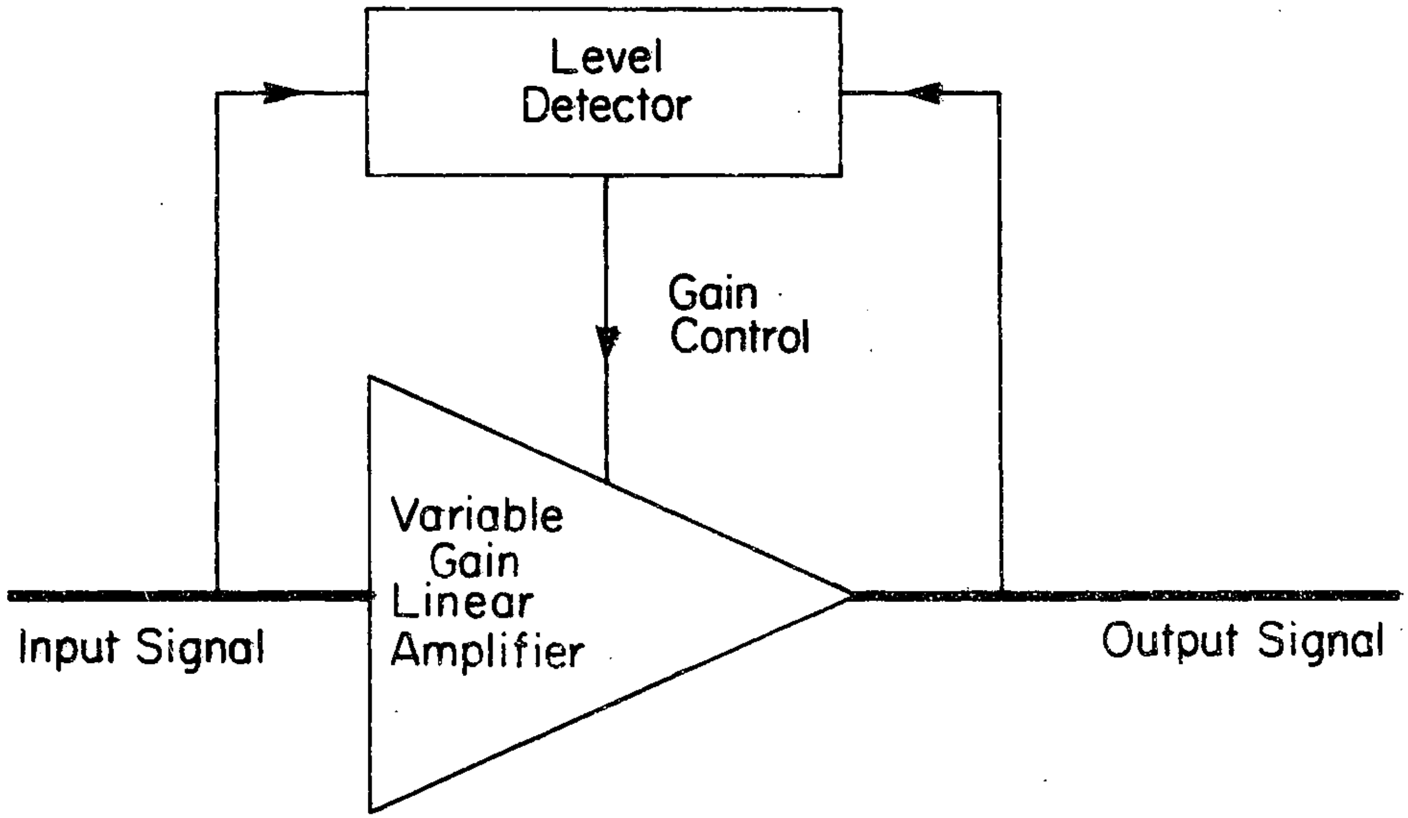

Level Detector
Gain Control
Variable Gain Linear Amplifier
Input Signal
Output Signal

FIGURE 2.

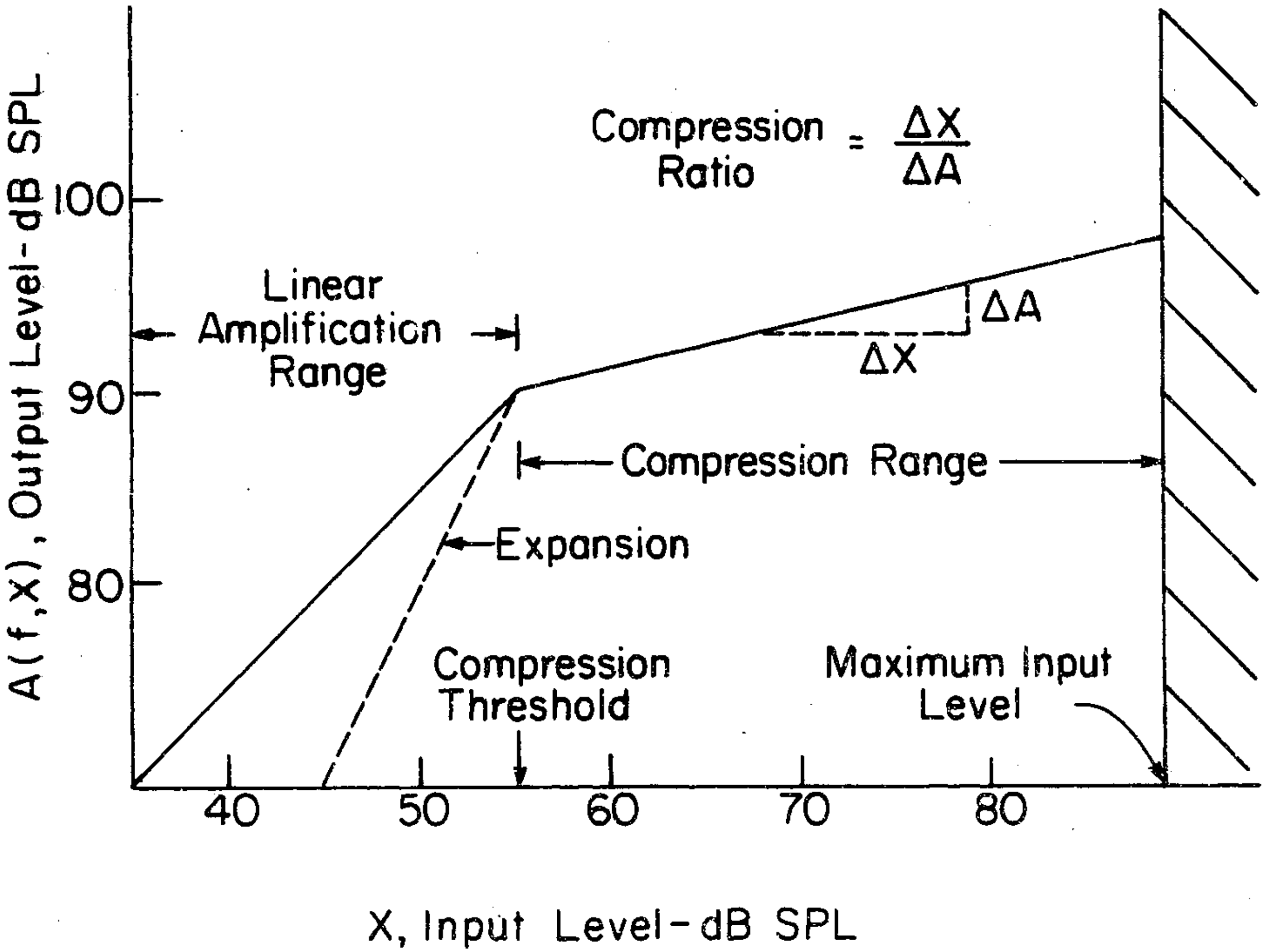
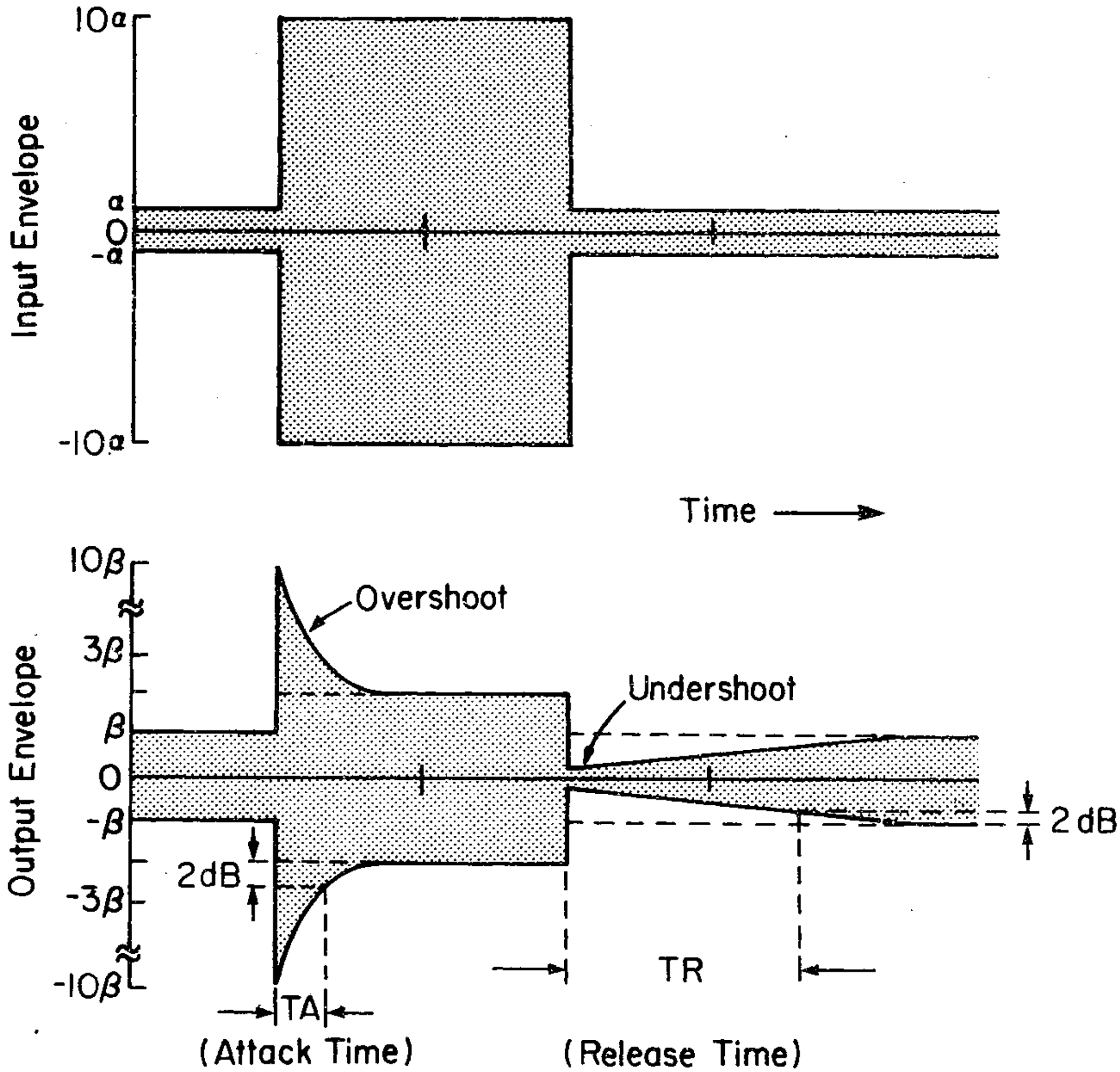

Compression Ratio = ΔX / ΔA
Linear Amplification Range
ΔA
ΔX
Compression Range
Expansion
Compression Threshold
Maximum Input Level
A(f,X), Output Level-dB SPL
100
90
80
40
50
60
70
80
X, Input Level-dB SPL

FIGURE 3.



10α
α
0
-α
-10α
Input Envelope
Time
10β
3β
β
0
-β
-3β
-10β
Output Envelope
Overshoot
Undershoot
2dB
2 dB
TA
(Attack Time)
TR
(Release Time)

FIGURE 4.

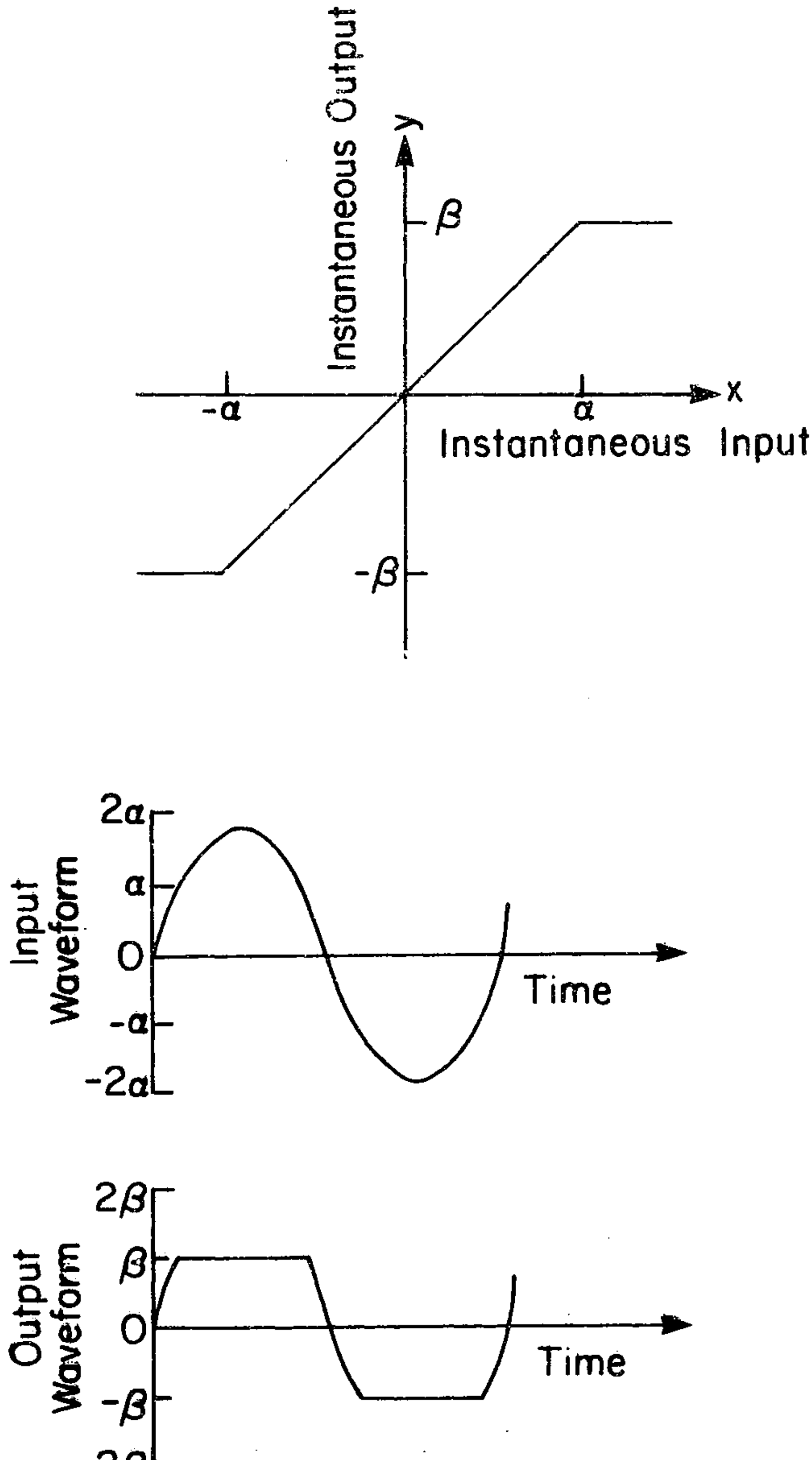

Instantaneous Output
y
β
-β
-α
α
x
Instantaneous Input
2α
α
0
-α
-2α
Input Waveform
Time
2β
β
0
-β
-2β
Output Waveform
Time

FIGURE 5.

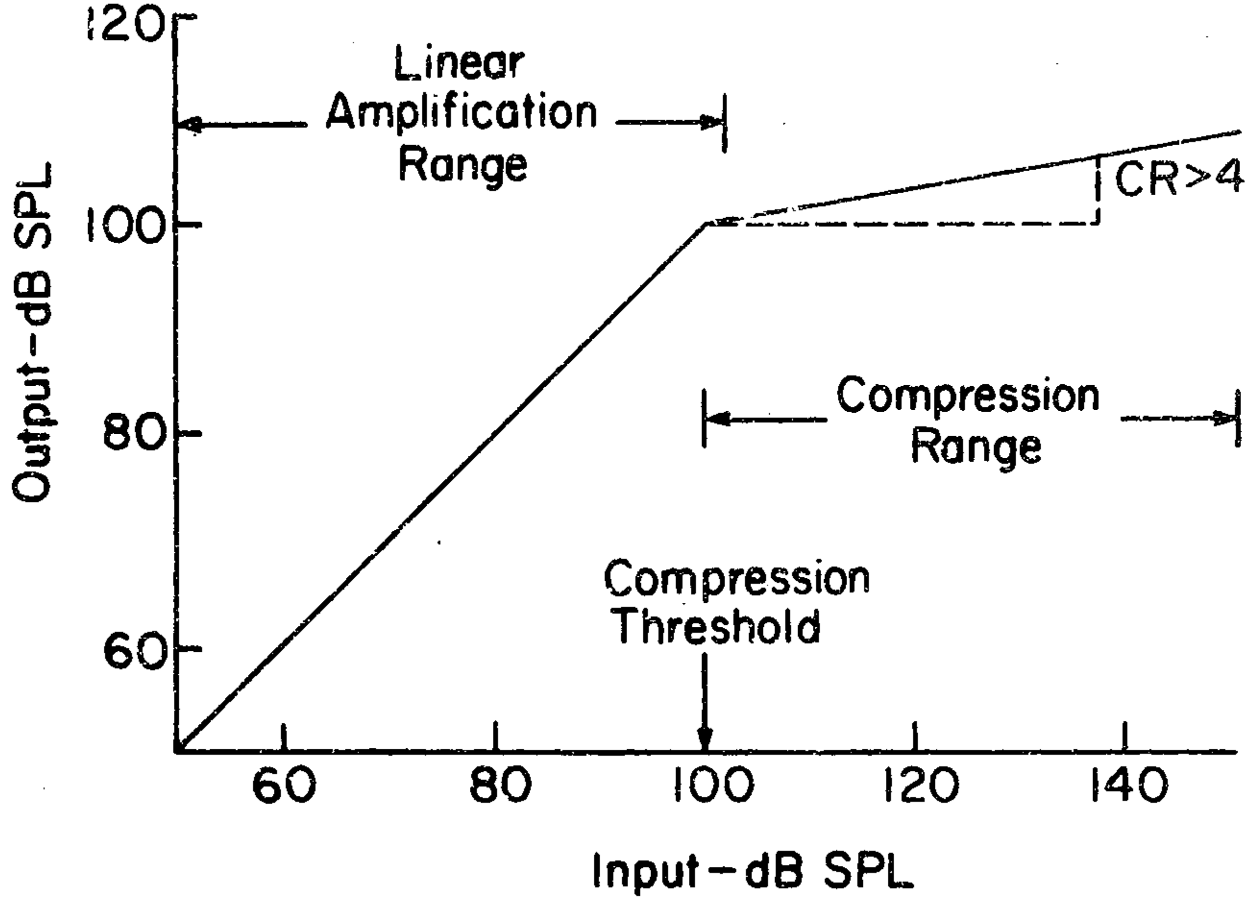

120
100
80
60
Output-dB SPL
Linear
Amplification
Range
CR>4
Compression
Range
Compression
Threshold
60
80
100
120
140
Input-dB SPL

FIGURE 6.

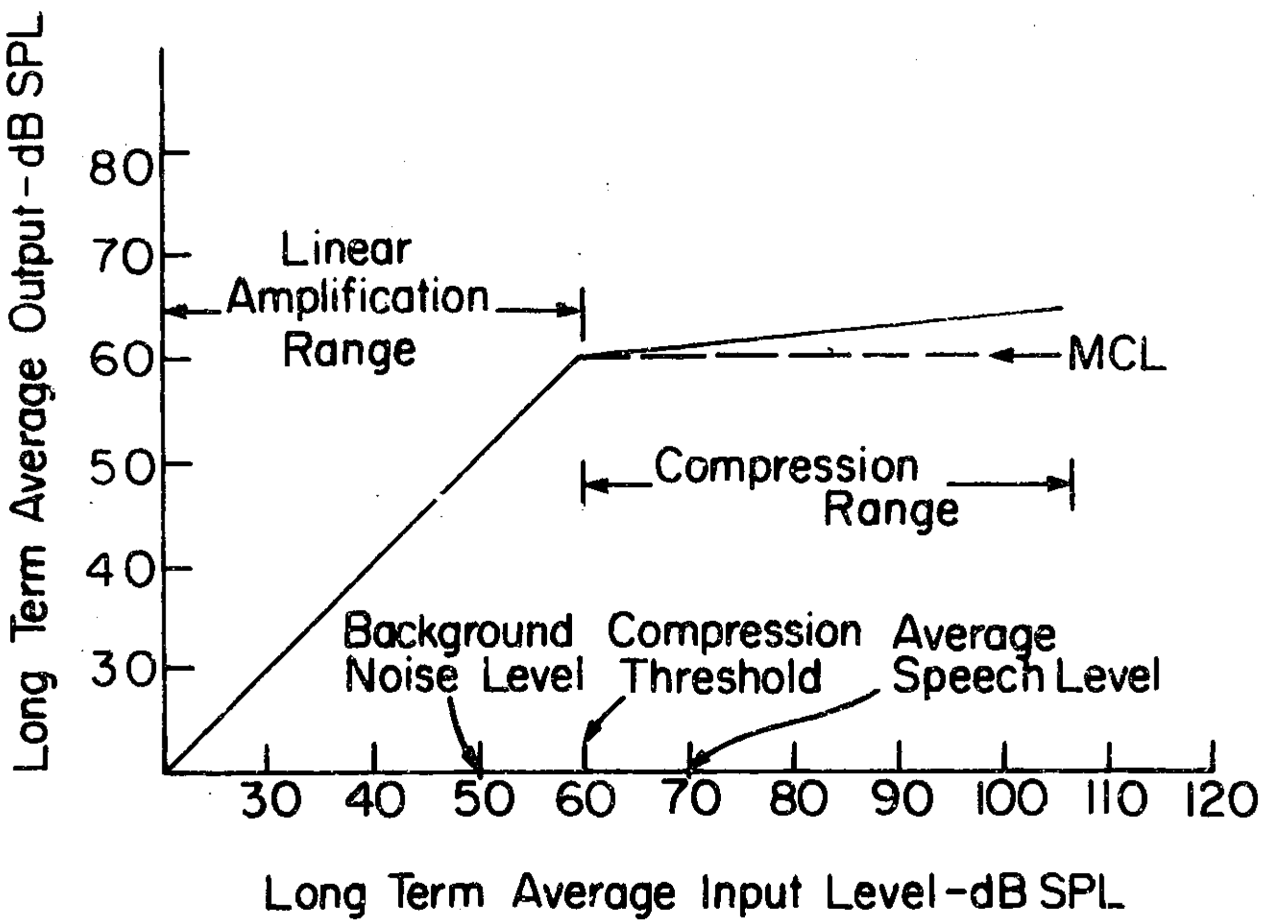

Long Term Average Output-dB SPL
80
70
60
50
40
30
Linear
Amplification
Range
MCL
Compression
Range
Background
Noise Level
Compression
Threshold
Average
Speech Level
30
40
50
60
70
80
90
100
110
120
Long Term Average Input Level-dB SPL

FIGURE 7.

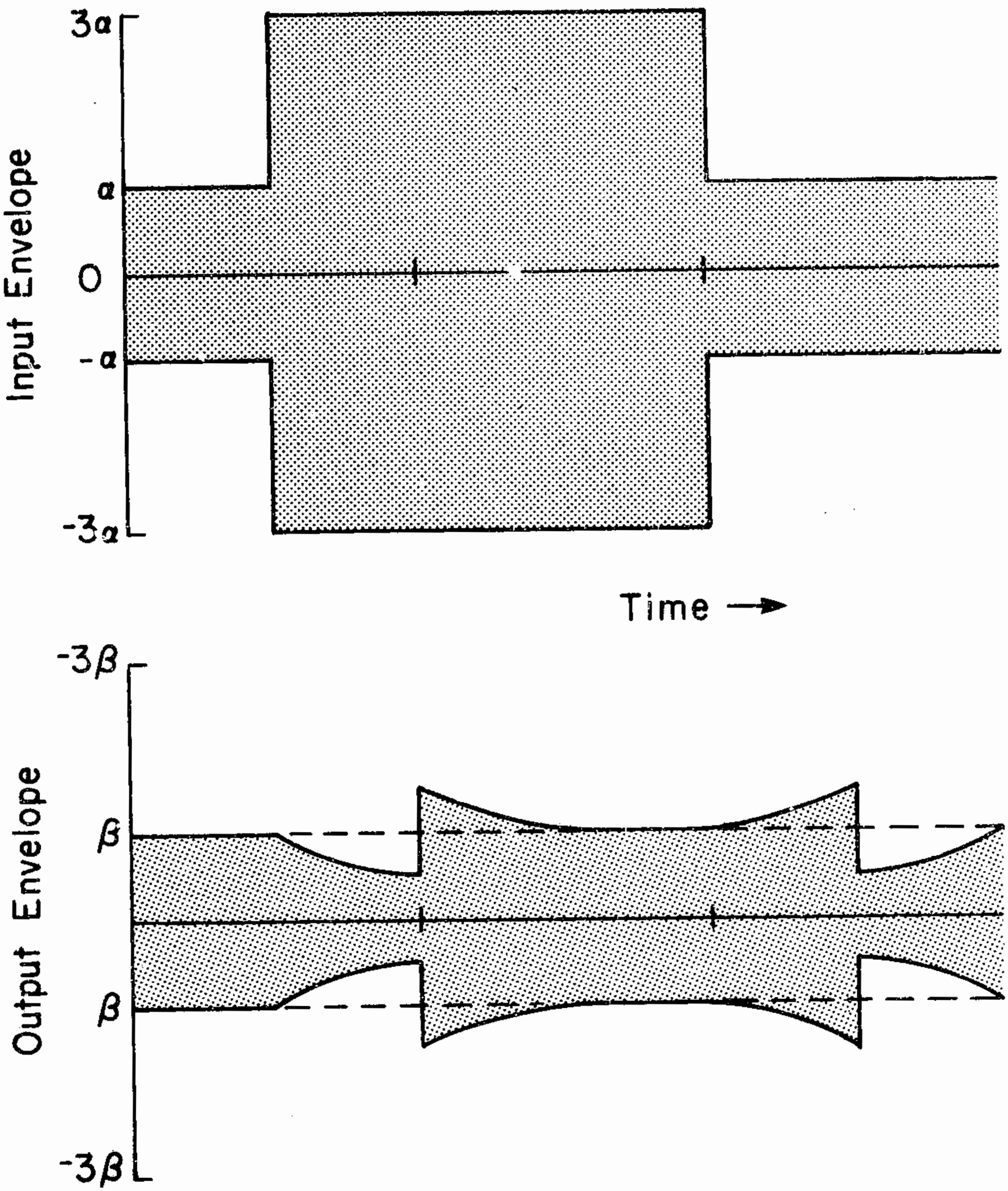


FIGURE 8.

Table 1. Characteristics of Compressor - Limiters

Limiter Type	Attack Time (msec)	Release Time (msec)	Average CR Above Threshold	Output at Compression Threshold (dB SPL 1kHz)	Compression Range (dB)	Distortion	
						Total Harmonic	Freq (Hz)
Nominal	<1	100 to 1000	>4	100 to 140	20 to 60	<1%	100 to 10k
Davis 1947	1	200	10	117	30	<8%	600 800 1k
Hudgins 1948	15 to 30	50 to 100	3	125	15	<10%	1k
Silverman 1951	?	100 to 1000	10	up to 140	25	<2%	?
Blegvad 1974	40	200	3.3	110	40	?	?

Table 2. Characteristics of AVC Compressors

AVC Type	Attack Time (sec)	Release Time (sec)	Average CR Above Threshold	Output at Compression Threshold (dB SPL 1kHz)	Compression Range (dB)	Distortion	
						Total Harmonic	Freq (Hz)
Nominal	.1 to 5	.15 to 5	>4	near MCL	20 to 60	<1%	100 to 10k
Aspinall 1951	.01 to 0.1	.25 to 3	>10	near MCL	30	?	?
Fleming and Rice 1969	.02	.2	2,3,5	near MCL	?	?	?

Table 3. Characteristics of Syllabic Compressors

Researcher	Compressor Type	Attack Time (msec)	Release Time (msec)	Average CR Above Threshold	Compression Range (dB)	Distortion: Total Harmonic	Distortion: Freq. (Hz)
Edgardh 1952	Single-Channel	1-2	20	10	30	low	200-6k
Parker 1953	Single-Channel	<10	10	2.5-3	25-35	(a)	
Kretsinger and Young 1960	Single-Channel	.15	22	>10	>20	4.6%	.3k-3k
	Peak Clipping	NA	NA	>10	>20	18.0%	.3k-3k
Lynn and Carhart 1963	Single-Channel	5-85	30-1200	>5	30	?	-
Caraway and Carhart 1967	3-Channel Instantaneous Cube and Square Rooting	NA	NA	2	30	13-15% (b)	250-500
				3		18-35% (b)	
Trinder 1972	Single-Channel Instantaneous Rooting	NA	NA	>1	>60	>10%	?
Vargo and Carhart 1977	Single-Channel	<1	50	1,2,5	<24	?	-
Burchfield 1971	Identical to Caraway and Carhart, 1964 Plus Filtering	NA	NA	2	30	2-6% (b)	250-500
				3			
Johansson 1973	Single-Channel	0.5-5	10-1000	?	?	?	-
Thomas and Sparks 1971	Infinite Peak Clipping	NA	NA	>10	?	(c)	
Ruhrberg and Esser 1973	Single-Channel	?	10-20	2.5	47	low	>300
Robinson and Huntington 1973	Single-Channel Time Delay	10-20	(d)	?	?	?	-
Gregory and Drysdale 1976	HFCC (f)	<10	<10	>10	>60	low	.4-2.5k
Yanick 1973	Single-Channel	1	30	2.6, 2, 1.4	60	?	-
Yanick 1975	Single-Channel	3	8	2	60	?	-
Nabelek and Robinette 1975	7 Single-Channel Commercial Aids	6-130	30-580	1.5-3	?	(e)	
Villchur 1973	2-Channel	<1	20	>1	>40	<3.5%	250-6.5k
Yanick 1976	1- and 2-Channel	<1	20	>1	40	<3.5%	.25-6k
Yanick and Drucker 1976	Single-Channel plus Expansion	<1	20	>1	40	<3.5%	.25-6k
Barfod 1976	1-,2-,& 3-Channel	6-24	6-24	>1	51	(g)	

a) Silent period noise was almost as loud as the speech signal which was passed through the compressor and re-recorded three times.

b) Distortion = $100\sqrt{(h_2^2 + h_3^2)}/h_1$, where h_i^2 is the energy of the ith harmonic.

c) Only zero crossings preserved.

d) See Figure 7.

e) Distortion during compression overshoot was extreme in some aids and the duration of this distortion ranged from 0 to 70 msec.

f) High-frequency carrier clipping (see text).

g) Distortion was not measured but specifications of system components indicate that total distortion values were similar to those given for Villchur (1973).

REFERENCES

ANSI (1976). S3.22, "Specification of Hearing Aid Characterisics," American National Standards Institute, New York, NY.

Armstrong, M.S. (1969). "Fast Response Dynamic Gain Control Circuit," U.S. Patent 3,581,223.

Aspinall, E. (1951). "A Master Hearing Aid," J. British Institution of Radio Engineers, 11, 45-50.

Barfod, J. (1972). Investigations on the Optimal Corrective Frequency Response for High-Tone Hearing Loss, The Acoustics Laboratory, Technical University of Denmark, Report No. 4.

Barfod, J. (1976). Multichannel Compression Hearing Aids: Effects of Recruitment on Speech Intelligibility, The Acoustics Laboratory, Technical University of Denmark, Report No. 11.

Bevan, W.R. (1973). "A professional Mixer/Compressor for Remote or Studio Use," Preprint 916(B-3), 45th Convention of Audio Engineering Society.

Bizaquet, G. and Veit, P. (1968). "Use of Advanced Apparatus on Deaf Children: Difficulties and Results," Review of Laryngol. Otol. Rhinol. 89, 448-456.

Blackmer, D.E. (1972). "A Wide Dynamic Range Noise-Reduction System," DB, 64, No. 8, 54-56.

Blegvad, B. (1974). "Clinical Evaluation of Behind-the-Ear Hearing Aids with Compression Amplification," Scand. Audiol. 3, 57-60.

Blesser, B.A. (1969). "Audio Dynamic Range Compression for Minimum Perceived Distortion," IEEE Trans. on Aud. and Electroacoustics, AU-17, 22-32.

Blesser, B. and Kent, A.R. (1968). "Analysis of a Feedback-Controlled Limiter Using a Llogarithmic Measuring Scale," IEEE Trans. Aud. and Electroacoustics, AU-16, 481-485.

Brink, B., Borrild, K. and Oving, K. (1975). "Body-Worn Stereophonic Compression Aids used by Profoundly Hearing-Impaired Children," Scand. Audiol. 4, 199-205.

Burchfield, S.B. (1971). "Perception of Amplitude Compressed Speech by Persons Exhibiting Loudness Recruitment," unpublished Ph.D. Thesis Dissertation, Michigan State University.

Burnett, E.D. and Bassin, M.A. (1976). "Attack and Release Characteristics of Compression Hearing Aids," National Bureau of Standards, Washington, D.C. NBSIR 76-1179.

Burnett, E.D. and Schweitzer, H.C. (1977). "Attack and Release Times of Automatic-Gain-Control Hearing Aids," J. Acoust. Soc. Am 62, 784-786.

Burwen, R.S. (1971). "Design of a Noise Eliminator System," Preprint No. 838 (B-8), 41st Convention of the Audio Engineering Society.

Caraway, B. (1964). "Effects on Speech Intelligibility of Dynamic Range Reduction with a Compressor Amplifier," unpublished Ph.D. Thesis Dissertation, Northwestern University.

Caraway, B.J. and Carhart, R. (1967). "Influence of Compressor Action on Speech Intelligibility," J. Acoust. Soc. Am. 41, 1414-1438.

Carter, R.O. (1964). "Theory of Syllabic Compandors," Proc. IEEE 111, 503-513.

Daniel, D.B. (1957). "The Consonant Amplifier-Limiter," Radio and Television News 57, 49-51.

Davis, H., Stevens, S.S., Nichols, R.H., Hudgins, C.V., Peterson, G.E., Marquis, R.J., and Ross, D.A. (1947). Hearing Aids - An Experimental Study of Design Objectives, Harvard University Press, Cambridge.

de Vos, A.W. (1969). Discussion following paper by Fleming and Rice (1969). Int. Audiology 8, 523.

Dolby, R.M. (1967). "An Audio Noise Reduction System," J. Aud. Engineering Society 15, 383-388.

Dudley H. and Gruenz, O.O. (1964). "Visible Speech Translators with External Phosphors," J. Acoust. Soc. Am. 18, 62-73.

Edgardh, B.H. (1952). "The Use of Extreme Limitation for the Dynamic Equalization of Vowels and Consonants in Hearing Aids," Acta-Otolaryng. 40, 376-382.

Egan, J. P. (1948). "Articulation Testing Methods," Laryngoscope 58, 955-981.

Fairbanks, G. and Miron, M.S. (1957). "Effects of Vocal Effort Upon the Consonant-Vowel Ratio within the Syllable," J. Acoust. Soc. Am. 29, 621-626.

Fleming, D.B. and Rice, C.G. (1969). "New Circuit Development Concepts in Hearing Aids," International Audiology 8, 517-523.

Fournier, J.E. (1951). Audiometrie Vocale, Librairie Maloine S.A., Paris.

Gregory, R.L. and Drysdale, A.E. (1976). "Squeezing Speech into the Deaf Ear," Nature 264, 748-751.

Hathaway, J.L. (1950). "Automatic Audio Gain Controls," Audio Engineering 34, September, 16-18.

Hecker, M.H.L. ((1974). "A Study of the Relationship Between Consonant-Vowel Ratios and Speaker Intelligibility," unpublished Ph.D. Thesis Dissertation, Stanford University.

Hellwarth, G.A. and Jones, G.D. (1967). "Automatic Conditioning of Speech Signals," Proc. 1967 Conference on Speech Communication and Processing.

Hirsh, I. J., Davis, H., Silverman, S. R., Reynolds, E. G., Eldert, E., and Benson, R. W., (1952). "Development of Materials for Speech Audiometry," J. Speech Hearing Disord. 17, 321-337.

Hood, J.D., and Poole, J.P. (1966). "Tolerable Limit of Loudness : Its Clinical and Physiological Significance," J. Acoust. Soc. Am. 40, 47-53.

House, A., Williams, C., Hecker, M., and Kryter, K. (1965). "Articulation- Testing Methods: Consonantal Differentiation with a Closed-Response Set," J. Acoust. Soc. Am. 37, 158-166.

Hudgins, C.V., Marquis, R.J. Nichols, R.H., Peterson, G.E., and Ross, D.A. (1948). "The Comparative Performance of an Experimental Hearing Aid and Two Commercial Instruments," J. Acoust. Soc. Am. 20, 241-258.

Huizing, H.C., and Rejntes, J.A. (1952). "Recruitment and Speech Discrimination Loss," Laryngoscope 62, 521-527.

IEEE (1969). "Recomended Practice for Speech Quality Measurements," IEEE Trans. Audio and Electro., AU-17, 225-247.

Johansen, P.A. and Simonsen, E. (1973). "Some Examinations to Determine the Best Frequency Characteristic of Hearing Aids for Patients with Hearing Loss in the High-Tone Range," Scandanavian Audiology 2, 55-58.

Johansson, B. (1973). "The Hearing Aid as a Technical Audiological Problem," Scandinavian Audiology Suppl. 3, 55-76.

Johansson, B. and Lindblad, A.V. (1971). "The Use of Compression and Frequency-Transposition in Hearing Aids," in: Electroacoustic Characteristics Relevant to Hearing Aids, H.W. Ewertsen (Ed.) Third Dadavox Symposium, Copenhagen, Denmark, Scandinavian Audiology Suppl. 1, 68-71.

Kaiser, A. and Bauer, B.B. (1962). "A New Automatic Level Control for Monophonic and Stereophonic Broadcasting," IRE Trans. on Audio AU-10, 171-173.

Krebs, D. (1972). "Output Limiting and Related Harmonic Distortion," Oticongress 2, 1-10.

Kretsinger, E.A. and Young, N.B. (1960). "The Use of Fast Limiting to Improve the Intelligibility of Speech in Noise," Speech Monographs 27, 63-69.

Kruel, E.J., Nixon, J. C., Kryter, K.D., Bell, D.W., Lang, J.S., and Schubert, E.D. (1968). "A Proposed Clinical Test for Speech Discrimination," J. Speech and Hearing Research 11, 536-552.

Ludvigsen, C. and Nielsen, H.B. (1975). "The Effect of Body-worn Stereophonic Compression Hearing Aids on Severely Hearing-Impaired Adults," Scandinavian Audiology 4, 191-197.

Lynn, G. and Carhart, R. (1963). "Influence of Attack and Release in Compression Amplification on Understanding of Speech by Hypoacusics," J. Speech and Hearing Disorders 28, 124-140.

McCandless, G.A. (1973). "Hearing Aids and Loudness Discomfort," Oticongress 3, 39-44.

Nabelek, I.V. (1973). "On Transient Distortion in Hearing Aids with Volume Compression," IEEE Transactions on Aud. and Electroacoustics, AU-21, 279-285.

Nabelek, I.V. and Robinett, L.N. (1975). "Comparison of Hearing Aids with Amplitude Compression," J. Acoust. Soc. Am. 57, 519(A).

NOBLE, J.J. and Bird, J.R. (1969). "A Dual-Band Audio Limiter," J. Aud. Engr. Soc. 17, 678-684.

Parker, C.D. (1953). "The Effects of the Reduction of 'Short-Time Fatigue' on Speech, for 'Perceptively' Deafened Individuals," unpublished Ph.D. Thesis Dissertation, University of Iowa.

Pascoe, D. P. (1975). "Frequency Response of Hearing Aids and Their Effects on the Speech Perception of Hearing-Impaired Subjects," Annals of Otol., Rhinol., and Laryngol., Supp. 23, Vol. 84.

Pestalozza, G. (1953). "Comparative Study of the Efficiency of Various Volume Control Devices of Hearing Aids," Archives of Italian Otolaryn. 64/2, 195-206.

Peterson, G. and Lehiste, I. (1962). "Revised CNC Lists for Auditory Tests," J. Speech and Hearing Research, 27, 62-72.

Poliakoff, A. (1950). "Hearing Aid Design," Wireless World 56, 274-276.

Portmann, M. and Portmann, C. (1961). Clinical Audiometry, Charles C. Thomas, Springfield, Ill.

Richards, A.M. (1973). "Loudness Growth Under Masking: Relation to True Sensorineural Impairment," J. Speech and Hearing Res. 16, 597-607.

Rintelmann, W.F. (1972). "Effects of Amplitude Compression Upon Speech Perception: A Review of Research," Scandinavian Audiology 1, 127-134.

Robinson, C.E. and Huntington,, D.A. (1973). "The Intelligibility of Speech Processed by Delayed Long-Time-Averaged Compression Amplification," J. Acoust. Soc. Am. 54, 314(A).

Ruhrberg, W. and Esser, G. (1973). "Die Wirkung der Dynamik-kompression bei Innenohrschwerhorightkeit und Retrocochlearer Horstorung," HNO 21, 247-255.

Salmon, R.D. (1970). "Talker Variation as Related to Intelligibility," Tech. Rept. 31, Communication Sciences Laboratory, University of Florida, Gainesville.

Shorter, D.E.L., Manson, W.I., and Stebbings, D.W. (1967). "The Dynamic Characteristics of Limiters for Sound Programme Circuits," BBC Eng. Monograph No. 70.

Silverman, S.R. (1949). "The Implications for Schools for the Deaf of Recent Research on Hearing Aids," Am. Ann. Deaf 94, 325-339.

Silverman, S.R. and Harrison, C.E. ((1951). "The National Research Council Group Hearing Aid Project, A progress Report," Am. Ann. Deaf 96, 420-431.

Steinberg, J.C. and French, N.R. (1946). "The Portrayal of Visible Speech," J. Acoust. Soc. Am. 18, 4-16.

Stevens, S.S. (1966). "Power Group Transformations Under Glare, Masking, and Recruitment," J. Acoust. Soc. Am. 39, 725-735.

Thomas, I.B. and Sparks, D.W. (9171). "Discrimination of Filtered-Clipped Speech by Hearing Impaired Subjects," J. Acoust. Soc. Am. 49, 1881-1887.

Tillman, T. W., Carhart, R., Wilber, L. A. (1963). "A Test for Speech Discrimination Composed of CNC Monosyllabic Words (N.U. Auditory Test No. 4)." Tech. Doc. Rept. No. SAM-TDR-62-135, USAF School of Aerospace Medicine, Aerospace Med. Div. (AFSC), Brooks AFB, Tex.

Tillman, T. W., and Carhart, R. (1966). "An Expanded Test for Speech Discrimination Utilizing CNC Monosyllabic Words (N. U. Auditory Test No. 6)." Tech. Doc. Rept. No. SAM-TDR-66-55, USAF School of Aerospace Medicine, Aerospace Med. Div. (AFSC), Brooks AFB, Tex.

Torick, E.L., Allen, R.G., Bauer, B.B. (1968). "Automatic Control of Loudness Level," IEEE Trans. Broadcasting BC-14, 143-146.

Trinder, E. (1972). "An Attempt to Correct Speech Discrimination Loss in Cochlear Deafness by Graded Instantaneous Compression," Sound 6, 62-67.

Vargo, S.W. and Carhart, R. (1972). "Amplitude Compression. I. Speech Intelligibility in Quiet with Normal and Pathological Hearing Groups," J. Acoust. Soc. Am. 53, 327(A).

Vargo, S.W. (1977). "Intelligibility of Amplitude Compressed Speech in Quiet: Normal and Pathological Hearing Groups," Audiologgy and Hearing Education, February/March, 16-17.

Viet, I. (1973). "Practical Possibilities in the Design of Hearing Aids with Automatic Gain Control," J. Audiol. Tech. 12, 186-200.

Villchur, E. (1973). "Signal Processing to Improve Speech Intelligibility in Perceptive Deafness," J. Acoust. Soc. Am. 53, 1646-1657.

Wallenfels, H.G. (1967). Hearing Aids on Prescription, Charles Thomas, Springfield, Ill.

Williams, C.E., Hecker, M.H.L., Stevens, K.N., and Woods, B. (1966). "Intelligibility Test Methods and Procedures for the Evaluation of Speech Communications Systems," Rept. ESD-TR-66-677, Decision Sciences Laboratory, Electronics Systems Division, Air Force Systems Command.

Yanick, P., Jr. (1973). "Improvement in Speech Discriminaation with Compression vs. Linear Amplification," J. Auditory Res. 13, 333-338.

Yanick, P., Jr. (1975). "Discrimination in the Presence of Competition with an AVC vs. a DRC Hearing Aid," submitted to J. Am. Audio. Society.

Yanick, P., Jr. (1976). "Effects of Signal Processing on Intelligibility of Speech in Noise for Persons with Sensorineural Hearing Loss," J. Amer. Audiol. Soc., 1, 229-238.

Yanick, P., and Drucker, H. (1976). "Signal Processing to Improve Intelligibility in the Presence of Noise for Persons with Ski-Slope Hearing Impairment," IEEE Trans. Acoust. Speech and Signal Processing, ASSP-24, 507-512.

Part 2

A Study of Multichannel Amplitude Compression and Linear Amplification for Persons With Sensorineural Hearing Loss

Introduction

There are three types of amplitude compression systems that have been suggested for use by persons with sensorineural hearing loss (Lippmann et al. 1978). Figure 1 illustrates these systems arranged in one aid. Each includes a variable-gain linear amplifier whose gain is a function of the input and/or output levels and all do not distort steady state input signals at normal input levels. Automatic Volume Control or AVC is designed to keep the long-term speech output level constant over intervals corresponding to phrases or sentences. This is useful for persons who obtain maximum speech intelligibility over only a limited range of presentation levels. The gain variation introduced by AVC is slow and does not affect the level variations that occur between or within words or syllables in a sentence or phrase. Compression-limiting is designed to prevent output levels from becoming excessively loud or painful. This is useful for persons with normal loudness discomfort levels who use amplification to overcome loss in absolute sensitivity. Compression-limiting acts rapidly to reduce the level of intense sounds and generally has little effect on the level variation of speech at lower, more normal, input levels. Syllabic, or fast compression is designed to modify the short-term (within or between words or syllables) level variations of speech in order to match the dynamic range of speech to the residual dynamic range of an impaired listener. This type of compression can, in effect, compensate for loudness recruitment (an abnormal growth of loudness with intensity) and a multichannel syllabic compression system can be adjusted to restore all equal loudness contours to normal. Past research on these three types of compression has been reviewed by Lippmann et al. (1978). Recent research has focused on multichannel syllabic compression because it is less well understood than the other types of compression and it holds more promise for aiding speech perception than do the other types of compression. Past research is inconclusive, however. The results of Barfod (1976) obtained with a 4-channel syllabic compression system differ from the more positive results of Villchur (1973) and Yanick (1976) obtained with 2-channel compression systems. Barfod's system when adjusted to restore normal equal loudness contours for subjects with sharply sloping mild-moderate losses performed no better than a linear system with 20-40 dB of high-frequency emphasis. Villchur's and Yanick's systems when adjusted by subjects with flat and sharply sloping mild-severe losses performed significantly better than linear systems with limited high-frequency emphasis.

The primary purpose of the present study was to further explore the effect of multichannel syllabic compression on the intelligibility of speech for persons with sensorineural hearing loss. This study was designed to investigate compression which restores normal equal loudness contours and to answer some of the questions concerning compression that have been raised by recent studies. Extensive training with compressed materials was specifically excluded from this study because it was assumed that compression would restore normal speech cues that subjects were experienced with. Important characteristics of this study include: 1) the use of a compression system that restores normal equal loudness contours for pure tones and of another compression system with reduced compression ratios and subject-adjusted equalization; 2) the use of four linear reference systems including a flat orthotelephonic (French and Steinberg, 1947) system plus three others chosen on the basis of recent research that have differing amounts of high-frequency emphasis; 3) the use of subjects with both flat and sloping moderate losses; 4) testing with both standard and specially designed speech materials in quiet/anechoic and noisy/reverberant environments with male and female talkers;

5) testing at each subject's most comfortable level and also at reduced input levels; 6) the use of a circumaural earphone that had been calibrated relative to free-field, and 7) the use of a 16-channel compression system in which the compression curve (output vs. input) in each channel can be independently adjusted and in which the attack and release times in each channel are fixed.

In the remainder of this paper we first describe the subjects used in this research and the three experiments that were performed and then discuss and compare the results of these experiments to the results of other recent research on amplitude compression and linear amplification. Experiment I involved comparisons of the 4 linear and 2 compression systems using PB-50 words in nonsense sentences (PBNS's) and CVC nonsense syllables spoken by male and female talkers, with and without added noise and reverberation and presented at each subject's most comfortable level. Experiment II involved comparisons of the orthotelephonic and the best linear and compression systems from experiment I using standard word and sentence tests. Experiment III involved comparisons of the best linear and compression systems from experiment I using nonsense syllables spoken by a female talker with input levels below the levels used in Experiment I.

1. Subjects

Subjects were required to have stable, moderate, sensorineural losses with recruitment and reduced dynamic range. These losses could be bilateral or unilateral; however, thresholds were required to be roughly normal at one or more frequencies between 100 and 8000 Hz in either ear. This allowed equal loudness contours to be related to normal equal loudness contours via either normal hearing at low frequencies or in the contralateral ear. Five subjects ranging in age from 24 to 60 were obtained through the Massachusetts Eye and Ear Infirmary.

1.1 Clinical Measurements

Subjects were examined by an otolaryngolist and audiologist both before and after the experimental program (Experiments I, II and III) and were diagnosed as having sensorineural losses. This diagnosis was based on each subject's history plus clinical measurements of air and bone conduction thresholds, speech reception threshold (SRT), speech discrimination (CID W-22 word lists), tone decay, and when possible, alternate binaural loudness balances (ABLB's). This information is presented in Appendicies 1 and 2. Two of the subjects had unilateral losses that were relatively flat and either congenital(IK) or associated with Meniere's disease(FM). The other three subjects had bilateral sloping losses. Subjects ED and HS had ISO HTL's that increased with a slope of roughly 20 dB per octave above 500 Hz and losses that were diagnosed as sudden (possibly a virus) and congenital. Subject ES had ISO HTL's that were less than 10 dB below 1000 Hz and greater than 60 dB above 1500 Hz and a loss that was attributed to noise exposure. Only ED and ES wore hearing aids. In the remainder of this paper all measurements refer to those made in the more severely impaired ear for IK, FM, HS, and ES, and in the ear in which a hearing aid was normally worn for ED.

1.2 Laboratory Measurements

Air conduction thresholds, loudness discomfort levels (LDL's) and equal loudness contours were measured in the Auditory Perception Laboratory at MIT in order to verify the clinical measurements and specify the characteristics of some of the systems that were to be studied. These measurements were made at the beginning of the experimental program in three to four sessions lasting two to three hours each and threshold measurements were repeated at the end of the experimental program. All measurements were made using a circumaural earphone that was calibrated relative to free-field (Lippmann, 1978) and masking noise was applied to the opposite ear using procedures similar to those suggested by Studebaker (1967). Also, all measurements were made using sinewaves with 300 msec durations and 25 msec rise and fall times. All pre-experimental measurements are presented in Appendicies 3, 4, and 5.

1.2a Thresholds

Air conduction thresholds were measured using a 4-interval forced-choice adaptive procedure that was similar to that used by Reed and Bilger[1] (1973) . The pre-experimental thresholds for all subjects are presented in Figure 2. The two groups indicated in Figure 2 were determined on the basis of threshold curves. Subjects IK and FM had fairly flat threshold curves and were placed in the flat-loss group and subjects ED, ES, and HS had sharply sloping threshold curves and were placed in the sloping-loss group. In the remainder of this paper we report group average results and note any significant individual differences. Group average pre-experimental thresholds are presented in Figure 3. The thresholds shown in Figure 3 did not change significantly during the experimental program. The difference between these thresholds and those measured after the experimental program averaged over the region 250-8000 Hz was 2 dB. The maximum individual difference (a threshold increase of 8 dB) occurred at 1500 Hz for FM. Variations in thresholds for this subject at 500, 1000, 2000, and 3000 Hz were all however, less than 2 dB. The average difference between pre-experimental laboratory threshold measurements (re ISO, 1961 binaural free-field thresholds) and all pre-experimental clinical threshold measurements (re ISO, 1964 and excluding FM) was small (less than 0.5 dB). Only two measurements at the extreme frequencies (250 and 8000 Hz) differed by more than 10dB.

1.2b Loudness Discomfort Levels

Loudness discomfort levels were measured using a tracking procedure that was similar to that used by Morgan et al. (1974)[2]. Individual LDL's for the five subjects are presented in Figure 2 and average LDL's for the two subject groups are presented in Figure 3. LDL's for three normal hearing subjects were roughly 110 dB SPL from 250-4000 Hz. LDL's for the impaired subjects were equal to or below these levels. This result is consistent with the clinical diagnosis of a sensorineural loss and demonstrates that the subject's losses did not have significant conductive or eighth nerve components(Hood and Poole,1966).

1.2c Loudness Balances

Monaural loudness balances (MLB's) and alternate binaural loudness balances (ABLB's) were performed in order to determine equal loudness contours for pure tones and relate these to normal contours (ISO, 1961). All loudness matches were performed using the method of adjustment, using balanced procedures to minimize the bias effects that are common to loudness matches (e.g. Stevens and Greenbaum, 1966; Hellman and Zwislocki, 1964). Loudness balances were not performed at high levels (90-100 dB SPL) because persons with sensorineural losses typically do not listen to speech at these levels[3]. The procedures used to perform loudness balances and the methods used to determine equal loudness contours from these balances and to relate these contours to normal are described in Appendices 4 and 5.

1.2d Average Equal Loudness Contours

Average equal loudness contours for the sloping and flat-loss groups are presented in Figure 3. At some frequencies between 250 and 6000 Hz segments have been extended above the measured MLB's to the 80 phon contour by assuming that the loss at and above the highest level measured was purely conductive[5]. The phon labels in this figure are derived from the assumptions presented in Appendix 5, the loudness balances, and normal equal loudness contours. These contours are representative of the results for individual subjects and of contours presented by Barfod (1976) and Reger (1936) in that the flat-loss curves are roughly parallel, the sloping-loss curves converge at high frequencies and high level contours flatten out relative to the threshold curve. The close spacing of the contours of the flat-loss group and the convergence of the contours of the sloping-loss group are further indications of recruitment.

2. Experiment I

The primary purpose of Experiment I was to determine the effect on intelligibility of a 16-channel syllabic compression system adjusted to restore normal equal loudness contours for pure tones for listeners with sensorineural loss. Some preliminary and past research, however, indicated that better intelligibility might be achieved with a compression system that had reduced compression ratios relative to the system which restored normal equal loudness contours and that had subject-adjusted high-frequency emphasis. Thus a compression system of this type was included. Ideally the compression systems would have been compared only to the "best" linear system. This is not presently possible. Although past research indicates that the "best" linear system should have characteristics which lie between those of a system with a standard orthotelephonic response and those of a system that mirrors the audiogram, it does not allow one to choose the "best" linear system for a given subject (see Braida et al., 1978 for a discussion of this issue). It was thus decided to include three linear systems that sample some likely candidates for the "best" linear system. It was also decided to include a standard orthotelephonic linear system for reference purposes. Thus, a total of two compression and four linear systems were studied. These systems were compared using PB-50 words in nonsense sentences (PBNS'S) and CVC nonsense syllables spoken by a male and a female talker and presented in a quiet/anechoic(Q/A) and noisy/reverberant(N/R) environment at each subject's most comfortable level.

2.1 Procedures

The five subjects described in Sec. 1 participated in nine sessions in which speech tests were administered to evaluate the six systems. The first of these was used for training with the different systems and the different talker-material-environment conditions. In each of the other eight sessions seven lists for one condition were used. The relationship between session number (within the above eight sessions) and condition tested in the session was:

1) MALE-SENTENCES-Q/A,
2) FEMALE-SENTENCES-Q/A,
3) MALE-SENTENCES-N/R,
4) FEMALE-SENTENCES-N/R,
5) MALE-CVC's-Q/A,
6) FEMALE-CVC's-Q/A,
7) MALE-CVC's-N/R,
8) FEMALE-CVC's-N/R.

The order in which systems were tested within a session was randomized to minimize order effects and also to minimize the effect of differences in the underlying intelligibility of lists used for one conditon. Seven lists were normally presented in one session. The first was used for practice and to measure stability of scores within a session (the system tested with this list was also tested with the last list of the session).

All speech tests were administered monaurally using circumaural earphones that were compensated for their free-field response. Feedback was not provided.

Materials were presented at each subject s most comfortable level (MCL). This was determined before the administration of every list using speech material consisting of the previous list or an extra list. It was described to subjects as a level that provides maximum intelligibility but is comfortable for long-term listening. For the 4 linear systems and the compression system with reduced compession ratios, the MCL was determined by having subjects adjust the system output level. A bracketing procedure was used for this adjustment. The average of two adjustments was taken as the MCL if they were within 3-4 dB of each other; otherwise, adjustments were repeated. The MCL for the compression system which restored normal equal loudness contours was determined by varying the input level to the system in 10 dB steps from an overall RMS level of 35 to 85 dB SPL and having subjects make binary comparisons. The input instead of output level of this compression system was varied when determining the MCL because equal loudness contours can be restored to normal only if the relationship between the system output and eardrum sound pressure is known and fixed. The input level to the other system was not varied because the overall equalization provided by that system was determined by varying the output level with the input level fixed and this equalization might not be appropriate for other input levels.

Masking was used in the non-test ear during speech tests. This was necessary because some subjects had low thresholds in the non-test ear, subjects listened at above threshold levels in the test ear, and the interaural attenuation obtained with earphones (typically 40 dB from 125 to 1000 Hz and 60 dB above 1000 Hz, e.g. Zwislocki, 1953; Chaiklin, 1967) is not large enough to totally isolate the two ears. Masking was used for subjects IK, FM and HS in frequency regions where thresholds in the non-test were 10 dB or more below thresholds in the test ear. White noise that was low-pass filtered at 5 or 6 kHz was used as a masker at a level that produced free-field thresholds of 23 dB at 250 Hz, 37 dB at 500 Hz and greater than 40 dB from 1000 to 6000 Hz. The overall level of this noise (65-70 dB SPL) was chosen to be comfortable for long periods of testing. Masking was not used for ED because thresholds in her non-test ear were higher than in her test ear. Masking was not used for ES because thresholds in his two ears differed by less than 15 dB up to 3kHz and above 3kHz tones in the tested ear would have had to exceed a level of roughly 110 dB SPL to be audible in the non-test ear.

Three normal hearing subjects were used in Experiment I to determine the variability and overall level of difficulty of the speech test lists. These subjects listened in quiet to unfiltered materials and in noise to low-pass filtered materials. All material was presented at a comfortable level (65-80 dB SPL) using a linear amplification system that was electrically flat and using TDH-39 drivers with .001A cushions. This system had a roughly orthotelephonic response from 250-6000Hz (DeGennaro, 1978). Subjects participated in sixteen sessions and in each session 8-9 lists from one condition were normally administered. The first one or two lists in these sessions were used for practice.

Filtered materials were presented in the first eight sessions and unfiltered materials were presented in the last eight sessions.

2.2 Systems

Characteristics of the systems studied are summarized in Table 1.

2.2a Linear Systems - Design

Four linear systems were included. One (ORTHO) had a flat functional gain or orthotelephonic response. This system simulated (except for overall gain) the sound transmission path that exists in a free field with talker and listener facing each other at a distance of one meter. The other three linear systems had different amounts of high-frequency emphasis and were chosen on the basis of past research and preliminary experiments. One system (MA) mirrored each subject's audiogram as suggested most recently by Pascoe (1975). Another, (OMCL) was suggested by preliminary experiments and was determined by having subjects listen to filtered bands of speech (110-700, 700-1400, 1400-2800 and 2800-9000 Hz) in isolation and adjust each band to the maximum level that would be "comfortable for long periods of listening". This was done both for male and female speech and the average relative gain in each band determined a piece-wise linear frequency gain characteristic which defined the OMCL system[6]. The fourth linear system (10%) attempted to restore normal loudness to the 10% cumulative levels of speech in each frequency region as suggested by Barfod (1972). The 10% cumulative level or simply 10% level represents the level exceeded during continuous speech by only 10% of all measurements made. For the 10% system these refer to levels measured in band-pass filters with widths equivalent to psychological critical bands (Zwicker, 1957). These levels were calculated (for a normal overall RMS speech level of 65 dB SPL) using the average male-female 10% RMS levels measured in 1/8 second intervals that are presented by Dunn and White (1940). They were converted to critical band measurements using the method suggested by Dunn and White, and are presented in Appendix 7. The frequency-gain characteristic of the 10% system was determined from these levels and the measured equal loudness contours. Since these levels fall off at high frequencies this system does not restore any single equal loudness contour to normal but restores normal loudness to high-level signals at low frequencies and to low-level signals at high frequencies.

2.2b Compression Systems - Design

The first of the two compression systems studied (EL) was designed to restore normal equal loudness contours for pure tones. This system has been suggested by a large number of researchers (e.g. Villchur, 1973; Barfod, 1976, etc.) but has never been tested. All past studies utilized compression systems with too few channels to allow compression characteristics to be varied with frequency as required for this system or included subject adjustments of compression characteristics. The characteristics of this system were determined from the relationship between equal loudness contours for each subject and normal equal loudness contours[7]. The accuracy with which this system restored normal equal loudness contours was limited by errors in the measured equal loudness contours, variation in the sound pressure level delivered to the eardrum, and equipment limitations. The combined error in the gain of the EL system at any level caused by the first two factors is estimated to be less than ±3 dB below 1-2 KHz and less than ±6 dB above 1-2 KHz. Equipment limitations were important only at high frequencies and are discussed in the next section. The compression characteristics of the second system (COMCL) were chosen on the basis of preliminary experiments, informal listening, and past research on compression. The compression ratio was 1 up to 500 Hz then increased to a maximum of 3 at and above 2 KHz[8]. The overall equalization provided by the COMCL system was determined using the same procedures used for the OMCL system except subjects listened to bands of compressed speech.

2.2c Implementation

All systems were implemented using the speech-processing system shown in Figure 4. Detailed descriptions of the hardware and software of this system are presented by Krasner (1974), Cuddy (1972), Boddie (1971), and Homan (1974). In this system, 16 channels are formed by 16 filters with center frequencies from 160-8000 Hz and bandwidths that approximate the widths of critical bands (1/3 octave filters except for the lowest two channels which have 1 and 2/3 octave widths and are formed by summing outputs of 1/3 octave filters). The output level of the filter in each channel is measured by a level detector, logged and sampled every 1.4 msec and used to control (via computer look-up tables) the settings of 16 computer-controlled digital attenuators that were adjustable from 0 to 63 dB in 1 dB steps. This allowed the compression curve in each channel to be specified to an accuracy of 1 dB over the input dynamic range of the system. The detector time constants largely determined the attack and release times of the system. The detectors were peak-detectors followed by RC filters, both chosen such that attack times were less than 1.5 msec (except in the lowest 4 channels where the transient response of the filters eliminated rapid increases in level and the attack times were 6, 3, 2.5, and 2 msec for channels 1, 2, 3, and 4) and the release times were roughly

20 msec (except in the lowest frequency channel where the release time was increased to 32 msec to prevent distortion[9] to low frequencies) .

High-frequency emphasis of 10 dB above 1250 Hz was included in the input filter section to make maximum use of the dynamic range of the system. With this emphasis the dynamic range of the detectors and also the compression range in each channel for speech signals was 55-60 dB and the maximum obtainable high-frequency emphasis was about 65 dB. This dynamic range was maximally utilized by adjusting the overall RMS level of the input speech signal such that 10% cumulative speech levels (as measured by the level detectors) were 15-20 dB below the clipping level of the system.

2.2d System Characteristics

2.2d.1 Output 10% Levels

The 10% cumulative output levels of the 6 systems used in Experiment I averaged over the 8 talker-material-environment conditions and over subjects in the flat and sloping subject groups are presented in Fig. 5. Characteristics for individual subjects are presented in Appendix 9. These curves were calculated from the most comfortable levels chosen by subjects, the 10% levels of the speech materials used, and the functional characteristics of the different systems. The average curves are representative of curves in the 8 conditions because the MCL's for each system did not change greatly in the different conditions (see section 2.2d.4) and because the 10% cumulative levels (re overall RMS level) of the different speech materials differed little (see Appendix 6). In Figure 5 the 10% output levels for the ORTHO linear system are identical (except for level) for both subject groups because they correspond to the uniformly amplified 10% input levels. Levels for the 10% linear and EL compression system and for the OMCL linear and COMCL compression system are represented by single curves because it was found that the average difference between these respective pairs of curves was less than 3dB when averaged over subects and less than 5 dB for individual subjects. This result suggests that the MCL is determined by peak speech levels and does not depend strongly on compression below the 10% levels.

The average curves for the sloping-loss group indicate that systems differ primarily in their output levels at low frequencies (1 kHZ). All except the ORTHO system have roughly similar output levels above 2-4 kHz. The 10% levels for the ORTHO system fall below threshold above 2-3 kHz while output levels for the other systems remain above threshold up to 4-6 kHz. Also, the MCL appears to be limited by low-frequency energy for the ORTHO system, but by mid-to high-frequency energy for the other systems. The same observations hold for curves of individual subjects in the sloping group.

The average curves for the flat-loss group indicate that output levels of all systems are very similar. This is also true for individual subjects in the flat-loss group. Because of slight differences in thresholds, however, levels for the ORTHO system fall below threshold above 2 kHz for FM while they remain above threshold up to 6 kHz for IK. It is generally not clear what determines the MCL for these subjects.

2.2d.2 Normalized Functional Gain

The functional gains of the 6 systems for the two subject groups used in Experiment I (normalized to zero dB at 500 Hz) are presented in Figure 6 (characteristics for individual subjects are presented in Appendix 10). These curves represent the normalized functional gain provided for pure tones at the average 10% input speech levels. They did not vary with most comfortable level for the linear systems and for the COMCL system, but did vary with most comfortable level for the EL system. The curve for this system has been averaged over the 8 conditions in Experiment I[10]. The functional gains of the 10%, MA, and EL systems of subjects ED, ES and HS were equipment limited at high frequencies[11]. The functional gain of the ORTHO system is flat by definition. The functional gains of the 10% linear and EL compression systems, and of the OMCL linear and OMCL compression systems are again represented by single curves because of the similarity of these respective pairs of curves.

For the sloping-loss group, the different systems clearly provided differing amounts of high-frequency emphasis. In the region 2-4 kHz this ranged from 0 dB for the ORTHO system to roughly 30 dB for the OMCL and COMCL systems, 40 dB for the 10% and EL systems, and 60 dB for the MA system. This method of comparing systems is, however, deceptive. As noted previously, because of the MCL's chosen by subjects the systems differed primarily in the amount of low-frequency speech energy presented. For the flat-loss group the functional gains of the system are very similar. The ORTHO system provided the least high-frequency emphasis (0 dB) and the OMCL and COMCL systems provided the greatest (15 dB in the 2-4 kHz region).

2.2d.3 Compression Curves

Detailed descriptions of the compression curves of the EL and COMCL systems are presented in Appendix 12. Briefly, the compression range of the EL system extended from roughly

25 dB below the 10% speech level in each channel to roughly 15 dB above this level. The compression range of the COMCL system extended from roughly 15 dB below the 10% speech level in each channel to 15-20 dB above this level. The compression curve in this region for the EL system was determined from the measured equal loudness contours and the speech input levels used[12]. It is important to note that at the most comfortable input speech levels chosen for the EL system the 10% input levels were always more than 10 dB below the maximum level at which loudness balances were performed. The assumption made in section 1.2c.4 concerning the equal loudness contours at high levels thus did not significantly affect results obtained with the EL system. The compression curve of the EL system was adjusted to be within 2 dB of that required to restore equal loudness contours except at high frequencies where it was equipment limited[13]. The compression curve for the COMCL system over its compression range was linear with a compression ratio that was independent of the input speech level and was chosen as described in section 2.1b.

The compression ratios for the EL and COMCL compression systems (in the region above compression threshold and below 10% input speech level) averaged over the 8 conditions of Experiment I and for the two subject groups are presented in Fig. 7. Data for individual subjects are presented in Appendix 12.

For the sloping group both compression systems functioned as a linear amplifier (compression ratio=1) below 500 Hz. The compression ratio of the COMCL system rises to 2.5 above 2.5 kHz while that of the EL system rises to 5.0 by 3-4 kHz and then falls off at high frequencies (due to equipment limitations). For individual subjects compression ratios for the EL system in general increased to a maximum at high frequencies and then decreased because of equipment limitations. The maximum compression ratio for any subject at high frequencies was 7.0, and the frequency above which compression ratios were equipment limited was roughly 7 kHz for ED, 3 kHz for ES and 4 kHz for HS. Compression ratios for the COMCL system were very similar for individual subjects. However, above 3 kHz the compression ratio for ES was 2.0 while the compression ratios for ED and HS were both 3.0.

For the flat group the compression ratios for the COMCL system are similiar to those of the COMCL system for the sloping group. The compression ratio of the EL system for IK falls from roughly 4.0 below 2 kHz to 2.0 above 4 kHz, but for FM rises from 1.6 below 1 kHz to greater than 4.0 above 4 kHz. The large difference between the compression ratios of IK and FM was not initially expected given the similar audiograms of these subjects. The difference was, however, required to restore normal equal loudness contours and reflects the higher average speech input level chosen by IK (60 dB SPL versus 45 dB SPL for FM), and differences between the phon equal loudness contours of FM and IK (e.g., closer spacing of IK's contours in the 40-60 phon region at low frequencies and closer spacing of all FM's contours at high frequencies). The different compression ratios are thus directly traceable to differences between the loudness balances performed by the two subjects at low frequencies and between the dynamic ranges at high frequencies.

2.2d.4 Most Comfortable Levels

The most comfortable levels chosen by each subject for a given system varied little over the 8 material-talker-environment conditions in Experiment I. (These levels are presented in Appendix 8.) This variation averaged over all systems (excluding the EL system in which the input level was varied) and subjects was roughly ±2 dB. For individual subjects and systems the variation was typically less than ±7 dB. For the EL system the variation in input level ranged from 0 dB for subject FM to 20 dB for IK (The variation in output level was however much smaller because of compression). Overall, the MCL for PBNS's was 3 dB higher than the MCL for CVC's, the MCL for the male talker's material was 1 dB higher than the MCL for female talker's materials, and MCL's in quiet and noise were roughly equivalent (less than 1 dB difference). The standard deviation of the MCL's measured in Experiment I (determined from the difference between MCL's measured at the beginning and end of a session for a fixed system) was roughly 3 dB.

2.3 Speech Tests

The speech test materials used in Experiment I consisted of PB-50 words in nonsense sentences (PBNS's) and CVC nonsense syllables spoken by a male and female talker and presented in a quiet/anechoic (Q/A) or noisy reverberant (N/R) environment. These tests were designed to: 1) sample a range of listening situations 2) obtain scores that were neither too high nor too low and thus could differentiate between systems and 3) obtain information concerning consonant confusions.

2.3a Materials

PBNS's were formed by placing four words from the 1000 PB-50 words (Egan, 1948) in the sentence frame "The __ __ told the __ __." e.g. "The <u>cuff</u> <u>golf</u> told the <u>hole</u> <u>dive</u>." Forty lists of 15 sentences each were formed:

each list contained 50 words from 1 PB-50 list and 10 words chosen at random from the 1000 PB-50 words. (Lists 1-20 and 21-40 were both formed by consecutively using PB-50 lists 1-20.) The phonetic balance of each list thus approximated that of the PB-50 lists.

Items used in the CVC nonsense syllable test were of the form /ə/ - CVC where /ə/ is the unstressed schwa. The CVC's following /ə/ included all 1536 combinations (excluding slang words) of 6 vowels and 16 consonants. The vowels used were /i, a, u, I, ɛ, ʊ/, and the consonants were /p, t, k, b, d, g, f, θ, s, ʃ, v, ð, z, dz, tʃ/ in the initial and final position, /h/ in the initial and /ɣ/ in the final position. The CVC syllables were arranged into 32 lists of 50 items each such that in each list each vowel occurred at least 8 times and each initial and final consonant occurred at least 3 times. The distribution of phonemes in each list was thus almost identical but the overall distribution of phonemes was not equal to the distribution in English.

2.3b Recording Procedures and Acoustic Characteristics

All materials were recorded in a sound-treated room using a omnidirectional dynamic microphone (Electrovoice RE55) located roughly 4 inches in front of and 3 inches above each talker's mouth, a professional 2-track tape recorder (OTARI MX-5050), and low print/low noise magnetic tape (Scotch 208, 1.5 mil.). Materials were spoken by one male and one female talker without marked regional characteristics. They were instructed to speak naturally and to avoid over-enunciating items. They were trained for at least 4 hours. Items were read at a rate of one every 4-5 seconds. Errors were removed by backing the tape up and repeating either the whole list or the mispronounced item. After all recordings were made, levels of all items were measured (using an absolute value detector followed by 60 msec RC averaging). These measurements were used to make adjustments in level in the dubbing process in order to equate the average level of each CVC and PBNS list. Level variation within a list and within each PBNS was not equalized. The level variation of syllables in the CVC lists and of sentences within the PBNS lists was less than ±4 dB. Also, the last word in the female PBNS's was typically 11 dB below the level of the first word and the last word in the male PBNS's was typically 4 dB below the level of the first word. The overall RMS levels of the lists were determined by first segmenting items (PBNS's or CVC's) from the lists using a digital computer. Items of each kind were then abutted until a 24 second segment of continuous CVC's or PBNS's was obtained. The average RMS level of these segments is reported.

Reverberation was added to half of the materials by playing them through a loudspeaker into a reverberant room and rerecording using a microphone located 6 feet from the loudspeaker. The room was a small classroom (20x16x9 ft.) with no windows or carpeting. It had plasterboard walls, a blackboard along one of the walls, and acoustic ceiling tiles. The average reverberation time of the room at 500, 1000 and 2000 Hz was 0.6 sec. The frequency response of the loudspeaker-room-microphone system (measured with bands of noise) was flat within 3 dB from 160 Hz to 8 kHz. When the reverberant lists were copied onto one track of the tape to be used to perform the tests, cafeteria noise provided by the Veterans Administration was copied onto the other track. In all tests involving the reverberant recordings this noise was added to the reverberant speech 10dB below its RMS level. This noise consisted predominately of speech babble and had previously been processed by a compressor-limiter (by the V.A.) to reduce level variation.

The 10% cumulative levels of the male and female CVC and PBNS material and of the cafeteria noise were determined (relative to the long-term RMS level of these materials) using the 16 detectors in the speech processing system and 24 seconds of each type of material. These levels are presented in Appendix 6 and have many dips and peaks as a function of frequency but are not significantly different from each other except for a 2-8 dB increase in level for the CVC's relative to the sentences above 2 kHz. The difference between the average 10% levels of all materials and the 10% levels of Dunn and White used to determine the characteristics of the 10% system is less than 3 dB from 300-2000 Hz. Outside this region our levels exceed Dunn and White's by 4-13 dB . The 10% and 90% cumulative levels of the cafeteria noise are also presented in Appendix 6. These levels differed by 7 to 10 dB and are similiar to the average 10% levels of all materials.

2.3c Responses

Responses for all words in one PBNS list were obtained by first presenting the list and having the subject write down the first (or last) two key words of each PBNS. The same list was then presented again and the subject wrote down the other two words of each PBNS. The choice of whether the first or last two key words were written first was random. This procedure was required because it was found to be difficult to remember and write down four unrelated words while it was easy to remember and write down two.

Responses for CVC's were written for normal hearing subjects (who had been trained in transcription). Impaired subjects repeated the CVC's after they heard them and these responses were then immediately transcribed by the experimenter. Both subject and experimenter were seated in a sound-proof room and only CVC responses were accepted. An attempt was made to limit responses and make transcription easier by describing the phonemes used in the CVC's before each test and giving subjects a list of these phonemes (along with key words) to refer to during each CVC test. This process was checked in preliminary experiments by having subjects verify the transcription of each item immediately after their oral response using the above list of CVC's and key words. The number of transcription errors was insignificant (maximum=3 out of a possible 150).

For the impaired listeners, ratings of the quality or naturalness of each list were obtained after each list was administered using a scale with eight equally spaced levels labeled "terrible, very poor, poor, moderate, fair, good, very good, and excellent." These ratings were converted to scores between 0 and 100 by assigning 100 points to "excellent", 0 points to "terrible" and spacing all other qualifying adjectives equally over the range 0 to 100.

All PBNS's and CVC's were delivered at a rate that allowed adequate time for responses and was comfortable for each subject. The rate was slowed, if necessary, by having the experimenter turn the tape recorder on and off.

2.3d Scoring

Lists were scored in terms of percent words, phonemes, initial consonants, final consonants and vowels correct. All words (including CVC's, VC's and CV's) were counted as 3 points if totally correct. Vowels, consonants, and consonant clusters were counted as 1 point if totally correct (/r,l,w/ were treated as consonants). A consonant that was expanded to a cluster or a consonant cluster that was only partially correct was counted as 1/2 point. Adding an initial consonant to a VC or a final consonant to a CV was equivalent to a consonant error. Finally, if the final consonant of one word in a PBNS was the same as the initial consonant of the next and only one consonant was recorded (e.g. "HIGH DOT" instead of "HIDE DOT") or if an initial or final consonant from one word was attached to the preceding or following word (e.g. "WHY DILL" instead of "WIDE ILL") then both consonants were counted as 1 point.

2.4 Results

2.4a Variability of Scores and of Quality Ratings

The equivalence of the 7 lists for each of the 8 talker-material-environment conditions of Experiment I was initially assured by phonetically balancing each list, recording lists in the same session, equalizing the average level of each list, and processing lists identically when dubbing and adding noise and reverberation. The variability of phoneme scores obtained with these lists was measured using three normal subjects who listened with added noise and low-pass filtering (2 or 3 kHz) that was sufficient to reduce scores to roughly 30-70% . The difference between average scores of the three subjects obtained on individual lists for one condition and average scores obtained on all 7 lists for one condition was compared to the expected binomial variance of these average scores. It was found that differences for 11 of the 46 lists exceeded 2 standard deviation bounds and only one exceeded 3 standard deviation bounds. The absolute difference for lists which exceeded 2 standard deviation bounds was roughly 6 points and the maximum differences were 10 and 6 points. The effect that these differences could have on the results of this study is small because these differences are small and the list used with each subject/system/condition combination was randomized. With regard to learning and training, in all 8 of the sessions (except the first) in which intelligibility of 7 lists for one condition was measured, no learning or training effects (as indicated by a systematic change in scores) were evident past the first practice list. In the first session stable scores were obtained after 2 lists were presented.

The variability of phoneme scores and quality ratings of the lists was measured for impaired listeners by administering two of the 7 lists using the same system in each session: one at the beginning and one at the end of the session. The differences between phoneme scores of these lists indicated that the variance of these scores did not exceed the expected binomial variance (out of 30 differences in scores none exceeded 2 binomial standard deviations). These comparisons also demonstrated that within-session learning or fatigue effects were minimal (the average difference between scores was less than one point). The differences between quality ratings indicated that the standard deviation of these ratings was 14 points which corresponds roughly to one increment in the scale used (e.g. "very poor" to "poor" or "fair" to "good"). Furthermore, the average difference between quality ratings obtained at the beginning and end of sessions was small (5 points).

2.4b Relationship Between Word or Syllable and Phoneme Scores

It was found that word scores for the PBNS's and the CVC's could be calculated from phoneme scores by the formula,

$$I = 100(P/100)^{N} \quad , \qquad (1)$$

where I represents the percent word or percent syllable correct score, P represents the percent phoneme correct score and N is an exponent between 1 and 3. This relationship is illustrated in Figures 8a-8d in which scores for all lists administered to both normal and impaired listeners are plotted. Good fits[15] were obtained with the values of N given in these Figures. For the CVC lists, N=3 for both normal and impaired listeners; for the PBNS lists, N=2.5 for the normal listeners and N=2.3 for the impaired listeners. These values of N suggest that phonemes in the CVC's were identified independently but that the identification of a phoneme within a word in a PBNS was dependent on the identification of the other phonemes in the word. The scatter diagrams also indicate that word formation was not a problem for words in the PBNS's except for tests with very low phoneme scores. Results similiar to the above were also obtained by Fletcher and Steinberg (1930) and Barfod (1976).

2.4c Overall Comparisons Between Conditions

2.4c.1 Phoneme Scores

The phoneme scores obtained by the normal hearing subjects listening in quiet (averaged over the 7 lists for each condition) and by the impaired subjects (averaged over lists used with the 6 systems for each condition and over subject groups) are presented in Table 2. Scores for the flat-loss group of impaired subjects for the PBNS's in the N/R environment are not presented because this condition proved too difficult for these subjects and scores were not obtained.

Under all conditions scores for the normals are highest (75-98 points) followed by scores for the sloping group (38-73 points) and then by scores for the flat group (26-65 points). The good performance of the normals is expected and the lower scores of the flat-loss group are consistent with the poorer low-frequency hearing of subjects in this group (see Figure 2). An indication of the severity of the hearing loss of the impaired subjects is given by the relatively low maximum scores obtained with these materials (73 and 65% phonemes correct corresponding to roughly 40 and 30% words correct).

In the Q/A environment for the sloping group and for the normal subjects scores for the PBNS's and CVC's were roughly equivalent. In the N/R environment and for the flat group scores for PBNS's were lower than scores for CVC's. This may have been caused by a reduction or absense of acoustic cues of words in the PBNS's (see Klatt and Stevens, 1973, for a discussion of this phenomenon) and by greater importance of low-frequency speech cues in the PBNS's.

Scores with the material spoken by the female talker were higher than scores for material spoken by the male talker for the flat-loss group (52% vs 43%), and roughly equivalent to scores for the male material for the sloping-loss group (61% vs 59%), and for the normal group (91% vs 94%). We presently do not understand the cause of these differences[16]. Furthermore, although these talkers are referred to as male and female the differences should not solely be attributed to the sex difference between talkers because of the use of only two talkers.

As expected, all scores for the N/R condition were lower than scores for the Q/A condition. The decrease in scores was greater for the PBNS's than for the CVC's (25 vs 11 points) and greater for the impaired than for the normal subjects (20 vs 12 points). The greater effect of noise and reverberation on scores for the PBNS's and for the impaired listeners may be related to the greater importance of low-frequency speech cues for the PBNS's and to the reduced auditory area of the impaired subjects.

2.4c.2 Quality Ratings

The quality ratings, averaged over systems, produced by the sloping and flat-loss groups are presented in Table 3. Ratings are higher for the sloping group than for the flat group (55 vs 42 points), and higher for the CVC's than for the PBNS's (60 vs 42 points). Ratings for the female talker's material are higher than ratings for the male talker's material for the flat-loss group (45 vs 39 points) and equivalent to ratings for the male talker's material for the sloping-loss group (55 vs 55 points). The relative ranking of conditions based on these ratings and on phoneme scores is equivalent. Also, many of the differences between ratings and phoneme scores obtained for different conditions are also roughly equivalent.

2.4c.3 Consonant and Vowel Scores

Average initial consonant, final consonant, and vowel scores for all lists administered to the impaired subjects and lists administered to the normal subjects are presented in Appendix 13 (Tables A13.12 - A13.14). Initial consonants were more intelligible than final consonants under all conditions. The average difference between these scores was roughly 11 points for the sloping-loss group, 7 points for the flat-loss group, and 6 points for the normals. For all subject groups, noise and reverberation caused a greater decrease in final than in initial consonant scores. In the N/R condition the difference between initial and final consonant scores increased to 17 points for the sloping-loss group, to 12 points for the normals, and to roughly 9 points for the flat-loss group[17]. The difference between average vowel and initial consonant scores was 11 points for the sloping-loss group, -7 points for the flat-loss group, and 1 point for the normals. For the sloping-loss group, the difference between vowel and initial consonant scores was much higher for the CVC's than for the PBNS's (20 points vs 3 points)[18].

Higher scores for initial than for final consonants in CVC's is a common finding and occurred for all subjects. Higher scores for vowels than for initial consonants is also a common finding and occurred for the sloping-loss and normal subject groups. It did not however, occur for the flat-loss group where vowel scores fell below initial consonant scores.

2.4d Comparisons Between Systems

2.4d.1 Phoneme Scores

Average phoneme scores for the flat and sloping-loss groups and the 6 systems are presented in Figures 9 - 12 and Table 4. Individual results are presented in Appendix 13. We emphasize comparisons based on these scores because they are closely related to the inherent intelligibility of individual speech sounds (Boothroyd, 1968), because they were found to be monotonically related to percent word correct scores, and because they have reduced variance relative to percent word correct scores (Boothroyd, 1967; Barfod, 1976). The standard deviations given for these scores in Figures 9 - 12 were calculated by assuming that the scores were binomially distributed and that the probability of identifying a phoneme correctly was 0.5.

2.4d.1a Linear Systems

Scores for the three linear systems with high-frequency emphasis (OMCL, 10%, MA) were roughly equivalent and significantly superior to scores for the ORTHO system. This superiority was greatest for the sloping group, as would be expected. The range between scores averaged over all conditions obtained with the OMCL, 10% and MA systems was only 2 points for the sloping-loss group and 3 points for the flat-loss group. Also, the difference between overall average scores with the above 3 systems and with the ORTHO system was 29 points (68% vs 39%) for the sloping-loss group and 12 points (53% vs 41%) for the flat-loss group.

The above relationships between scores were observed under all test conditions for all subjects except IK. She obtained roughly equivalent scores with all linear systems. However, the functional gain and 10% output levels of her linear systems were also roughly equivalent and thus the observed similarity in scores was not surprising. The inferior scores obtained with the ORTHO system for the other subjects are consistent with the reduction of speech energy at high frequencies for the ORTHO system. For all subjects in the sloping-loss group and for FM, speech energy above 2-4kHz was below threshold with the ORTHO system while speech energy was above threshold up to 4-8 kHz with the other linear systems (see Figure 5).

The equivalence of scores for the OMCL, 10% and MA systems is understandable for subjects in the flat-loss group because the 10% output levels of the systems for these subjects were roughly equivalent. The equivalence of scores for the sloping-loss group is however somewhat surprising given the large differences between the low-frequency output levels of these systems for this subject group. As shown in Figure 5, levels in the region 250-500 Hz differed by roughly 25 dB. These levels are all, however, more than 20 dB above threshold and less than roughly 70 dB SPL. The equivalence of the 3 systems is thus consistent with the idea that low-frequency speech energy must be audible (e.g. 10% level 20 dB SL), but need not be too intense, to contribute to overall intelligibility.

2.4d.1b Compression Systems

Scores with the COMCL compression system were generally greater than or equal to scores obtained with the EL compression system. The overall difference in group average scores was small (2 points) for the sloping-loss group and slightly larger (6 points) for the flat-loss group. The advantage of the COMCL system was greatest (22 points) in quiet with the female sentences for the flat-loss group. Individual results concerning these observations were generally consistent with the group average results.

These results indicate that it is not necessary to restore equal loudness contours to normal. Although this finding is not fully understood, one important factor in the poorer performance of the EL system for the flat-loss group may be the higher compression ratios of this system at low-mid frequencies.

2.4d.1c Linear Amplification versus Compression

Scores obtained with the best linear system were usually greater than or equal to scores obtained with the better compression system. The overall difference in scores between these systems was roughly 9 points for both subject groups. This difference was smaller in quiet (5 points) than in noise (14 points) and in quiet it was smaller for the female talker (1 point) than for the male talker (10 points). Furthermore, for the flat-loss group the better compression system was slightly superior to the best linear system (47% versus 40%) for female PBNS's in quiet.

It is important to determine why compression performed significantly better than linear amplification only for the flat-loss group in quiet with the female PBNS's. The better performance of compression under this condition is probably related to the greater word-to-word level variation in the female PBNS's, the greater loss at low frequencies for the flat-loss group, and the equalization of word-to-word level variation caused by compression. In normal sentences the level of words varies naturally and depends on factors including stress, function, and position in a sentence. For the female PBNS's the levels of the key words decreased towards the end of each sentence and the last key word in each sentence was, on the average, 11 dB below the level of the first key word. For the best linear system this reduction in level would be expected to cause a significant decrease in the audibility of speech components at low-mid frequencies for the flat-loss group, but not for the sloping-loss group (see Figure 5). The results obtained are consistent with this prediction (with the best linear system scores for the last two key words in the PBNS's were 12 points lower than scores for the first two key words for the flat-loss group but only 2 points lower for the sloping-loss group)[19]. Furthermore, it is likely that compression compensated for this level variation by providing more gain for low-level words. Evidence for this compensation is that the increase in scores with the COMCL system versus the best linear system was small for the first two key words in the sentences (2 points) and much higher for the last two words (12 points). Thus, if there had been no reduction in level of the last two words, scores with compression might have increased only two points. The equalization of levels by compression would not be expected in to be as effective in noise because the noise controls the gain of the compressor for low level signals. Also, the equalization of levels would not be expected to be important for the male PBNS's or the CVC's because the range of levels encountered in these materials was much less than the range for the female PBNS's (the average range was 4dB for the male PBNS's and 6dB for the CVC's).

Scores obtained with the better compression system were consistently superior to those obtained with the ORTHO linear system. The difference in scores (averaged over all conditions) was 21 points for the sloping and 4 points for the flat group.

2.4d.2 Quality Ratings

Average quality ratings for the flat and sloping-loss subject groups made in quiet and noise are presented in

Table 5. Ratings of individual subjects are presented in Appendix 13.

For the sloping-loss group quality ratings are roughly consistent with phoneme scores. The main inconsistency is that the OMCL system is rated higher and the MA system is rated lower than indicated by phoneme scores. Although the difference in phoneme scores between these two systems is less than 1 point when averaged over all conditions, the difference in quality judgments is 14 points (roughly that between "fair" and "good"). It is interesting to note that of the three linear systems which provided roughly equivalent phoneme scores (OMCL, 10%, MA), subjects preferred the system with the least amount of high-frequency emphasis (OMCL). This is reasonable because the boost to high frequency energy provided by the MA and 10% systems emphasized fricative sounds to such an extent that some subjects commented that these sounds were too loud and limited the level at which they could comfortably listen. Also, one subject complained that speech processed by these systems was "irritating".

For the flat-loss group, quality ratings for subjects IK and FM differed. IK tended to use a wider range of ratings than FM and the average results of the flat-loss group are thus dominated by her ratings. IK's ratings indicate that the ORTHO, OMCL, and 10% systems are equivalent. Ratings of the MA system were generally lower (13 points) than ratings of the other linear systems. Also, ratings of the two compression systems were, on the average, equivalent and roughly 24 points below ratings of the ORTHO, OMCL, and 10% systems. For FM ratings of the MA, 10%, and OMCL systems were roughly equivalent. and slightly greater (6 points) than ratings of the ORTHO system. Ratings of the two compression systems were similar and slightly lower (6 points) than the ratings of the MA, 10%, and OMCL systems.

2.4d.3 Consonant Confusions

Consonant confusion matrices were formed for the flat and sloping groups for the ORTHO, OMCL, COMCL and EL systems by combining results for all consonants obtained for the CVC lists in the Q/A condition. The resulting matrices contain roughly 37 and 25 responses per consonant respectively for the sloping and flat groups, and are given in Appendix 14. These matrices were analyzed in terms of information transfer (Miller and Nicely, 1955) and the results of this analysis are presented in Table 6.20. In all matrices a bias toward responding /v/ instead of /ð/ and /f/ instead of /θ/ was observed. This bias was roughly the same for all systems.

For the sloping-loss group error patterns for the linear systems are similar to the patterns reported by Miller and Nicely (1955) with low-pass filtering and by Bilger and Wang (1976) for subjects with sensorineural losses and sloping audiograms. If features not common to all studies (e.g. nasality, sibilation) are omitted, then the feature voicing was most salient followed by frication, duration and place for all studies. The equivalence of error patterns for subjects with sloping audiograms and for normals with low-pass filtering evident in our results was also noted by Wang et al. (1977). It can be explained by the inaudibility of high-frequency energy for subjects with sloping-losses. Error patterns for the ORTHO system were similar to those obsereved by Miller and Nicely for low-pass filtering at roughly 600 Hz, while those of the OMCL system were similar to those for low-pass filtering at 2500 Hz. This difference is consistent with the greater audibility of high-frequency energy for the OMCL system indicated by Figure 5[21]. Confusion patterns obtained with the compression and linear systems were similar. The effect of compression can generally be described in terms of the audibility of speech energy in different frequency regions. For example, changes in error patterns with compression were minimal between the OMCL and COMCL systems which had very similiar 10% output speech levels. Also, changes in error patterns between the OMCL and EL system were similar to changes observed by Miller and Nicely with additional low-pass filtering. This is consistent with the lower 10% output levels of the EL compression system in the region 1-3 kHz.

For the flat-loss group error patterns for the linear systems are roughly similar to patterns reported by Miller and Nicely (1955) with high-pass filtering, but the similarity of patterns is not nearly as good as for the sloping-loss group and low-pass filtering. Bilger and Wang (1976) also found this to be true for subjects with flat sensorineural losses. If features not common to all these studies (e.g. sibilance, high) are omitted, then duration and voicing are most important while place and frication are least important for all studies. The rough equivalence of error patterns for subjects with flat losses and normals with high-pass filtering is not easily explained on the basis of audibility of speech energy in different frequency regions. It may be related to supra-threshold perceptual distortions that are more severe for the flat-loss group. The error patterns for the two compression systems for the flat-loss group are similar to each other, but they differ from the error patterns of the linear systems. The main difference is that the relative transmitted information for the features duration and place are lower with compression (74 versus 41% and 44 versus 22% respectively)[22]. This change makes the error patterns of the compression systems more similar to those obtained with low-pass filtering.

Errors for both subject groups in all confusion matrices were primarily caused by errors in the feature place. This is indicated by the low percent transmitted information for this feature in Table 6 and by the orderly pattern of responses in the confusion matrices. Furthermore, this feature was perceived best with the OMCL linear system. These results indicate that any more complicated speech processing system must focus primarily on making the acoustic cues for place (eg. short-term spectra, formant transitions) more discriminable. For the amplitude compression systems studied percent relative transmitted information for place fell. It is not known whether this was caused by a lack of training with compressed materials or by a blurring of important acoustic cues for place caused by reduced differences between the spectra of short speech segments with compression.

3. Experiment II

The purpose of Experiment II was to compare the ORTHO, OMCL and COMCL systems using standard tests. The ORTHO system was included in this experiment because it is a standard reference, the OMCL system was included because it is representative of the linear systems with high frequency emphasis, and the COMCL system was included because it was the better of the two compression systems in Experiment I. The same 5 subjects used in Experiment I were used in Experiment II.

3.1 Speech Tests

The speech tests used were 1) CID W-22 word lists: 50 words per list (Hirsh et al., 1952), 2) SPIN Test lists 50 words per list - 25 words at the end of low predictability (PL) sentences and 25 words at the end of high predictability (PH) sentences (Kalikow et al., 1977), and 3) Harvard Sentences: 10 sentences per list with 5 key words per sentence (IEEE, 1969). Versions of these materials that had been recorded and used previously by others (Hirsh et al, 1952; Kalikow et al, 1977; Williams and Hecker, 1968) were used. Each test had been recorded by a male talker in a quiet/anechoic background. All materials were administered in quiet; the SPIN test was also administered in a background consisting of the 12-voice babble provided on the SPIN test tape at a speech to babble ratio of 10 dB. The RMS input level of the SPIN test lists (determined from the provided calibration tone) was at the same RMS input level used for the materials in Experiment I. The input levels of the other tests were adjusted such that the maximum level observed in each of the 16 channels was at least 3 dB below the clipping level of the system. Detailed measurements of the frequency-level distribution of the speech materials were not made. However, the decrease in level between the first and last key word of the Harvard Sentences and between the first content word and last word of the SPIN test sentences was measured. The average decrease in level was 2 dB for the Harvard Sentences and 3 dB for the SPIN sentences.

3.2 Methods and Procedures

All tests were administered in 2-3 sessions which lasted 2-3 hours each and which followed the last session of Experiment I. Tests were administered in the following order: SPIN-Quiet, SPIN-Noise, CID W-22, Harvard sentences. The order in which systems were tested was randomized for each subject and test. All lists were administered at each subject's MCL (determined as in Experiment I) and the first list of each type was used for practice and to determine presentation levels for the remaining lists of that type. Feedback was not provided, all responses were written, and masking was used as in Experiment I. Each system was tested with 1-CID W-22 word list in quiet, 1-SPIN Test list in quiet, 1-SPIN Test list in noise, and 2-Harvard Sentence Lists in quiet. Word and phoneme scores were obtained as in Experiment I using the 5 key words in each Harvard Sentence and the designated words in the other tests.

3.3 Results and Discussion

3.3a Most Comfortable Levels

The most comfortable levels chosen in Experiment II are presented in Appendix 15[23]. The differences between the MCL's chosen in Experiment II and in Experiment I were generally small. Averaged over all subjects, tests, and systems the levels in Experiment II were roughly 2 dB above the levels used in Experiment I. Furthermore, for all tests, systems, and subjects (excluding ES) all levels used in Experiment II were within 6 dB of the levels used in Experiment I. Subject ES used levels that were higher than those used in Experiment I by 1-11 dB (average increase = 6 dB).

3.3b Speech Tests

Phoneme scores for Experiments I and II are presented in Figures 13-17 and Table 7. (Scores for individual subjects in Experiment II are presented in Appendix 16.) The 2 standard deviation bounds in these figures were calculated as described previously and are averaged over both experiments.

3.3b.1 Overall Level of Scores

Scores in Experiment II were generally higher than scores in Experiment I and in many cases were very near 100% correct. This reflects the greater amount of contextual information in certain of the Experiment II tests and the fact that many words in these tests were enunciated very carefully and perhaps overemphasized. The relationships between scores obtained with the different tests were different for the two subject groups and in quiet and noise. In general, however, scores were highest for the PH SPIN sentences and the Harvard Sentences, and lowest for the PBNS's.

3.3b.2 ORTHO versus OMCL

For each test in Experiment II scores with the OMCL system were higher than scores with the ORTHO system. The difference in scores averaged over all tests was roughly 13 points for both subject groups. This difference was generally smallest for tests with scores above 90% correct (Harvard Sentences and words in PH SPIN sentences) but otherwise varied unsystematically from test to test. Furthermore, this difference was generally larger for the sloping group than for the flat group, as expected. Results for individual subjects are similar to group average scores except for subject IK for whom scores for the ORTHO and OMCL systems were similar. As noted previously, however, these systems were also functionally similar for IK. In general the relation between the OMCL and the ORTHO systems in Experiment II differs little from that of Experiment I. This confirms the superiority of the OMCL system over the ORTHO system, particularly for the sloping group.

3.3b.3 COMCL versus OMCL

In Experiment II the COMCL system was roughly equal to the OMCL system in quiet, but slightly inferior in noise. The difference in scores was 0.5 points for both subject groups in quiet and -6 points in noise. The best performance of the COMCL system in quiet (3-4 points above the OMCL system) occurred for the low predictability SPIN sentences but was not statistically significant. Similar results occurred for individual subjects. A comparison of these results to those of Experiment I indicates that the performance of the COMCL system relative to that of the OMCL system was slightly better in Experiment II (by roughly 4 points). The difference between scores with the COMCL and OMCL system averaged over all subjects was -5 and -9 points for the Q/A and N/R conditions in Experiment I, and 0.5 and -6 points in quiet and noise in Experiment II. Whatever factor was important in causing the improvement in the relative performance of the COMCL system in Experiment II, it was probably not level variation of words in the Experiment II tests. This variation was roughly equivalent to that in the CVC's or in the male PBNS's used in Experiment I. The improvement was small however. These results thus confirm those of Experiment I.

3.3b.4 Word Scores

Group average word scores are presented in Table 8 and word scores for individual subjects are presented in Appendix 16. Comparisons between systems made using these scores generally confirm the phoneme scores, but exhibit somewhat greater variability. The relatively large increase in scores with the COMCL system relative to the OMCL system (9 points) for the sloping-loss group and the CID W-22 test and PL SPIN sentences in quiet largely reflects the performance of subjects ED and ES. Scores for subject ED increased by 23 points and scores for ES increased by 9 points while scores for HS decreased by 4 points. The cause of the larger increase for ED (which corresponds to an increase of 6 points in terms of phoneme scores) is unknown.

3.3b.5 Quality Judgments

Group average quality judgments for all Experiment II tests are presented in Table 9 and individual results are presented in Appendix 16. For the sloping-loss group in quiet the OMCL and COMCL systems are roughly equivalent and better by 30 points than the ORTHO system. In noise the performance of the COMCL relative to the OMCL system drops 30 points and the OMCL system is best. For the flat-loss group in quiet the COMCL system performs better than the OMCL and ORTHO systems but in noise the ORTHO system performs better than the COMCL and OMCL systems.

4. Experiment III

The primary purpose of this experiment was to investigate the effect of reduced input levels on speech intelligibility. Four of the five impaired subjects were able to participate in this test. Each was tested with the COMCL system and the linear system which led to best performance in Experiment I (ORTHO for IK, OMCL for the other subjects). Only the quiet/anechoic female CVC lists were used in Experiment III.

4.1 Methods and Procedures

Each subject was tested in one session with both systems. The input speech levels were set at 0, 4, 8, 16 and 24 dB below the levels corresponding to each subject's MCL for each system. The order in which systems and levels were tested was randomized over subjects. In general procedures were the same as in Experiment I, however, the MCL was determined only once for each system and quality judgments were not obtained.

4.2 Results and Discussion

The MCL's in Experiment III were similar to those of Experiment I (less than 4 dB difference) except for the linear systems of IK and ES for which the MCL increased by 7 and 14 dB respectively. Phoneme scores obtained at each subject's MCL were slightly higher (3 points) than in Experiment I. For individual subjects this increase ranged from -6 to 10 points. The average difference between scores with the compression and linear systems in Experiment III (-2 points) was very similar to the difference in scores for the female CVC's in Experiment I (-1 point).

The phoneme scores for Experiment III are presented in Figure 18. As can be seen, the linear and compression system are roughly equivalent at MCL but the compression system always provides superior scores at lower levels. The average difference in scores with compression and linear amplification at 16 and 24 dB below MCL is 14 points. This difference was greatest for subjects IK and ED who had the greatest losses at 500, 1000 and 2000 Hz.

These results demonstrate the ability of compression to compensate for reduced input levels. The compression system provided increased gain for low level inputs while the linear system provided fixed gain for all inputs; thus the compression system performed better at low levels. These results are consistent with the increased intelligibility with compression of low-level words in the female sentences in quiet for subjects IK and FM that was found in Experiment I.

5. Discussion

5.1 Summary of Present Study

Two 16-channel compression systems and four linear systems were evaluated using 2 subjects with flat and 3 subjects with sloping sensorineural losses. Initial tests on all systems were performed using PB-50 words in nonsense sentences (PBNS's) and CVC nonsense syllables. These materials were spoken by a male and a female talker and presented in both a quiet/anechoic and noisy/reverberant environment. After the best linear and better compression system were determined from these initial tests they were evaluated along with another linear system having a standard orthotelephonic response. This second evaluation utilized the CID W-22 test (Ira Hirsh version) in quiet , the SPIN test in quiet and noise, and Harvard sentences in quiet. All of the above materials were presented at each subject's most comfortable level. Supplementary tests were also performed at reduced input levels using the best linear and better compression system from the initial tests.

One of the two compression systems (EL) was adjusted to restore normal equal loudness contours. Its characteristics were determined on the basis of an extensive set of loudness balances. There were three factors limiting the accuracy with which this system restored normal equal loudness contours: variance of the loudness balances, equipment limitations, and variation in the sound pressure levels delivered to the eardrum. Although the errors introduced by these factors were generally small in dB, they may have been significant for these subjects in regions where their dynamic range was small. The compression ratios in the EL system depended on the loudness balances and on the most comfortable levels used and varied from subject to subject (especially between the two subjects with flat losses). The other compression system (COMCL) provided linear amplification at low frequencies and limited compression ratios (<3) at high frequencies. The equalization of this system was subject adjusted to present individual bands of compressed speech at a most comfortable level. The compression ratios for this system varied little over subjects.

Of the four linear systems, one (ORTHO) provided flat functional gain. The second (MA) mirrored each subject's audiogram. The third (10%) restored normal loudness to 10% cumulative levels of speech, and the fourth (OMCL) amplified bands of compressed speech presented in isolation to a most comfortable level. The high-frequency emphasis provided by the three fitted linear systems ranged from roughly 30-60 dB for the sloping-loss subject group and 10-20 dB for the flat-loss group. At the most comfortable levels chosen by subjects in the sloping-loss group the output 10% cumulative levels at mid-high frequencies were roughly equivalent but the levels at low frequencies differed by roughly 25 dB. For all subjects the output 10% cumulative levels for the COMCL compression and OMCL linear system were roughly equivalent. This was also true for the EL compression and 10% linear system.

Under all conditions, the linear systems with high-frequency emphasis yielded nearly equivalent scores which were significantly higher than scores for the ORTHO system. This was more pronounced for subjects with sloping-losses (29 points) and for one of the subjects with a flat loss (24 points). For the other subject with a flat loss the frequency-gain characteristics of the four systems were roughly equivalent and all scores were roughly equivalent. Quality judgements made by subjects in the the sloping-loss group indicated that of the 3 systems with equivalent scores, subjects preferred the system with least high-frequency emphasis (OMCL). For the subjects with flat losses, quality ratings of the linear systems did not differ significantly.

The COMCL system performed better than or equal to the EL system in terms of both scores and quality ratings. However, the overall difference in scores was small for the sloping-loss group (2 points) and only slightly larger for the flat-loss group (6 points). The advantage of the COMCL system was greatest (22 points) for the subjects with flat losses in quiet with the PBNS's spoken by the female talker.

The best linear system typically was superior to the better compression system (COMCL) in terms both of scores and quality judgements. The advantage in scores was greater in noise (14 points) than in quiet (5 points) and in quiet it was greater for the male talker (10 points) than for the female talker (1 point).

The best system tested for each subject failed to provide normal speech perception for the impaired listeners studied. For example, while normal subjects obtained scores of roughly 95% on the CVC's and PBNS's presented in quiet, subjects in the sloping group obtained maximum scores of only 80% . Those in the flat group obtained maximum scores of roughly 60% . Furthermore, in noise, the difference between scores for normal and impaired listeners increased by roughly 10 points.

Consonant confusions were analyzed for CVC's presented in quiet and processed by the ORTHO, OMCL, COMCL and EL systems. For the sloping-loss group the error patterns for the linear and compression systems were similar and resembled those of normals with low-pass filtering. The differences between the error patterns with the linear and compression systems resembled differences obtained for normals with different amounts of low-pass filtering and were consistent with variations in the audibility of high-frequency speech elements. For the flat-loss group the error patterns with the linear systems were, however, less structured and more similar to those of normals with high-pass filtering while the error patterns with the compression systems were more similar to those of normals with low-pass filtering. The cause of this result is presently not known.

The only condition in which a compression system (COMCL) performed significantly better than the best linear system (47% vs 40%) was in quiet with female PBNS's for both subjects with flat losses. This appears to be related to the level variation of words in the female sentences and the greater loss at low frequencies for these subjects. The average level of the final word in these sentences was 11 dB below the average level of the initial word. This reduction in level was typically associated with reduced scores for the linear system but not for the compression system. This effect was further explored by testing at reduced input levels. At input levels corresponding to each subject's most comfortable level there was no difference between scores obtained with compression and linear amplification. At input levels that were 16-24 dB below these levels scores with compression were 3-38 points higher than scores with linear amplification.

5.2 Comparison to Previous Research on Linear Amplification

The importance of providing high-frequency emphasis which has been examined in this study and the recent studies of Barfod (1972), Pascoe (1975), and Skinner (1976) is illustrated in Table 10. (Detailed reviews of these studies and comparisons of these studies to previous research are presented in Braida et al., 1978.) As can be seen from this table, systems with high-frequency emphasis perform significantly better than systems with a flat response both for subjects with flat and sharply sloping-losses. For subjects with sharply sloping-losses the increase in scores with high-frequency emphasis is higher (29 points) in the present study than in the studies of Barfod and Skinner (19 and 12 points). This difference may be related to our use of subjects with greater losses. The increase in scores with high-frequency emphasis for subjects with flat or gently sloping audiograms is almost identical (11 and 12 points) in the present study and in the study of Pascoe.

The present study found that the three systems with high-frequency emphasis had similar performance for subjects with sloping-losses. These results are similar to those of Skinner (1976) and Barfod (1972). Both studied systems with differing amounts of high-frequency emphasis using subjects with sharply sloping-losses. Skinner, however, varied presentation levels over a wide range (including each subject's most comfortable level) while Barfod maintained a constant unprocessed speech level (65 dB SPL). These studies demonstrated little change in scores for variations in high-frequency emphasis of 22-44 dB and of 30-40 dB. Skinner also found, as did the present study, that at each subject's most comfortable level, systems with various amounts of high-frequency emphasis differed primarily in the presentation level of low-frequency speech elements.

All recent studies have demonstrated the inadequacy of linear amplification as a means of providing normal speech perception for impaired listeners. Under certain conditions in the present study and in the studies of Barfod, Skinner, and Pascoe normal subjects would obtain essentially perfect scores. Under these same conditions with the best linear systems studied, subjects with sharply sloping losses obtained maximum scores of between 70% and 80% and subjects with flat or gently sloping losses obtained maximum scores of roughly 70% and 75% .

5.3 Comparison to Previous Research on Multiple Channel Syllabic Compression

The major previous studies of multiple channel syllabic compression are those of Villchur (1973), Yanick (1976) and Barfod (1976). Detailed reviews of these studies and comparisons of these studies to previous research are presented in Lippmann et al., 1978. The results of the present study and of these studies concerning comparisons of compression and linear systems are summarized in Table 11. These comparisons are based on results obtained under conditions common to all studies in which 1) the long-term level of speech material was held constant before processing and there was minimal word-to-word level variation in this material 2) subjects had, on the average, moderate losses 3) compression characteristics were generally adjusted in an attempt to restore normal equal loudness contours, and 4) minimal training was provided with compressed materials. The results of the present study are presented twice in Table 11, once using the ORTHO and once using the OMCL system as a linear reference to evaluate the effect of compression.

As can be seen in Table 11, in Villchur's and Yanick's recent studies and in the present study when the ORTHO system is used as a linear reference, scores increase with compression. On the other hand, in Barfod's study and in the present study when the OMCL system is used as a linear reference, they do not. There are a number of factors that may have contributed to this result. The first is the differences in the linear reference systems used to evaluate the effect of compression. Recent research on linear amplification has demonstrated that 15-30 dB of high-frequency emphasis above 800-1000 Hz will provide substantially increased scores over orthotelephonic conditions. In all the studies which demonstrated no benefits with compression, linear reference systems with such high-frequency emphasis were used. On the other hand, in all the studies where scores increased with compression, linear reference systems which failed to provide such high-frequency emphasis were used. The use of these systems may have created artificial increases in scores with compression.

The two studies which demonstrated increased scores with compression are those of Villchur and Yanick. Villchur purposefully limited the amount of high-frequency emphasis provided by his linear reference system for practical reasons. He felt that significant high-frequency emphasis would amplify the high-frequency energy of real-life environmental sounds into the discomfort region. He allowed subjects to choose a low-frequency roll-off of 18-24 dB/octave below a cutoff frequency of 100, 200, 500, or 700 Hz. The average cutoff frequency chosen (400 Hz) was significantly lower than that which is suggested from recent research on linear amplification and may not have provided sufficient high-frequency emphasis. It is interesting to note that the two subjects in Villchur's study who demonstrated the smallest increase in scores with compression, chose cutoff frequencies of 700 Hz, while the subject who demonstrated the greatest increase chose a cutoff frequency of 250 Hz. Yanick also limited the high-frequency emphasis of his reference linear system. This was done because unpublished experiments showed no benefits from this emphasis. He allowed subjects to choose between low-frequency roll-offs of 0, 6, 12, or 18 dB/octave below a cutoff frequency of 1500 Hz. The average slope chosen (9 dB/octave) might have been adequate if he had not used insert receivers. As it was, only minimal high frequency emphasis was provided.

The two studies which demonstrated no increase in scores with compression are the present study and the study of Barfod. In both studies one or more linear reference systems which were individually fitted were included. These systems were chosen on the basis of recent research and all provided significant high-frequency emphasis. Furthermore, in the present study, these linear systems and both compression systems outperformed the ORTHO linear system which lacked high-frequency emphasis.

The use of different compression systems accounts for a second difference between studies. The work of Villchur and Yanick demonstrated increased scores with compression utilizing 2-channel compression systems. Both permitted subject adjustments of the compression ratio in each channel, of "bass" and "treble" equalization and of the relative high-frequency channel level. On the other hand, the best system of Barfod utilized 4 channels and was adjusted solely on the basis of loudness balances to restore normal equal loudness contours as was done for the EL system of the present study. The accuracy with which normal equal loudness contours were restored by his system was, however, limited by the number of compression channels used as well as by measurement and sound presentation errors that also occurred in the present study. The present study's better system utilized 16 channels and had equalization, but not compression ratios which were subject adjusted.

A third difference between studies is in the shape of the subjects' audiograms and the severity of their losses. In all studies except Barfod's subjects had both flat and sharply sloping audiograms. In most studies there was little difference in results obtained for these two groups. However, in the present study more positive findings with compression were obtained for subjects with flat losses. These subjects and the subjects in Villchur's study had the most severe losses (average losses of 52, 52, and 63 dB and of 41, 47, and 67 dB at .5, 1, and 2 kHz). Barfod's subjects and the subjects with sloping-losses in the present study had the least severe losses (average losses of 22, 36, and 62 dB and of 6, 23, and 62 dB at .5, 1, and 2 kHz). In Yanick's study subjects had losses of 40-70 dB in the "speech frequencies". Other differences between studies occurred in the etiologies of the subjects used (a factor which was not carefully controlled or reported in many studies) and in the speech materials and procedures used to evaluate the systems.

The above discussion covers results which were obtained with constant speech input levels before processing. Reduced input levels were tested in quiet both by Villchur (1973) and in the present study. In both studies scores with compression increased relative to scores with linear amplification. At sufficiently reduced input levels there was always an advantage for compression.

It's important to compare compression to linear amplification in both quiet and noisy backgrounds. All recent studies of compression have included tests in noise backgrounds, but only Villchur (1973), Barfod (1976) and the current study included tests in quiet as well. In the present study cafeteria noise (S/N = 30 dB) and reverberation was added to non-reverberant speech, while Villchur added two talker interference (S/N = 10 dB) to reverberant speech and Barfod added speech spectrum noise (S/N = 0 - 15 dB) to nonreverberant speech. In the present study scores with compression decreased relative to scores with linear amplification in noise. In Barfod's study scores with compression and linear amplification were roughly equivalent both in quiet and noise. In Villchur's study the absolute difference in scores (and also the percentage improvement in scores with compression) did not vary in quiet and noise. The detailed explanation of the present study's differing results is not known but may be related to the number of compression channels used.

5.4 Overview - Linear Amplification

In this discussion we would like to focus on the results of recent studies (including the present) that demonstrate 1) the importance of providing adequate high-frequency amplification for subjects with sharply sloping and flat losses 2) the relative invariance of speech test scores to increases in high-frequency emphasis over a nominal value of 20-30 dB for subjects with sharply sloping losses and 3) the failure of linear amplification to restore normal or near normal speech perception. The first result reflects the importance of the audibility of speech cues available in the higher frequencies and the ability of subjects to make use of these cues. High frequency emphasis is required to reduce the relative level of intense low-frequency speech elements and to prevent these elements from limiting the presentation level and audibility of mid-high frequency elements of speech presented at a listener's most comfortable level. The second result reflects the large dynamic range between threshold and discomfort level for subjects with sloping-losses in the lower frequencies and the insensitivity of scores to variation in the presentation level of speech elements in these frequencies when these elements are above threshold and below discomfort level. These first two results are consistent with a major postulate of Articulation Theory (French and Steinberg, 1947; Kryter, 1962) which states that amplifying most speech energy in a given frequency region to be above threshold and below discomfort level allows that energy to contribute as much as possible to overall intelligibility for a given listener and speech test. Articulation theory has also been shown to be reasonably capable of predicting the relative scores obtained by impaired listeners for different linear systems and a given speech test (Fletcher,1952; Barfod, 1972; Macrae and Brigden, 1973). Articulation theory and the concept of the primary importance of audibility are, however, limited, and capable of predicting only relative scores with a specific speech test. They do not, for example, predict the third result: the failure of linear amplification to restore normal speech perception. They also do not explain why this failure was more severe for subjects with flat losses and for certain speech materials (e.g. nonsense sentences). These results are not fully understood.

Future research on linear amplification should aim towards a more thorough understanding of speech perception by impaired listeners. Emphasis should first be placed on developing a more adequate description of the limitations of speech perception by impaired listeners and developing a method of predicting the effect of variations in the frequency-gain characteristics of linear amplification systems on the relative intelligibility of speech in normal environments. Once these two goals have been achieved, development of improved methods to fit hearing aids could follow. Improved fitting procedures for hearing aids could also be developed through empirical studies including further investigation of the OMCL system which provided good results in the present study.

5.5 Overview - Multiple Channel Syllabic Compression

All studies are consistent in demonstrating no advantage for compression over properly chosen linear amplification under conditions where 1) the long-term input level of speech material is held constant before processing and the speech material exhibits minimal word-to-word or item-to-item level variation 2) subjects with moderate losses are used and 3) compression is adjusted roughly to restore normal equal loudness contours. More positive results have, however, been obtained when these limitations have been removed. For example, when there is significant word-to-word level variation, or when reduced input levels are examined, performance with compression is superior to that with linear amplification, particularly for subjects with more severe losses at low frequencies. Also, results of the present study demonstrated advantages for a compression system that included subject adjusted equalization, but no advantage for one that attempted to restore normal equal loudness contours. One contributing factor to the positive results of Yanick (1976) and Villchur (1972) may have been the use of 2-channel compression systems whose characteristics were subject adjusted.

Some of the above results can be explained by the importance of the audibility of speech energy in the mid- to high-frequency regions for speech perception and by the relative invariance (noted in the previous section) of intelligibility scores to variation in the level of speech energy in a given frequency region once it is above threshold and as long as it is below discomfort level. For example, the improvement for subjects with more severe losses and speech materials with more level variation may be due to the lack of audibility of low-level words under these conditions and the extra amplification applied to these words with compression. Also, the general lack of improvement with compression may reflect the ability of properly chosen linear amplification to provide adequate audibility without discomfort for the subjects used who had a relatively large dynamic range between threshold and discomfort. The reduction of scores with compression relative to scores with linear amplification in noise in the present study may have been caused by the use of an a relatively large number of channels (16). In systems with only 2 to 4 channels (e.g., Villchur, 1973 and Barfod, 1976), this reduction was not observed. Finally, for the subjects with flat losses in the present study, the poorer performance of the compression system which was adjusted to restore normal equal loudness contours may be related to the high compression ratios at low frequencies required by this system.

The above discussion indicates that our understanding of syllabic compression is incomplete and points to the need for further research. This research should differ from past research in a number of important areas. First, it should utilize speech materials with more natural level variation. Equalization of the levels of words in sentences may be the most important function of syllabic compression. Because of this, it may be important (as suggested by Barfod, 1976) to evaluate syllabic compression in real-life situations. Second, further research should utilize a wider range of subjects including those with flat audiograms and with more severe losses. Third, research should investigate the effect of training on performance with compression. No study of compression has explored this issue primarily because it has been assumed that compression restores normal speech cues which an impaired listener should have experience with. The acoustic cues introduced by compression may not however be as natural or as salient as assumed and may require training to be utilized. Fourth, a wider range of compression systems should be investigated. The promising results for the COMCL and OMCL system in the present study suggest one approach to this investigation. Specifically, output peak speech levels in each frequency region of the compression systems that are studied could be determined as was done for the above two systems and the number of channels and compression ratio at each frequency could be the main variables. Also, the 2-channel systems used by Villchur and Yanick should be re-evaluated and compared to a properly chosen linear reference system. Finally, an analytic study of the effects of compression on speech cues and of the perceptual importance of these effects needs to be initiated with both normal listeners (possibly listening in noise) and with impaired listeners.

Footnotes

1. The method used differed only in that decisions (based on two trials) caused the attenuation to remain the same, increase by 2 dB, or decrease by 6 dB.
2. Our method differed in that subjects were instructed to respond when the tone bursts reached a level where they would not want to listen to more than one burst, a measurement was terminated if 4 successive response levels were within a 4 dB range, and the measured LDL was taken to be the average of the 4 successive levels.
3. See section 2.2d.3 for a discussion of the relationship between the maximum levels at which subjects listened to the speech materials in these experiments and the levels at which loudness balances were performed.
4. This was indicated by a negligible change in the threshold (< 5 dB) in the impaired ear when noise with an effective masking level of 40-50 dB was introduced in the better ear. Actual threshold differences between ears at frequencies where ABLB's were performed were 54 and 52 dB at 500 Hz, 43 and 61 dB at 1000 Hz, and 49 dB at 4000 Hz.
5. This affected the average 80 phon contours between 250 and 6000 Hz little. In this region, for individual subjects, either the measured MLB's and ABLB's exceeded (or differed little) from the predicted 80 phon con-

tours or the error introduced by assuming a conductive loss was small (as demonstrated by a small error in prediction the highest level MLB's from the next highest MLB's). For subjects IK, FM, and HS the measured balances exceeded the predicted 80 phon contour for roughly 2/3 of the frequencies at which measurements were made and they were never lower than the predicted 80 phon contour by more than 3 dB. For subjects ED and ES the measured balances exceeded the 80 phon contour or were within 3 dB of this contour for 8 of the 14 frequencies at which measurements were made. At the remaining frequencies the maximum error in predicting the highest level MLB from the next highest was less than 2dB at 3 and 4 KHz for ES; less than 2 dB at .5 and 4 KHz for ED, and roughly 5 dB for ES at 2 and 6 KHz and for ED at 6 KHz. If these errors are representative of the errors that would have occurred at higher levels, then the assumption of a conductive loss could have caused an error of less than roughly 3 dB in the 80 phon contour of Figure 3. Furthermore, this error could have occurred only at .5, 3, 4, and 6 KHz.

6. The frequency gain characteristic of the OMCL system used by HS had the required relationship between the average gain within each band but increased slightly within each band.

7. The gain of the EL compression system in dB at frequency

f and input level L was calculated by G(L,f) = LI(f:L1) - L(f:L1), where L(f:L1) is the level of a tone at frequency f presented to a normal listener that is as loud as a tone at 1 KHz at level L1 presented to a normal listener and LI(f:L1) is the level of a tone at frequency f presented to an impaired listener that is as loud as a tone at 1 KHz at level L1 presented to a normal listener. In this formula, all levels are in free-field dB SPL, LI(f:L1) is calculated from each subject's phon equal loudness contours and L(f:L1) is obtained from ISO R226 (1961).

8. Above 500 Hz the compression ratio was equal to the maximum of 1) a curve that increases linearly (log frequency) from 1 to 3 over the frequency region 500-2000 Hz and 2) the normal range between threshold and the 80 phon contour divided by the impaired range.

9. When used as a linear amplifier the system in Figure 4 has distortion of less than 0.2%. When used as a compressor distortion cannot be defined in the standard sense because of the use of discrete 1 dB attenuation steps. An indication of the behavior of the system when used as a compression is obtained by examining the ripple in the lowest frequency channel where this ripple is most severe. At a point immediately following the log converter this ripple has a peak-to-peak value of 1.4 mv for a steady state sine wave within the input

dynamic range of the system. At this point in all channels a 1 dB change in the input corresponds to roughly 33 mv and the voltage range corresponding to the input compression range is roughly 2 volts. At higher frequencies this ripple decreases inversely with frequency.

10. Individual characteristics of the EL system for all input levels chosen in Experiment I are presented in Appendix 12.

11. See Appendix 10 for the details of this limitation.

12. In practice the compression curves instead of the speech input level were varied for the EL system. This varied the "effective" speech input level and made maximum use of the dynamic range of the system.

13. See Appendix 12 for the details of this limitation.

14. This may be caused by the talkers used or by the different types of averaging performed by the peak detectors used in this study and the RMS detectors used by Dunn and White.

15. In Figure 8c many of the points from lists administered in noise and with low-pass filtering to the normal listeners lie below the fitted curve. This was caused by a large asymmetry between vowel and consonant scores that occurred only in this condition. Vowel scores were roughly 90% while consonant scores were roughly 50%.

16. See Hecker (1973) for a discussion of characteristics of speech that can influence the intelligibility of a recorded test.

17. The difference in noise for the flat loss group cannot be compared to the preceding difference because scores were not obtained in noise for the PBNS's. For the CVC's and the flat loss group this difference was 3 points in quiet and 9 points in noise.

18. For the normals scores were saturated (>95%) and the difference between vowel and initial consonant scores increased slightly for the CVC's. For the flat loss group average scores are not directly comparable because of the missing scores for the PBNS's in noise. If scores in quiet are compared then the difference between vowel and initial consonant scores increases from -12 points for the PBNS's to -8 points for the CVC's. This difference was probably caused by the limited number of vowels (6) used in the CVC's.

19. For the 3 normal listeners scores for the first and last two words were greater than 95% for both male and female PBNS's in the Q/A environment. In the N/R environment scores for the last two words were 15 points below scores for the first two words for the female PBNS's and 14 points below scores for the first two words for the male PBNS's.

20. The consonants /ch/, /dz/, and /h/ were not used by

Miller and Nicely. Features for these were specified as follows: /ch/ = 0-voicing, 1-affrication, 1-duration, 2-place, /dz/ = 1-voicing, 1-affrication, 1-duration, 2-place, /h/ = 0-voicing, 1-affrication, 0-duration, (no place for /h/).

21. The agreement is not exact because Figure 5 indicates that speech energy is audible (10% levels >0 dB SL) below 2.5 kHz for the ORTHO system and below 4-5 kHz for the OMCL system. These frequencies can, however, be lowered considerably if the criterion for audibility is made more liberal (e.g. 10% levels >20 dB SL).

22. Errors associated with the decrease in percent transmitted information for the feature duration were: respond /f/ for /s/ or /sh/, respond /v/ for /z/ or /zh/, respond /t/ for /ch/ and respond /d/ for /dz/. Errors associated with the decrease in percent transmitted information for the feature place were: respond /t/ for /p/, /d/ for /g/, /b/ for /d/, /f/ for /s/ or /sh/, and /v/ for /z/.

23. The RMS levels used to calculate the presentation levels in Appendix 15 were calculated for the CID W-22 words from VU measurements by assuming that the relationship between these measurements and the overall RMS level of these words was the same as for the male CVC syllables. The RMS level of the Harvard sentences was determined from VU measurements by assuming that the

relationship between these measurements and the overall RMS level of the sentences was the same as for the SPIN test sentences.

Figure Captions

1. Hearing aid which incorporates the three basic types of amplitude compression systems.

2. Individual thresholds and loudness discomfort levels of subjects in the sloping-loss (a) and flat-loss (b) groups.

3. Average thresholds, equal loudness contours, and loudness discomfort levels for subjects in the sloping-loss (a) and flat-loss (b) subject groups.

4. 16-Channel speech processing system used to implement all linear and compression systems.

5. Average 10% cumulative output speech levels (see text) for the 4 linear and 2 compression systems of experiment I, and for the sloping-loss (a) and flat-loss (b) subject groups. Levels for the 10% linear and EL compression systems, and also for the OMCL linear and COMCL compression systems, are represented by single curves.

6. Functional gain (normalized to 0 dB at 500 Hz) of the 4 linear and 2 compression systems for the average 10% input speech levels of experiment I, for the sloping-loss (a) and flat-loss (b) subject groups. The Normalized functional gains of the 10% linear and EL

compression systems, and also of the OMCL linear and COMCL compression systems are represented by a single curve.

7. Average compression ratio in the region above compression threshold, and below the average 10% input speech levels of experiment I, for the COMCL and EL compression systems and the sloping-loss (a) and flat-loss (b) subject groups.

8. Scatter diagrams for word and phoneme scores and for syllable and phoneme scores for all PBNS and CVC lists administered to the normal and impaired subjects.

9. Average percent phoneme correct scores for the male and female PBNS's in the quiet/anechoic environment (———), and in noisy/reverberant environment (- - - -), for the sloping- loss subject group in experiment I.

10. Average percent phoneme correct scores for the male and female CVC's in the quiet/anechoic environment (———), and in the noisy/reverberant environment (- - - -), for the sloping-loss subject group in experiment I.

11. Average percent phoneme correct scores for the male and female PBNS's in the quiet/anechoic environment for the flat-loss subject group in experiment I,

12. Average percent phoneme correct scores for the male and female CVC's in the quiet/anechoic environment (——), and in the noisy/reverberant environment (---), for the flat-loss subject group in experiment I.

13. Average percent phoneme correct scores for standard tests in experiment II (——), and for CVC's and sentences in experiment I (---), for the sloping-loss group in quiet.

14. Average percent phoneme correct scores for the SPIN test in experiment II (——), and for CVC's and sentences in experiment I (---), for the sloping-loss group in noise. Key for symbols is in Figure 13.

15. Average percent phoneme correct scores for standard tests in experiment II (——), and for CVC's and sentences in experiment I (----), for the flat-loss group in quiet. Key for symbols is in Figure 13.

16. Average percent phoneme correct scores for the SPIN test in experiment II (——), and for CVC's in experiment I (---), for the flat-loss group in noise. Key for symbols is in Figure 13.

17. Overall average percent phoneme correct scores in quiet and in noise for standard tests in experiment II (——), and tests in experiment I (---), for sloping-loss (a) and for flat-loss (b) subject groups.

Scores for the flat-loss group in experiment I in noise are for CVC's only. Curves for the flat-loss group for scores in noise in experiments I and II, and for scores in quiet in experiment I, have been shifted slightly for clarity.

18. Individual percent phoneme correct scores from experiment III for female CVC's presented in quiet at reduced input levels for the best linear (---) and the best compression (——) system from experiment I.

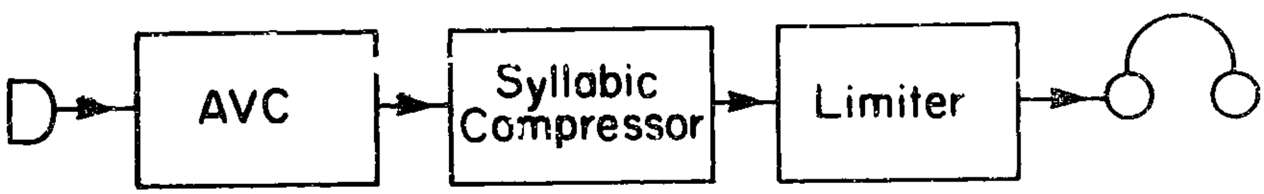


FIGURE 1

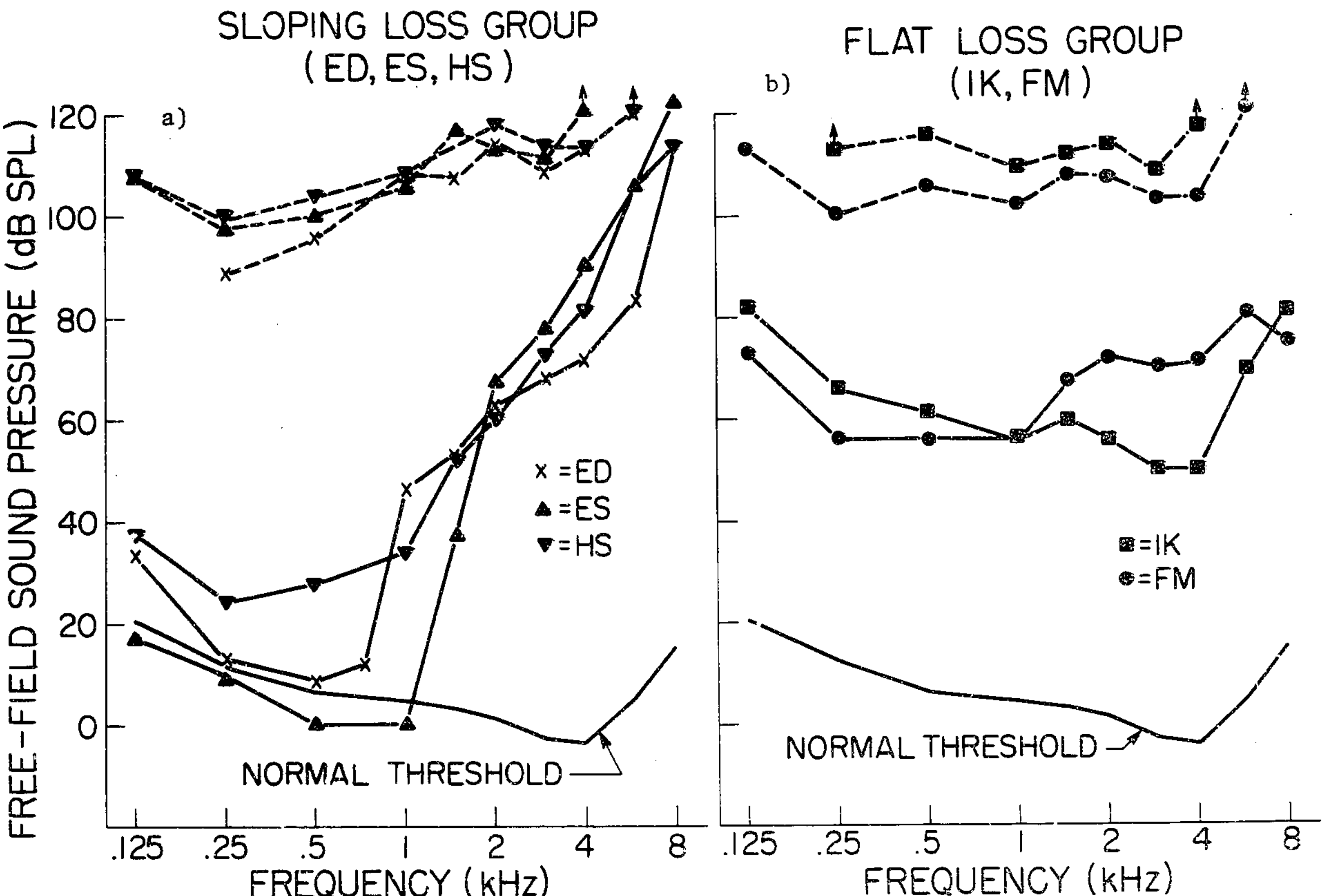

SLOPING LOSS GROUP
(ED, ES, HS)
a)
FLAT LOSS GROUP
(IK, FM)
b)
FREE-FIELD SOUND PRESSURE (dB SPL)
120
100
80
60
40
20
0
× = ED
▲ = ES
▼ = HS
■ = IK
● = FM
NORMAL THRESHOLD
NORMAL THRESHOLD
.125
.25
.5
1
2
4
8
FREQUENCY (kHz)
FREQUENCY (kHz)

FIGURE 2

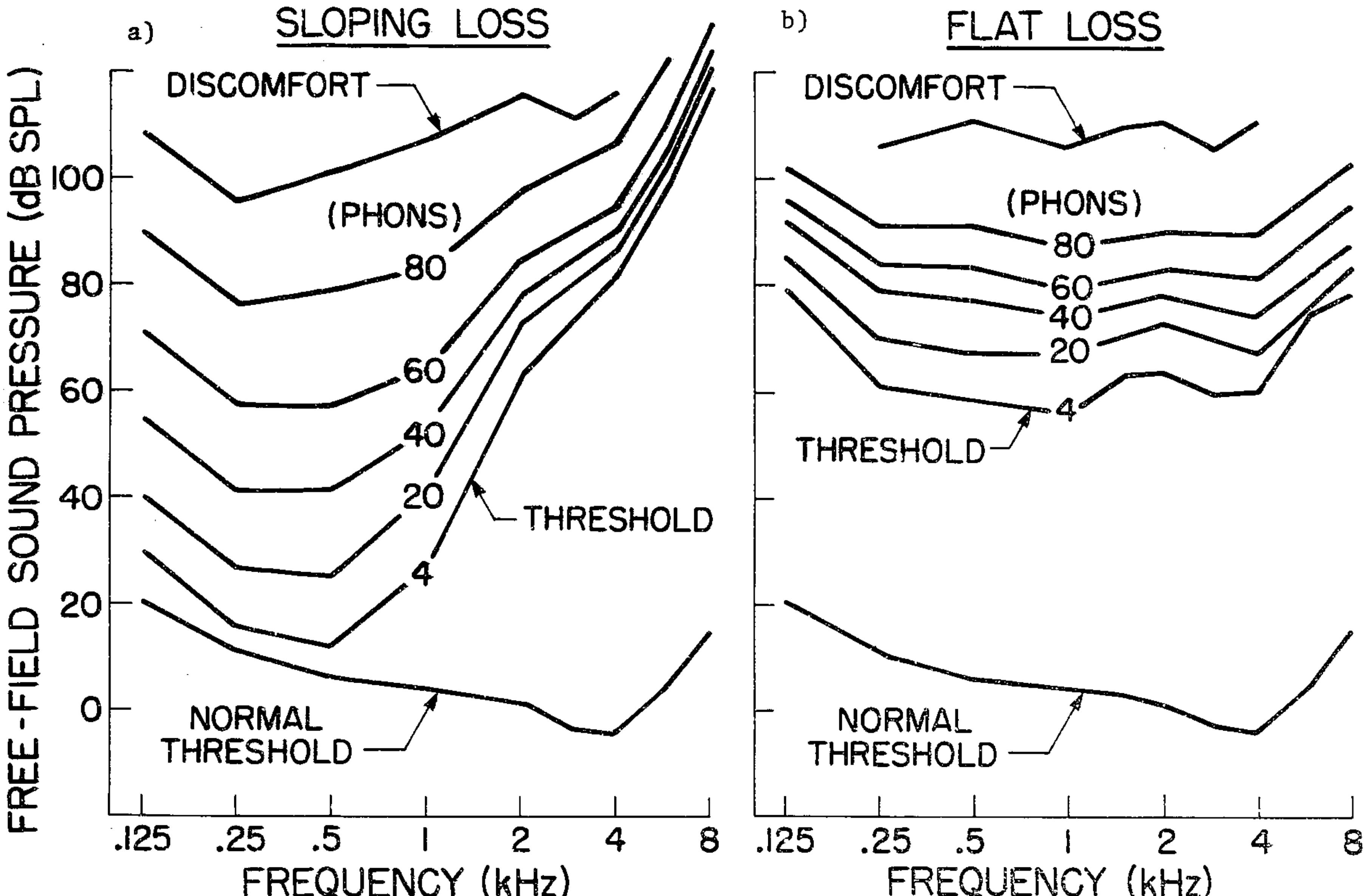

a)
SLOPING LOSS
b)
FLAT LOSS
FREE-FIELD SOUND PRESSURE (dB SPL)
DISCOMFORT
(PHONS)
80
60
40
20
4
THRESHOLD
NORMAL THRESHOLD
100
80
60
40
20
0
.125
.25
.5
1
2
4
8
FREQUENCY (kHz)

FIGURE 3

SPEECH PROCESSING SYSTEM

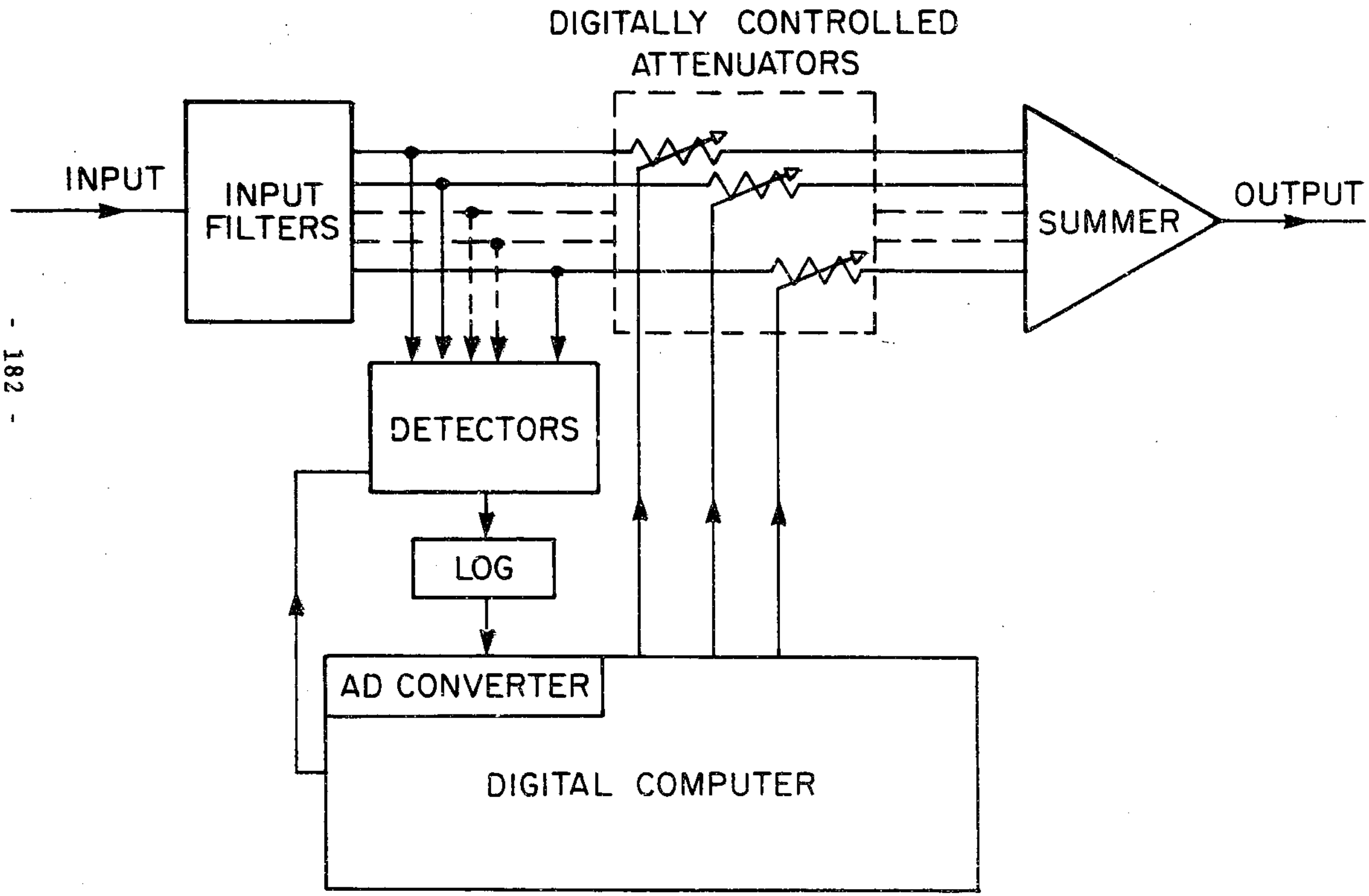


FIGURE 4

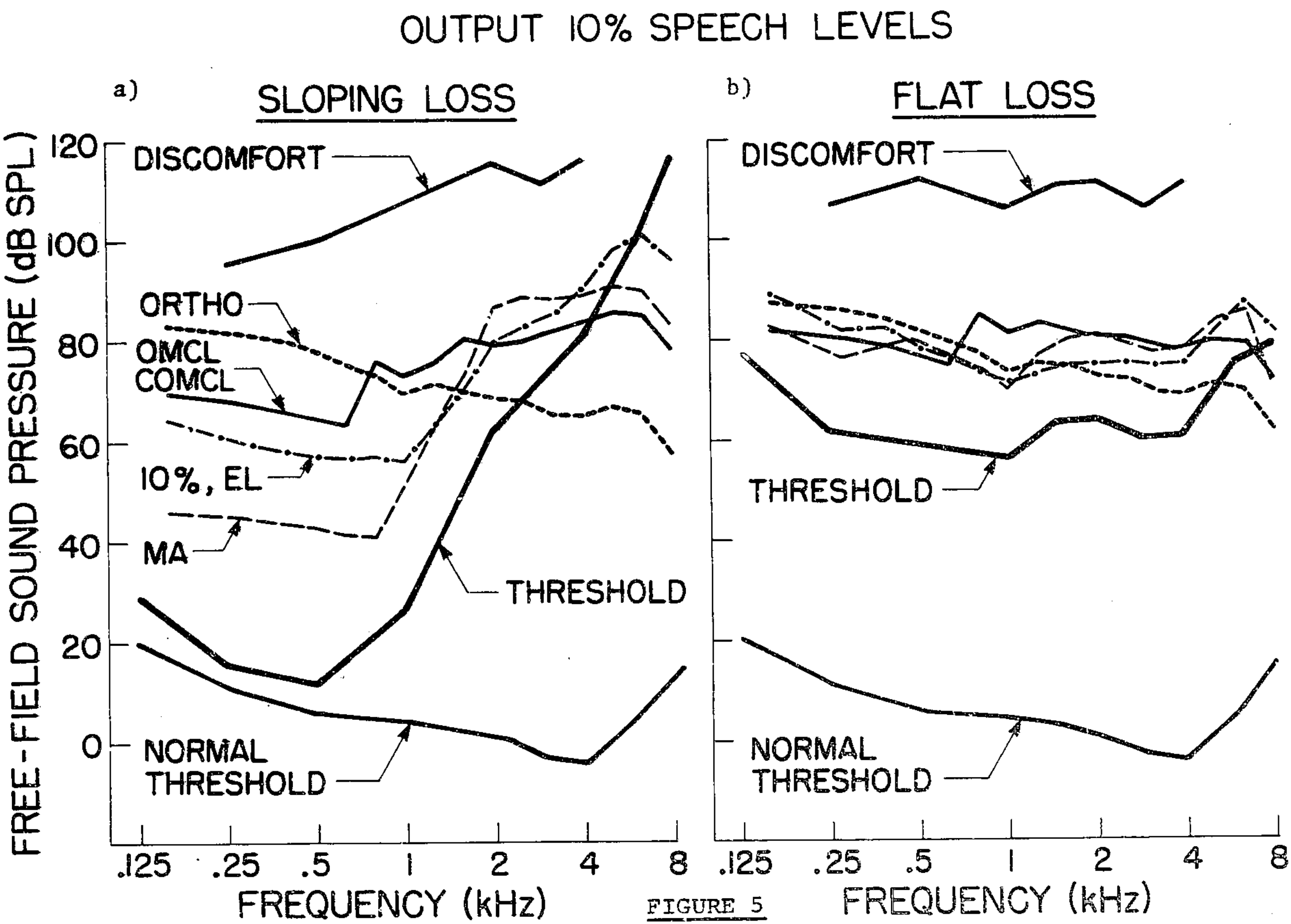

OUTPUT 10% SPEECH LEVELS
a)
SLOPING LOSS
b)
FLAT LOSS
FREE-FIELD SOUND PRESSURE (dB SPL)
120
100
80
60
40
20
0
DISCOMFORT
ORTHO
OMCL
COMCL
10%, EL
MA
THRESHOLD
NORMAL THRESHOLD
DISCOMFORT
THRESHOLD
NORMAL THRESHOLD
.125
.25
.5
1
2
4
8
FREQUENCY (kHz)
.125
.25
.5
1
2
4
8
FREQUENCY (kHz)

FIGURE 5

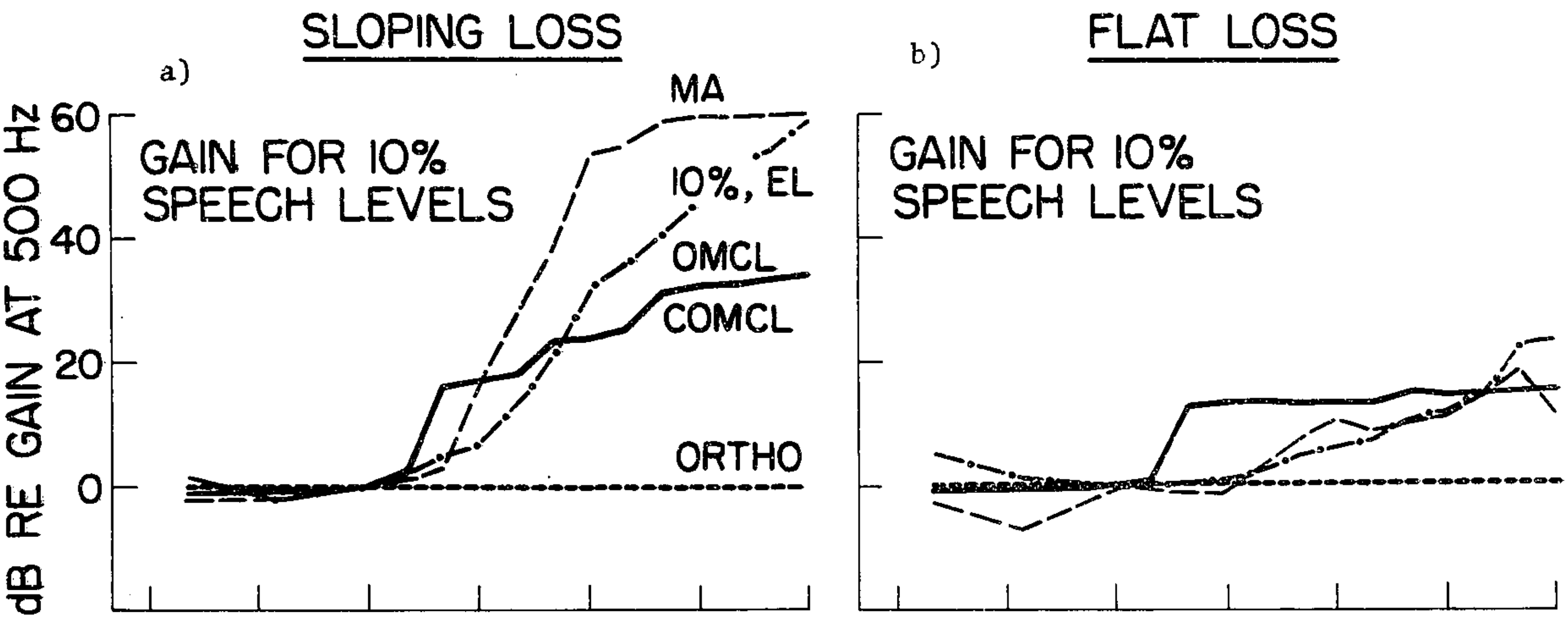


FIGURE 6

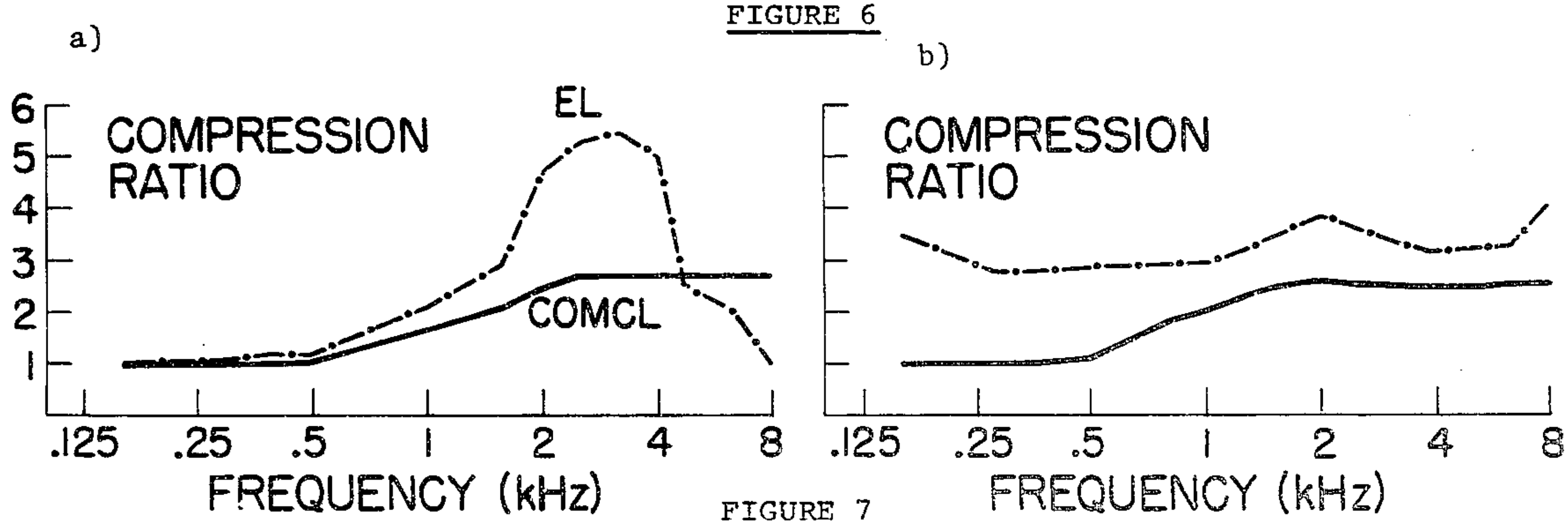


FIGURE 7

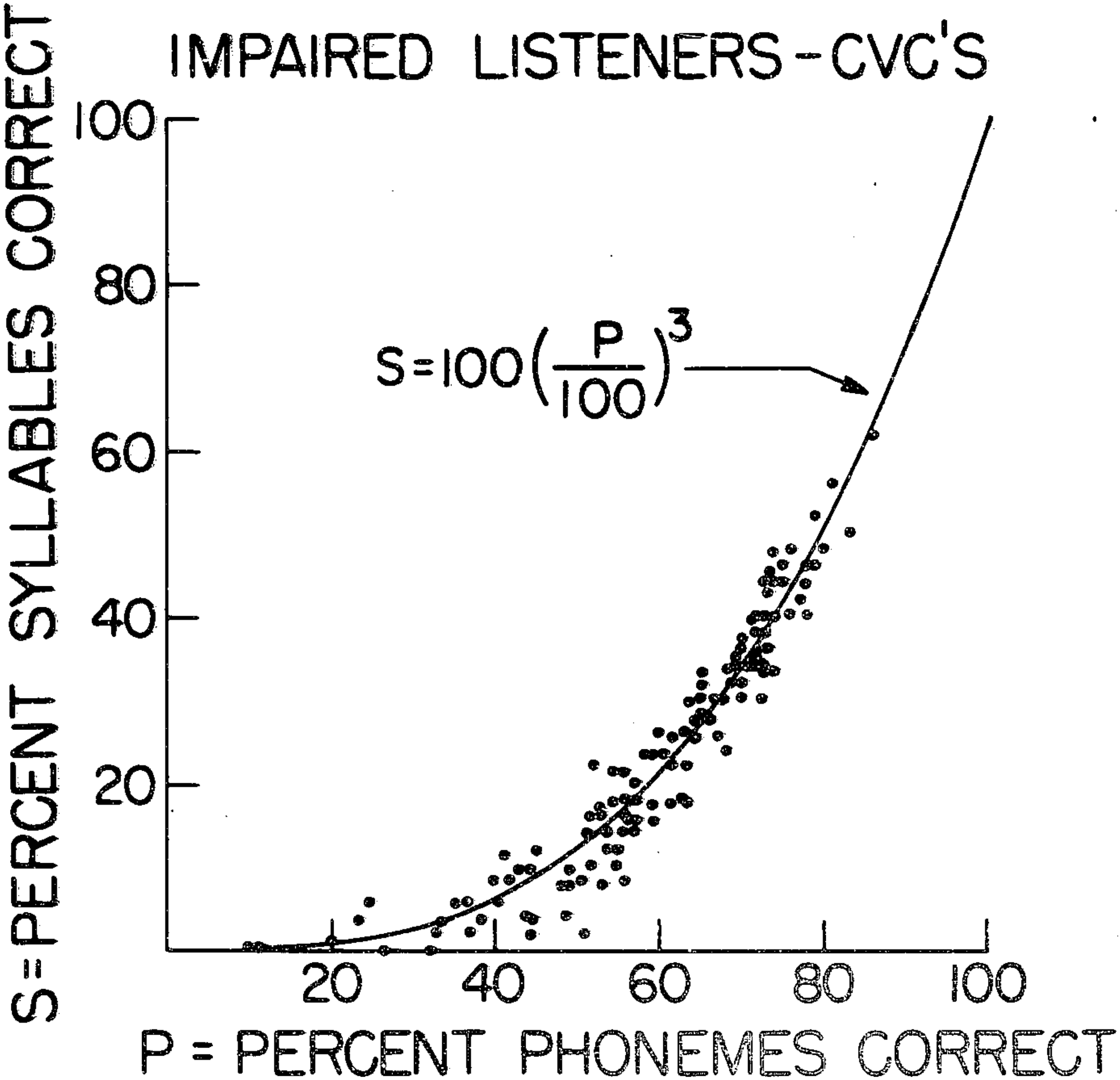

IMPAIRED LISTENERS - CVC'S
S = PERCENT SYLLABLES CORRECT
P = PERCENT PHONEMES CORRECT
$S = 100\left(\frac{P}{100}\right)^3$
100
80
60
40
20
20
40
60
80
100

FIGURE 8a

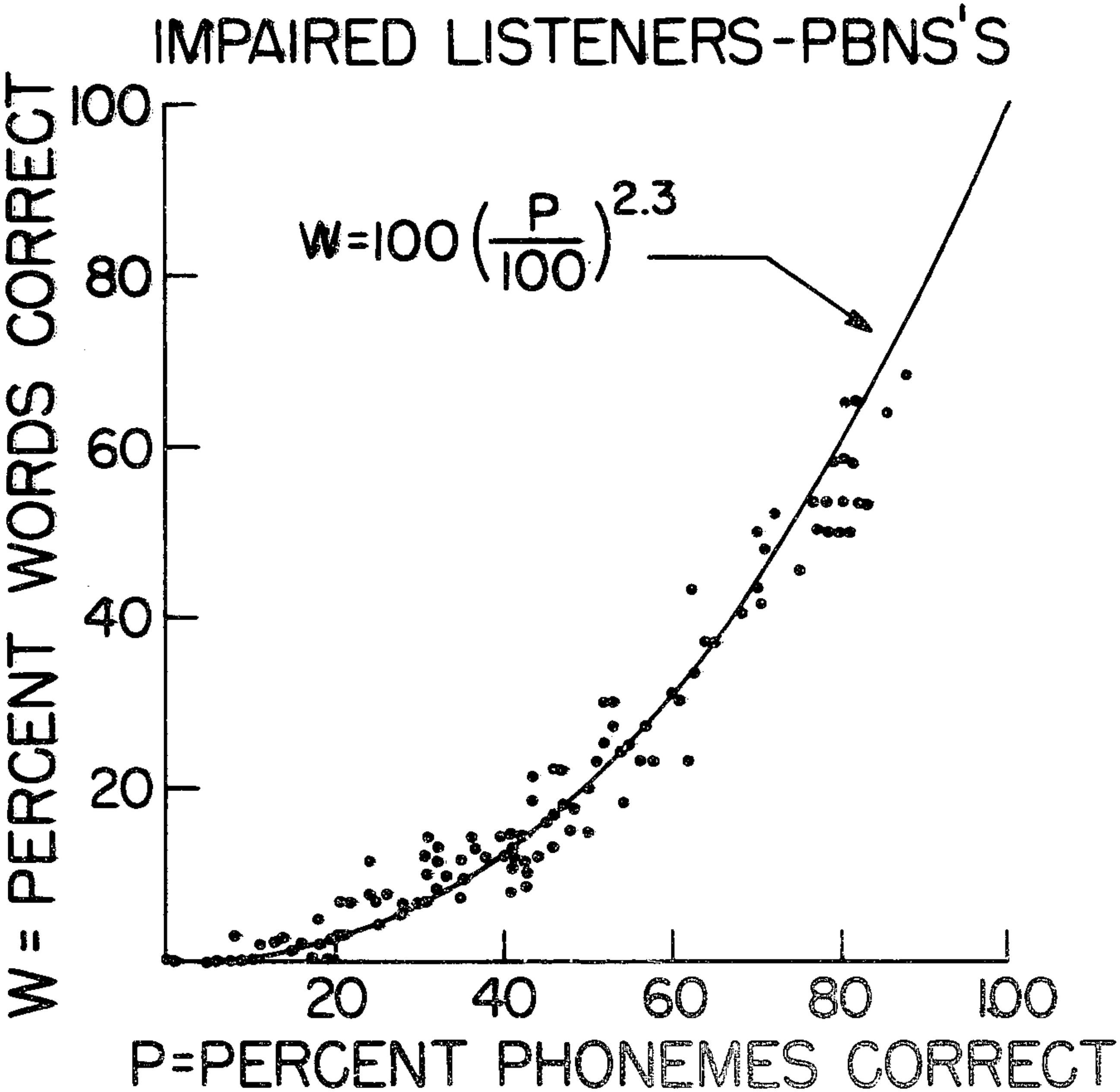

IMPAIRED LISTENERS-PBNS'S
W = PERCENT WORDS CORRECT
100
80
60
40
20
$W=100\left(\frac{P}{100}\right)^{2.3}$
20
40
60
80
100
P=PERCENT PHONEMES CORRECT

FIGURE 8b

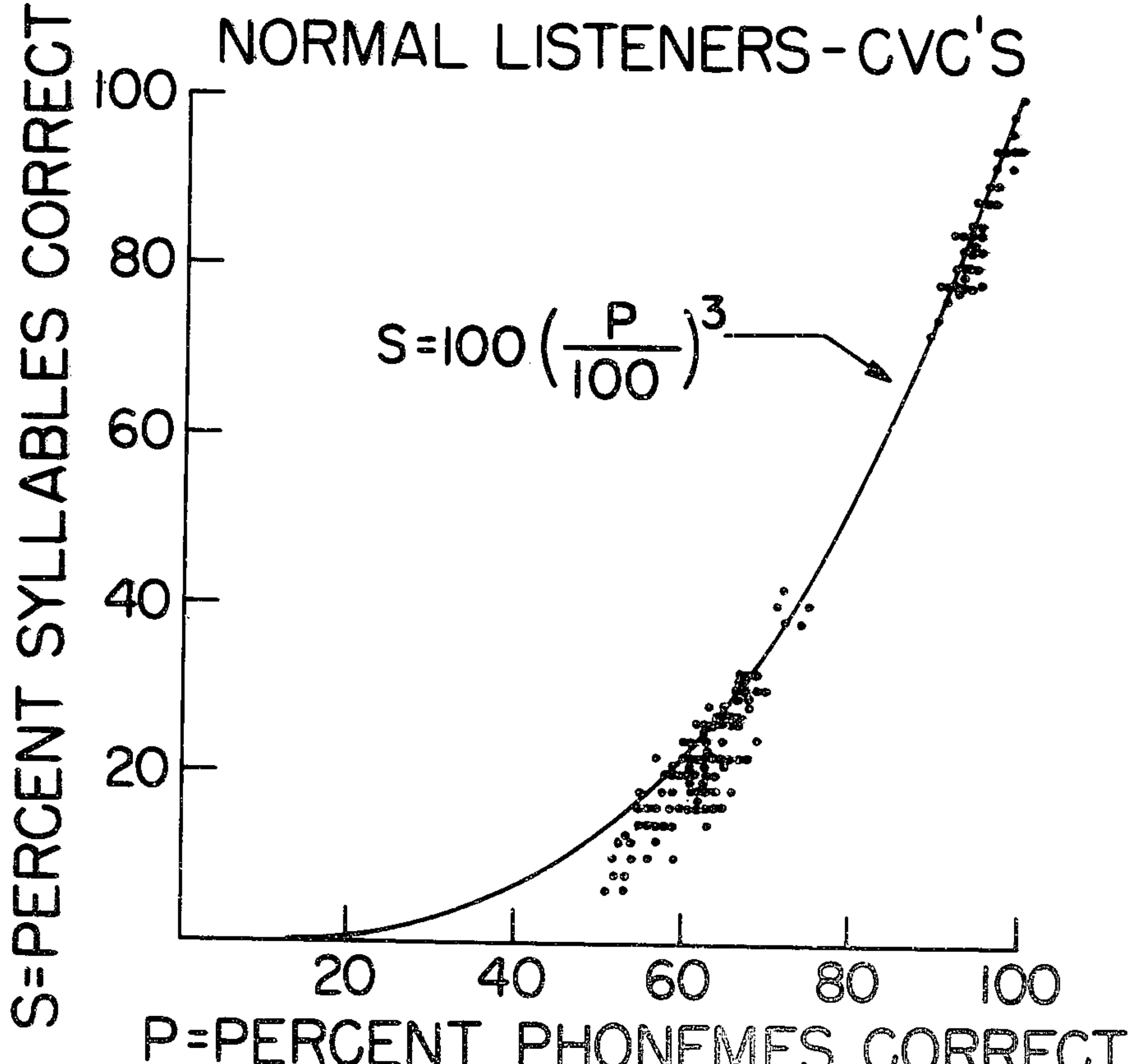

NORMAL LISTENERS - CVC'S
S=PERCENT SYLLABLES CORRECT
P=PERCENT PHONEMES CORRECT
$S = 100\left(\frac{P}{100}\right)^3$
100
80
60
40
20
20
40
60
80
100

FIGURE 8c

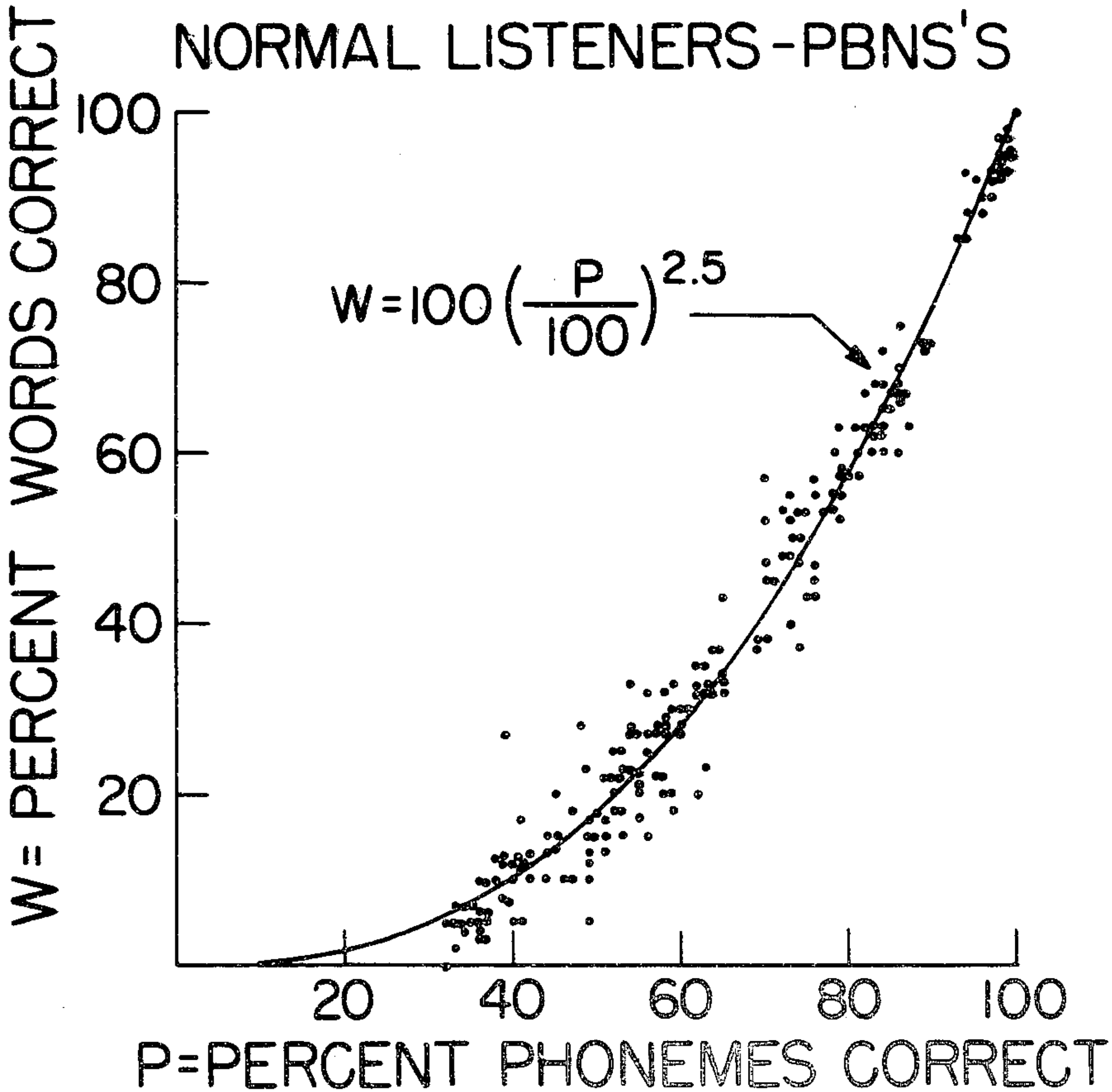

NORMAL LISTENERS-PBNS'S
W= PERCENT WORDS CORRECT
100
80
60
40
20
20
40
60
80
100
P=PERCENT PHONEMES CORRECT
W=100 (P/100)^2.5

FIGURE 8d

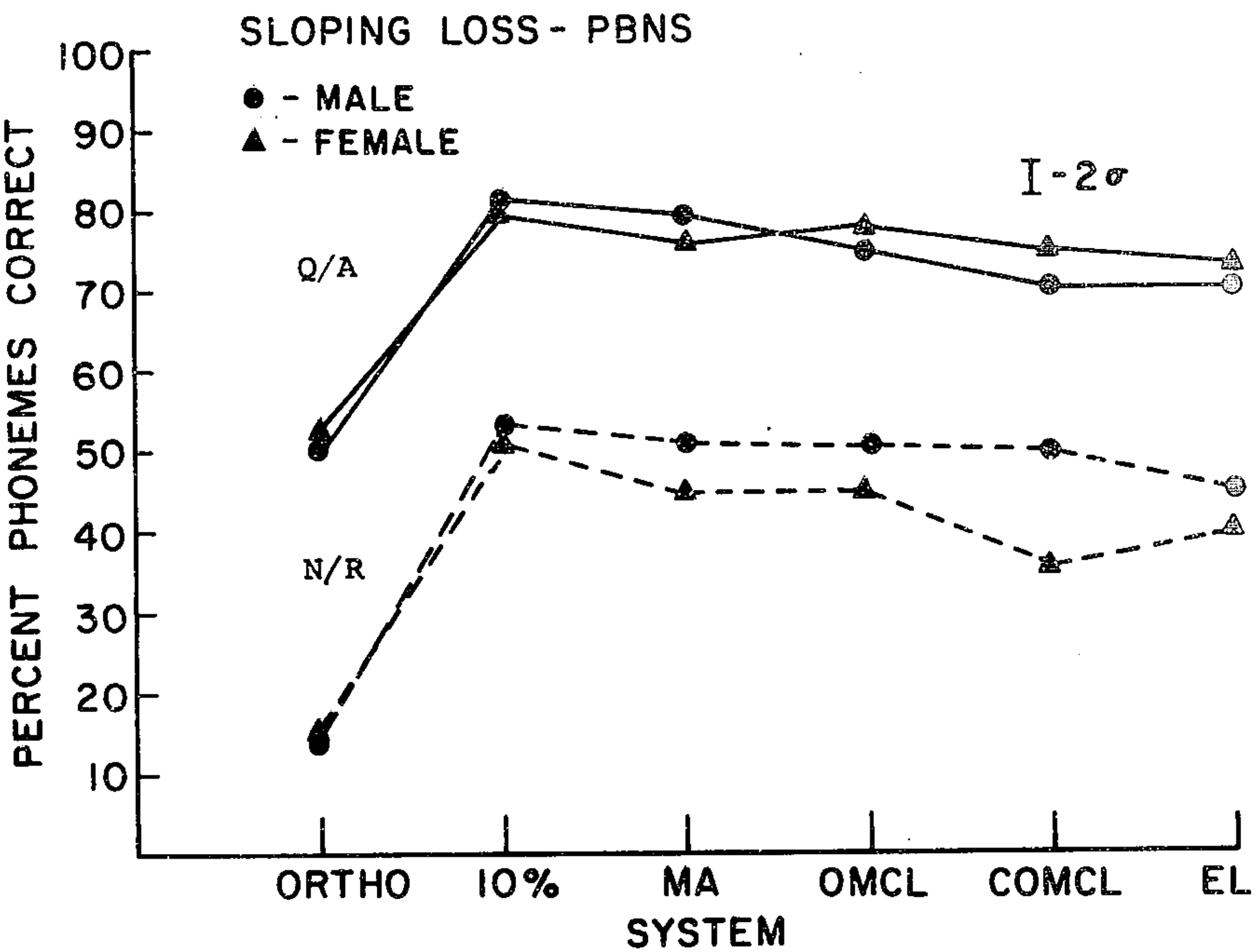

SLOPING LOSS - PBNS
● - MALE
▲ - FEMALE
I - 2σ
Q/A
N/R
PERCENT PHONEMES CORRECT
100
90
80
70
60
50
40
30
20
10
ORTHO
10%
MA
OMCL
COMCL
EL
SYSTEM

FIGURE 9

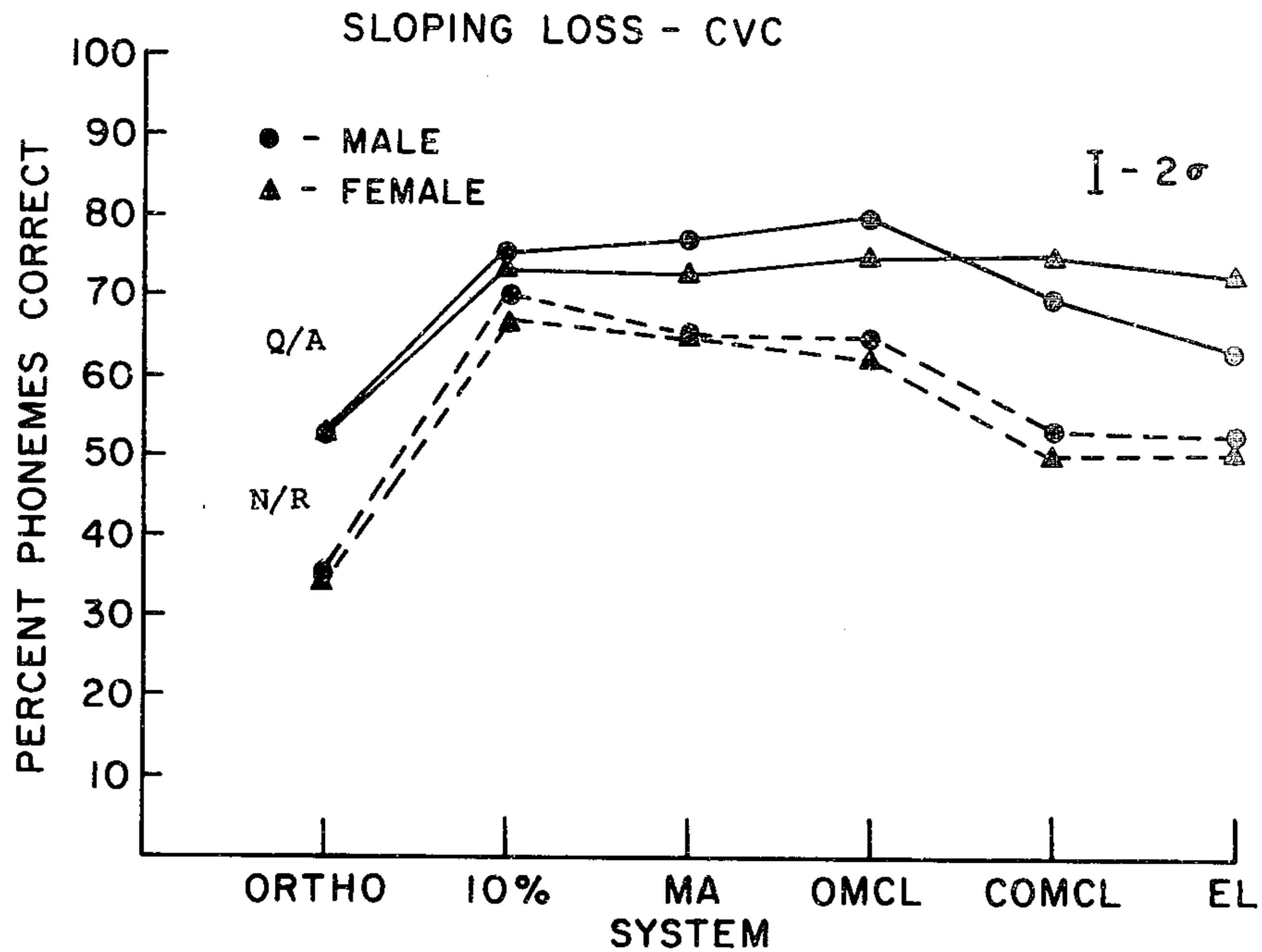

SLOPING LOSS - CVC
● - MALE
▲ - FEMALE
I - 2σ
Q/A
N/R
PERCENT PHONEMES CORRECT
100
90
80
70
60
50
40
30
20
10
ORTHO
10%
MA
OMCL
COMCL
EL
SYSTEM

FIGURE 10.

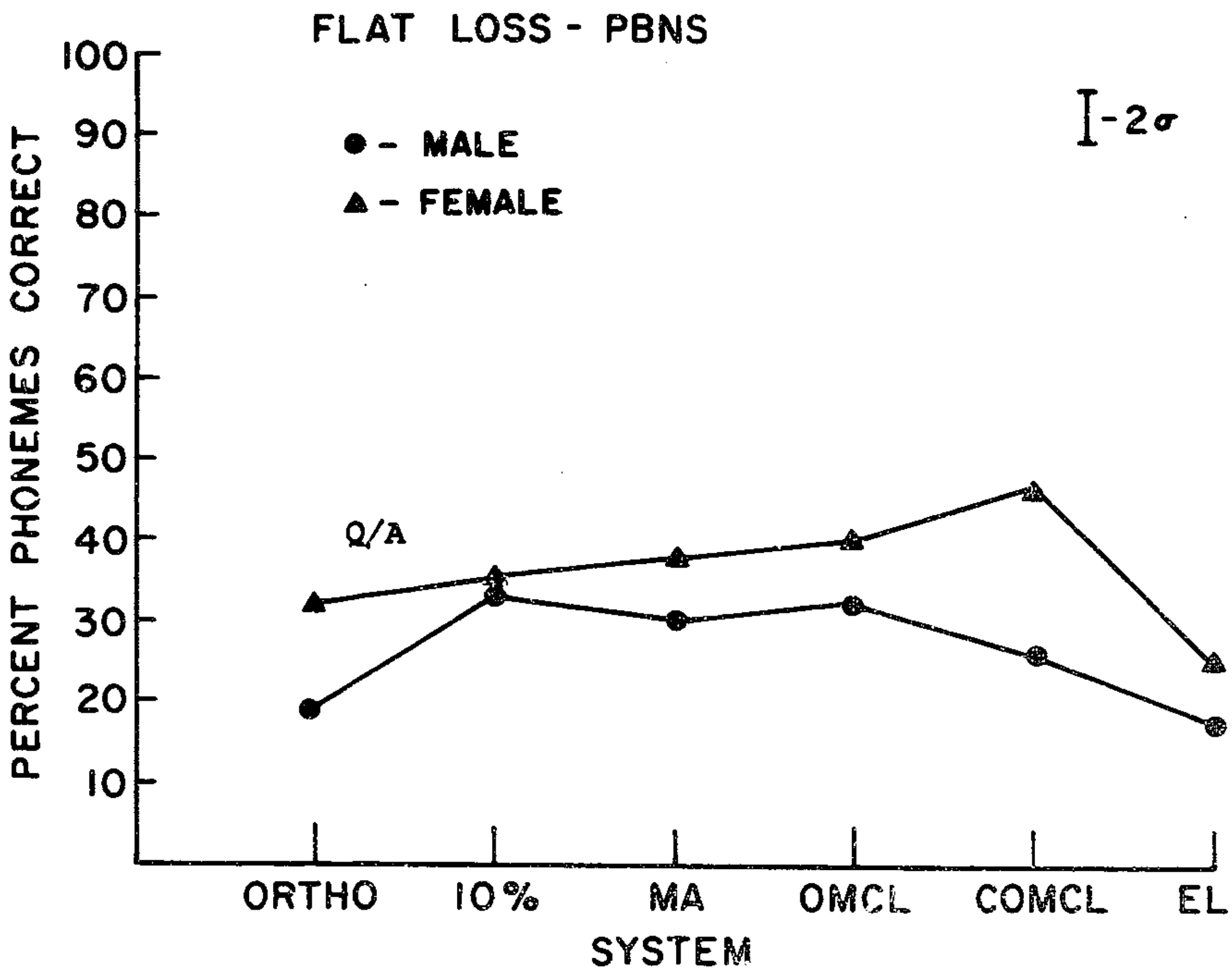

FLAT LOSS - PBNS
100
90
80
70
60
50
40
30
20
10
PERCENT PHONEMES CORRECT
● - MALE
▲ - FEMALE
I-2σ
Q/A
ORTHO
10%
MA
OMCL
COMCL
EL
SYSTEM

FIGURE 11.

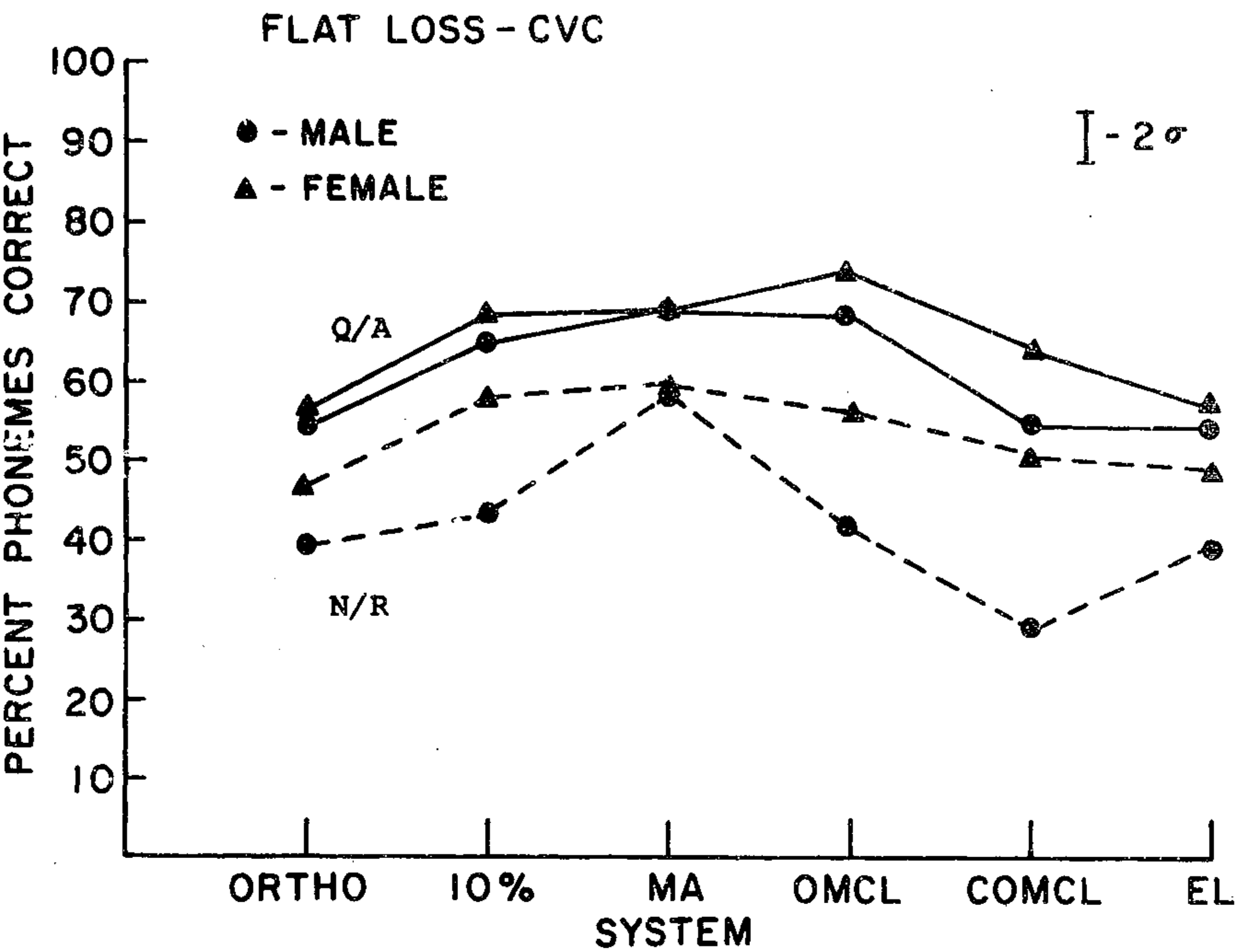

FLAT LOSS - CVC
● - MALE
▲ - FEMALE
I - 2σ
Q/A
N/R
PERCENT PHONEMES CORRECT
100
90
80
70
60
50
40
30
20
10
ORTHO
10%
MA
OMCL
COMCL
EL
SYSTEM

FIGURE 12.

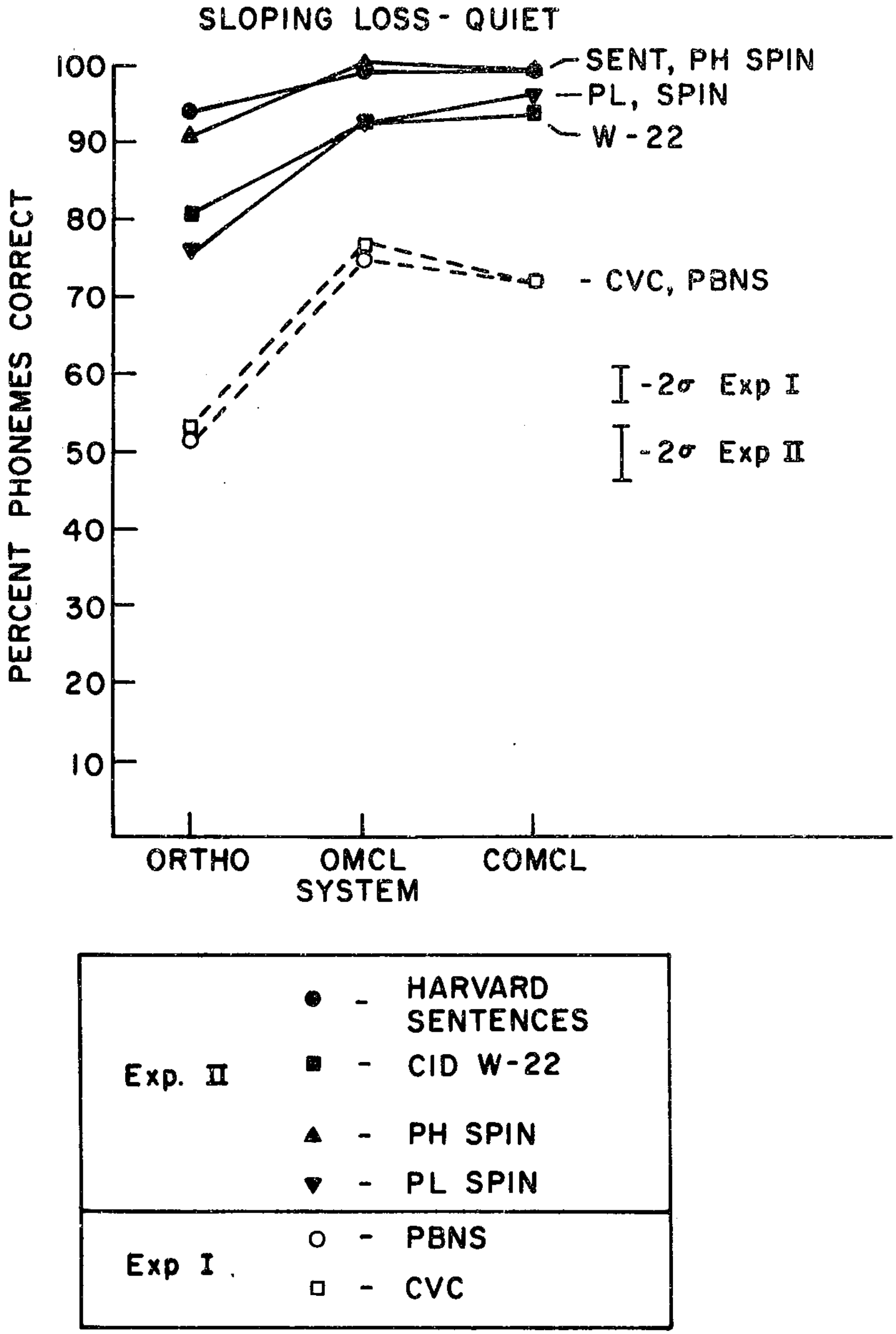

SLOPING LOSS - QUIET
PERCENT PHONEMES CORRECT
100
90
80
70
60
50
40
30
20
10
SENT, PH SPIN
PL, SPIN
W - 22
- CVC, PBNS
-2σ Exp I
-2σ Exp II
ORTHO
OMCL
COMCL
SYSTEM
Exp. II
● - HARVARD SENTENCES
■ - CID W-22
▲ - PH SPIN
▼ - PL SPIN
Exp I
○ - PBNS
□ - CVC

FIGURE 13.

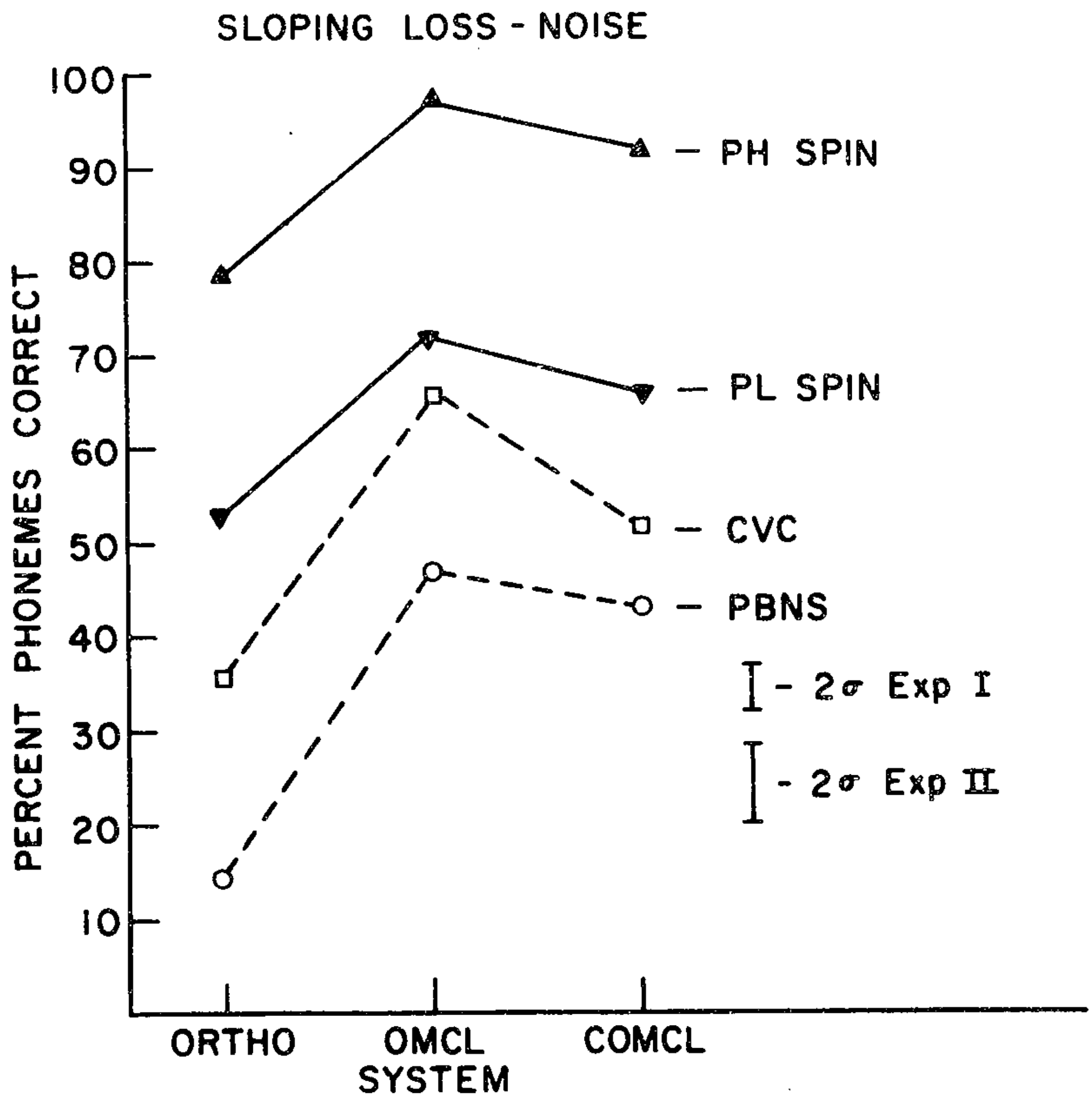

SLOPING LOSS - NOISE
PERCENT PHONEMES CORRECT
100
90
80
70
60
50
40
30
20
10
PH SPIN
PL SPIN
CVC
PBNS
- 2σ Exp I
- 2σ Exp II
ORTHO
OMCL
COMCL
SYSTEM

FIGURE 14.

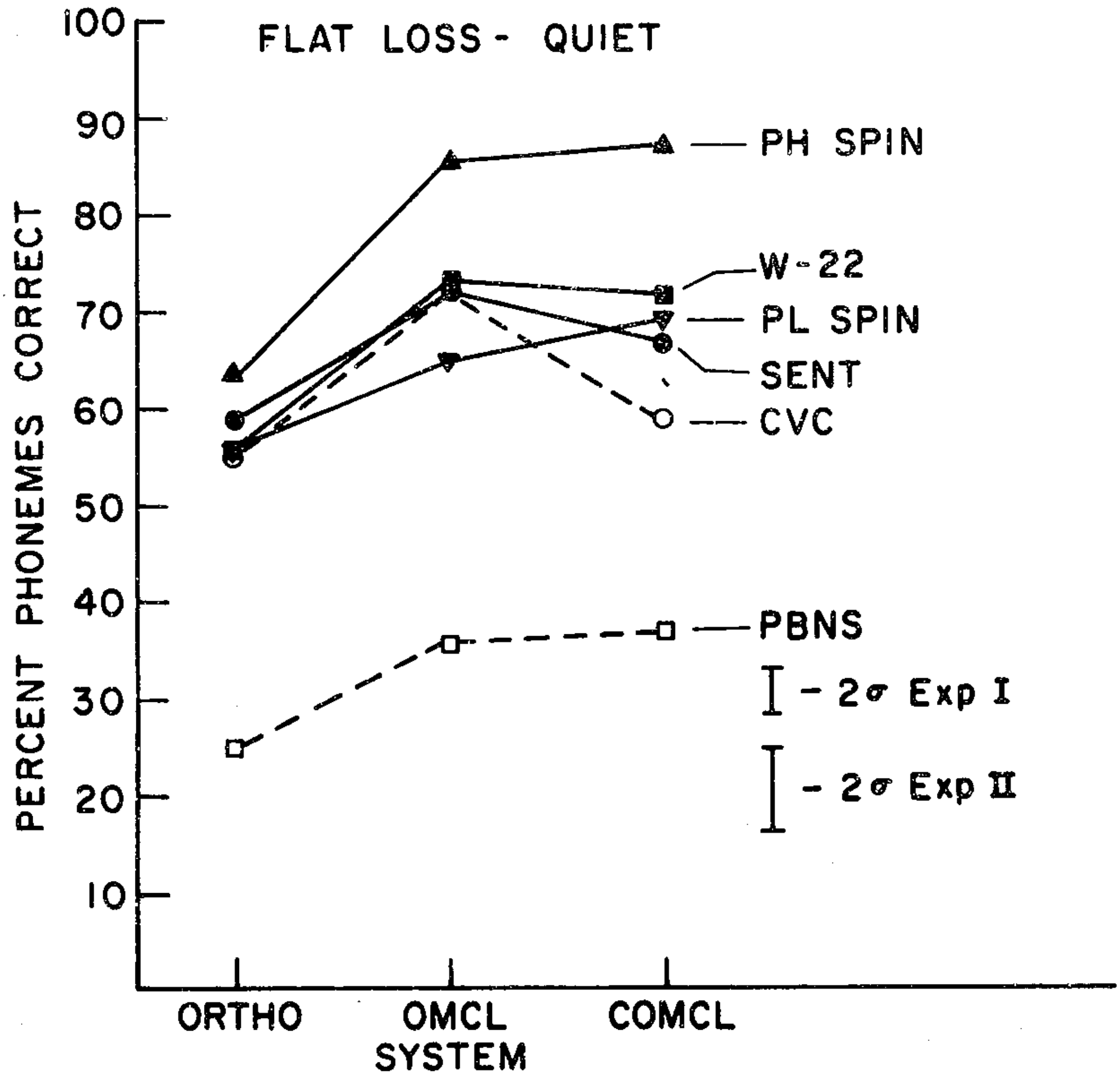

FLAT LOSS - QUIET
PERCENT PHONEMES CORRECT
100
90
80
70
60
50
40
30
20
10
PH SPIN
W-22
PL SPIN
SENT
CVC
PBNS
- 2σ Exp I
- 2σ Exp II
ORTHO
OMCL
COMCL
SYSTEM

FIGURE 15.

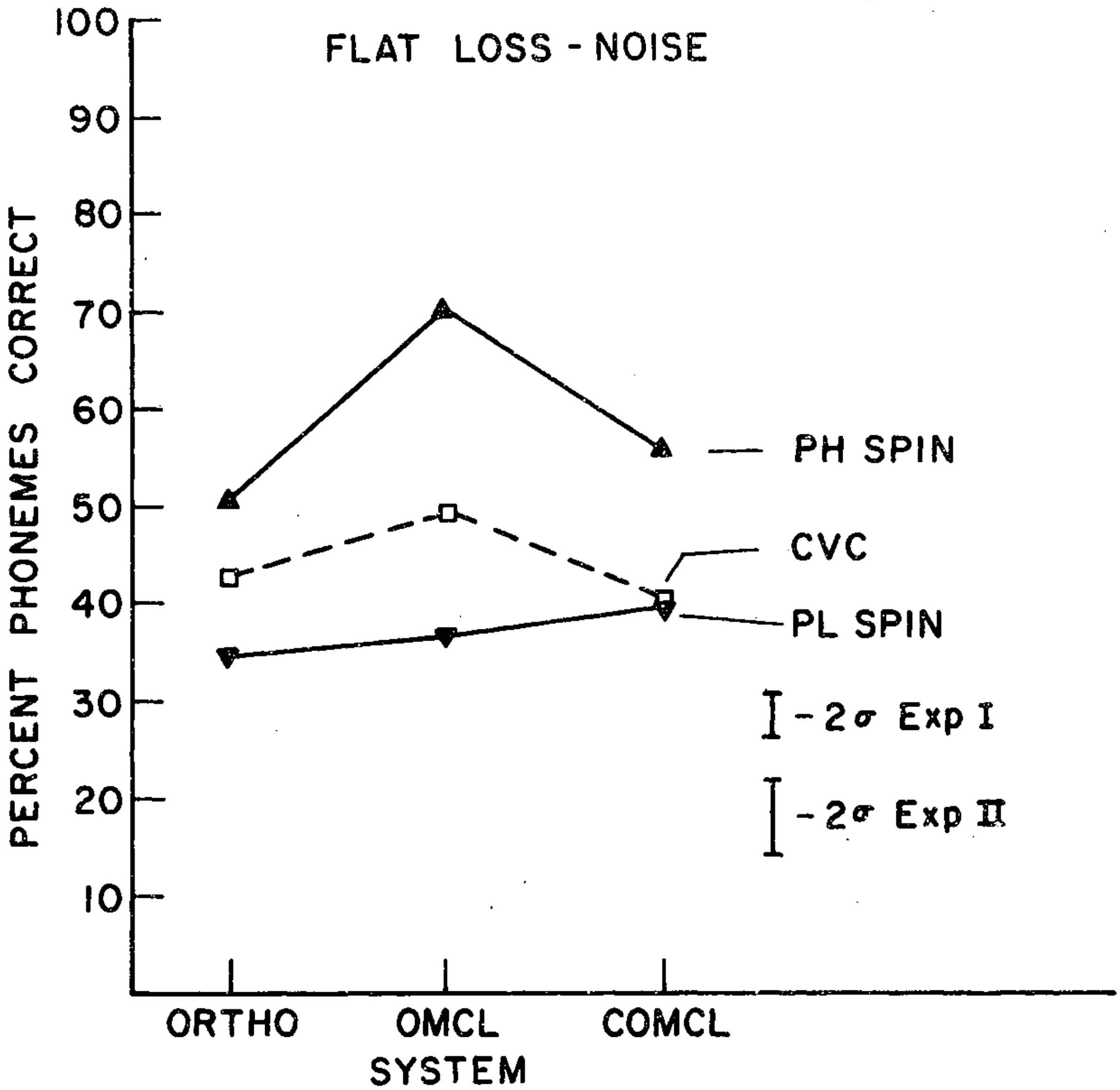

FLAT LOSS - NOISE
PERCENT PHONEMES CORRECT
100
90
80
70
60
50
40
30
20
10
PH SPIN
CVC
PL SPIN
- 2σ Exp I
- 2σ Exp II
ORTHO
OMCL
COMCL
SYSTEM

FIGURE 16.

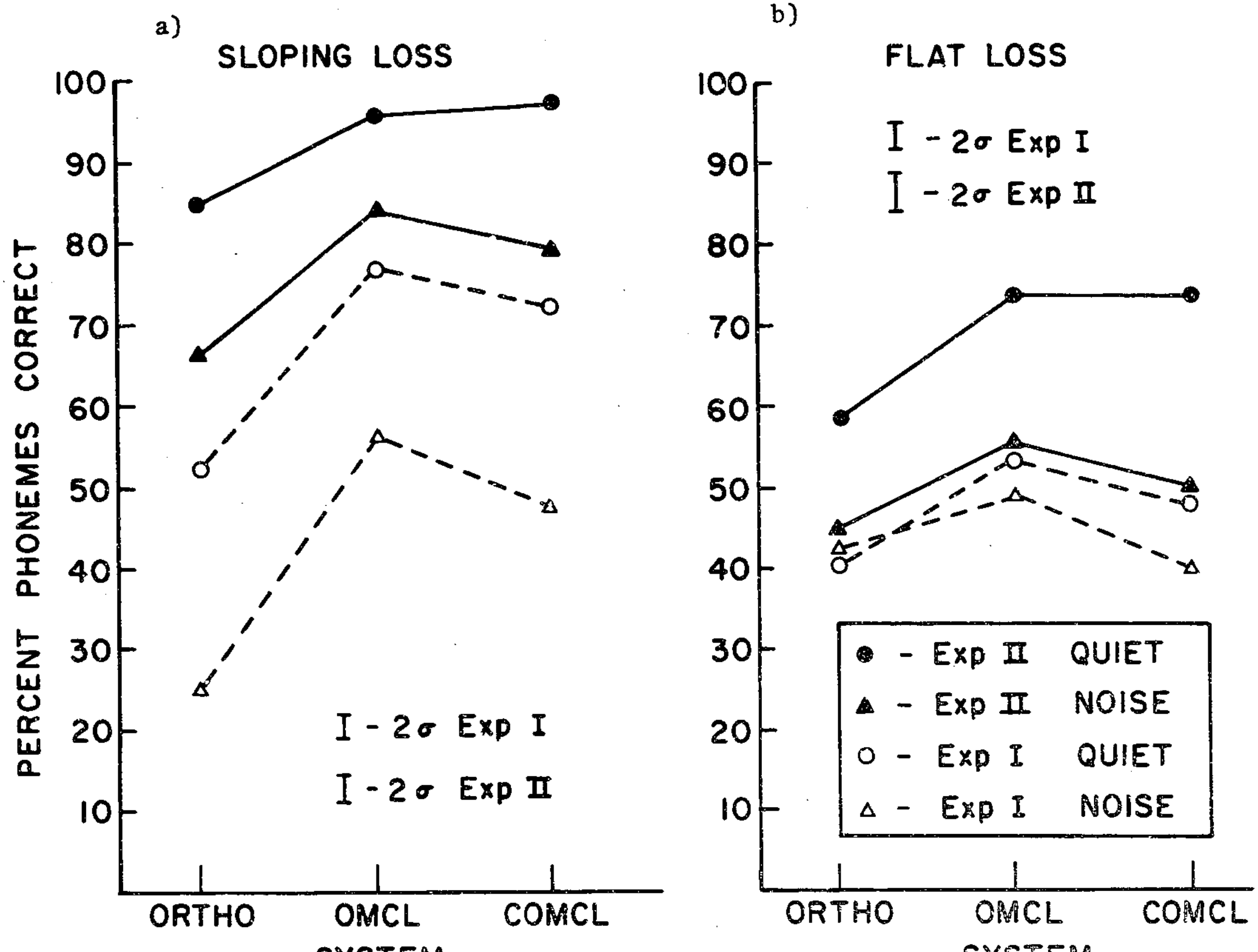

a)
SLOPING LOSS
PERCENT PHONEMES CORRECT
100
90
80
70
60
50
40
30
20
10
I - 2σ Exp I
I - 2σ Exp II
ORTHO
OMCL
COMCL
SYSTEM
b)
FLAT LOSS
100
90
80
70
60
50
40
30
20
10
I - 2σ Exp I
I - 2σ Exp II
● - Exp II QUIET
▲ - Exp II NOISE
○ - Exp I QUIET
△ - Exp I NOISE
ORTHO
OMCL
COMCL
SYSTEM

FIGURE 17.

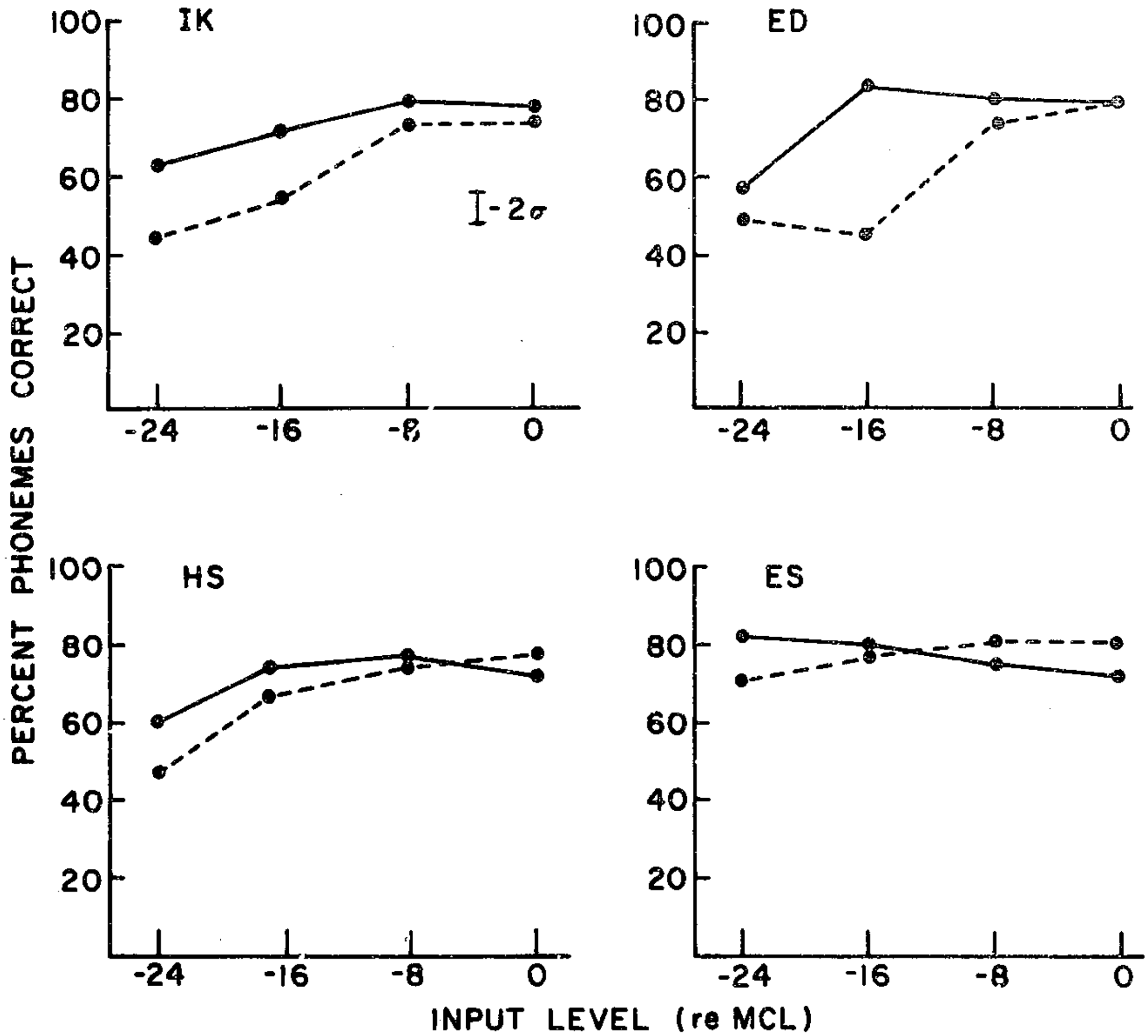

PERCENT PHONEMES CORRECT
IK
ED
HS
ES
I-2σ
100
80
60
40
20
-24
-16
-8
0
INPUT LEVEL (re MCL)

FIGURE 18.

Table 1. The Six Systems Used in Experiment I.

LINEAR SYSTEMS

1.	ORTHO	-	(Orthotelephonic) - Flat functional gain.
2.	MA	-	(Mirror Audiogram) - Functional gain proportional to hearing loss.
3.	OMCL	-	(Octaves at most comfortable levels) - Bands of speech in isolation at most comfortable levels.
4.	10%	-	Restore normal loudness to tones at 10% speech levels.

COMPRESSION SYSTEMS

1.	EL	-	(Equal Loudness) - Restore normal equal loudness contours.
2.	COMCL	-	(Compressed octaves at most comfortable levels) - Bands of compressed speech in isolation at most comfortable levels.

Table 2. Overall Percent Phoneme Correct Scores - Impaired and Normal Subjects.

Mat.	Talker	Env.	Impaired Sloping Loss	Impaired Flat Loss	Normal
PBNS	Male	Q/A	73	26	98
PBNS	Male	N/R	44	-	83
PBNS	Female	Q/A	72	38	98
PBNS	Female	N/R	38	-	75
CVC	Male	Q/A	70	61	99
CVC	Male	N/R	57	42	96
CVC	Female	Q/A	70	65	98
CVC	Female	N/R	55	53	93

Table 3. Overall Quality Judgements for Impaired Subjects.

Mat.	Talker	Env.	Sloping Loss	Flat Loss
PBNS	Male	Q/A	61	15
PBNS	Male	N/R	44	-
PBNS	Female	Q/A	60	30
PBNS	Female	N/R	39	-
CVC	Male	Q/A	74	64
CVC	Male	N/R	41	37
CVC	Female	Q/A	70	64
CVC	Female	N/R	49	41

Table 4. Percent Phoneme Correct Scores Averaged over Major Conditions of Experiment I.

SUBJECT GROUP	CONDITION	SYSTEM ORTHO	10%	MA	OMCL	COMCL	EL	2σ
SLOPING	QUIET	52.3	77.7	76.2	76.8	72.5	59.7	2.1
SLOPING	NOISE	25.0	60.5	57	56.5	47.2	47	2.1
SLOPING	PBNS's	33	66.5	62.7	61.9	57.7	57	2.1
SLOPING	CVC's	44.2	71.7	70.5	71.7	62	59.7	2.1
SLOPING	MALE	38.3	70	68.3	69	60.8	57.5	2.1
SLOPING	FEMALE	39	68.3	65	65.3	59	59.3	2.1
SLOPING	OVERALL	38.6	69.1	66.6	66.8	59.8	58.3	1.5
FLAT	QUIET	40.1	50.5	51.7	53.7	48.1	38.6	2.5
FLAT	NOISE(a)	43	50.7	59	49.2	40.2	44	3.5
FLAT	PBNS's(b)	25.3	34.2	34.3	36.3	36.9	21.7	3.5
FLAT	CVC's	49	58.7	64	60.1	49.7	47.3	2.5
FLAT	MALE	37.1	47.1	56.6	47.7	36.5	33.8	2.9
FLAT	FEMALE	45	53.0	55.5	55.6	54.6	43.6	2.9
FLAT	OVERALL	41.1	50.6	54.1	52.2	45.5	40.4	2.0

a) Averaged only over scores for CVC's.
b) Averaged only over scores in quiet.

Table 5. Average Quality Judgements in Experiment I

SUBJECT GROUP	CONDITION	SYSTEM ORTHO	10%	MA	OMCL	COMCL	EL	2σ
SLOPING	QUIET	49	69	63	81	73	62	8.2
SLOPING	NOISE	14	60	49	58	41	39	8.2
FLAT	QUIET	49	48	41	51	44	32	9.8
FLAT	NOISE*	50	50	43	39	14	36	14

* These averages are only with CVC's and not sentences.

Table 6. Percent Relative Transmitted Information for Initial and Final Consonants in CVC's Presented in Quiet in Experiment I

SUBJECT GROUP	SYSTEM	VOICING	FRICATION	DURATION	PLACE	TOTAL
SLOPING	ORTHO	89.7	57.1	26.4	16.4	55.3
SLOPING	OMCL	98.4	83.3	51.8	42.3	75.2
SLOPING	COMCL	97.0	81.3	43.0	34.6	71.0
SLOPING	EL	91.0	61.6	35.9	29.4	64.3
FLAT	ORTHO	69.2	47.7	70.2	37.4	63.3
FLAT	OMCL	68.2	64.6	83.7	52.0	74.8
FLAT	COMCL	70.7	62.7	43.0	25.2	60.6
FLAT	EL	89.1	70.1	40.2	18.8	62.2

Table 7. Percent Phoneme Correct Scores Averaged over Quiet and Noise in Experiment II.

SUBJECT GROUP	CONDITION	ORTHO	OMCL	COMCL	2σ
SLOPING	QUIET(a)	85.2	96.1	97	2.4
SLOPING	NOISE(b)	66.2	85	79.2	4.7
FLAT	QUIET(a)	58.8	74	73.9	2.9
FLAT	NOISE(b)	43.3	53.8	48	5.8

a) Averaged over scores for Harvard Sentence Lists, SPIN Sentence Lists (PH and PL sentences counted seperately) and CID W-22 word lists.
b) Averaged over scores for SPIN Sentence Lists.

Table 8. Average Percent Word Correct Scores in Experiment II.

		SLOPING LOSS GROUP SYSTEMS				FLAT LOSS GROUP SYSTEMS			
TEST	CONDITION	ORTHO	OMCL	COMCL	2σ	ORTHO	OMCL	COMCL	2σ
Harvard Sent.	Quiet	89.7	99	95	5.8	27	65.5	59	7.1
CID W-22	Quiet	59.3	80.7	90	8.2	20	54	50	10
SPIN PL Sent.	Quiet	48	78.7	88	11.6	20	36	36	14.2
SPIN PH Sent.	Quiet	86.7	100	98.7	11.6	36	82	82	14.2
SPIN PL Sent.	Noise	17.3	33.3	33.3	11.6	8	12	14	14.2
SPIN PH Sent.	Noise	68	86.7	90.7	11.6	12	62	38	14.2

Table 9. Average Quality Judgements in Experiment II.

SUBJECT GROUP	CONDITION	SYSTEM ORTHO	OMCL	COMCL	2σ
Sloping	Quiet	62	91	92	9.4
Sloping	Noise	31	74	43	16.2
Flat	Quiet	50	55	62	11.6
Flat	Noise	29	14	21	20.2

Table 10. Recent Research on Linear Amplification

STUDY	LOSSES	BEST HF EMPHASIS (a)	SCORES ORTHO (b)	SCORES BEST	MATERIALS	ITEM SCORED
Barfod (1972)	Sharply Sloping, Mild-Moderate	32 dB	31%	50%	CVC's	CVC
Skinner (1976)	Sharply Sloping, Mild-Moderate	23 dB	72%	84%	High Frequency Word List (c)	Word
Lippmann (1978)	Sharply Sloping, Moderate	31 dB	39%	68%	CVC's	Phoneme
Pascoe (1975)	Flat to Gently Sloping, Moderate	18 dB	62%	73%	High Frequency Word List (c)	Word
Lippmann (1978)	Flat, Moderate	15 dB	41%	53%	CVC's	Phoneme

a) Our estimate of functional gain in the 2-4 kHz region relative to functional gain in the 250-500 Hz region. High-frequency emphasis was roughly constant above 1600 Hz for Barfod, above 2000 Hz for Skinner, and above 800 Hz for Lippmann (flat losses). High-frequency emphasis increased at a slope of roughly 5 dB/octave above 1000 Hz for Pascoe and above 800 Hz for Lippmann (sloping losses).

b) Barfod's system was not exactly orthotelephonic because of the use of an electrically flat system and Beyer DT-48 earphones.

c) See Pascoe (1975).

Table 11. Recent Research on Amplitude Compression

STUDY	LOSSES	BEST COMPRESSION CHANNELS	BEST COMPRESSION SCHEME	HF EMPHASIS OF REFERENCE LINEAR SYSTEM dB (a)	HF EMPHASIS OF REFERENCE LINEAR SYSTEM f_o (b)	SCORES (c) LINEAR	SCORES (c) COMPRESSION	MATERIALS	ITEM SCORED
Villchur (1973)	Flat & Sloping, Moderate-Severe	2	EL (d)	9 dB	400 Hz	59%	74%	CVC's	Phoneme
Yanick (1976)	Flat & Sloping, Mild-Severe	2	EL (d)	9 dB	1500 Hz	55%	88%	Harvard Sentences	Key Word
Lippmann (1977)	Flat & Sloping, Moderate	16	COMCL	0 dB (ORTHO)	-	47%	62%	CVC's	Phoneme
Barfod (1976)	Sloping, Mild-Moderate	4	EL	33 dB	800 Hz	79%	76%	CVC's	Phoneme
Lippmann (1977)	Flat & Sloping, Moderate	16	COMCL	25 dB (OMCL)	800 Hz	68%	62%	CVC's	Phoneme

a) Our estimate of functional gain in the 2-4 kHz region relative to functional gain in the 250-500 Hz region. For Yanick this value has been reduced by 10 dB to account for the effect of using insert receivers.

b) Frequency above which functional gain is roughly flat.

c) Scores are for materials presented in quiet for all but Yanick's study where scores correspond to S/N = 6 dB.

d) The characteristics of these systems were modified by subject adjustments and these systems only roughly restored normal equal loudness contours.

REFERENCES

Barfod, J. (1976). "Multi Channel Compression Hearing Aids," Report No. 11, The Acoustics Laboratory, Technical University of Denmark.

Barfod, J. (1972). "Investigations on the Optimum Corrective Frequency Response for High-Tone Hearing Loss," Report No. 4, The Acoustics Laboratory, Technical University of Denmark.

Boddie, J. R. (1973). "A Real-Time Signal Processor for Simulating a Sound Source in Free Space," unpublished M.S. Thesis Dissertation, Massachusetts Institute of Technology.

Boothroyd, A. (1967). "Developments in Speech Audiometry," paper presented at the Inaugural meeting of the British Society of Audiology.

Boothroyd, A. (1968). "Statistical Theory of the Speech Discrimination Score," J. Acoust. Soc. Am. 43, 362-367.

Bilger, R.C. and Wang, M. D. (1976). "Consonant Confusion in Patients with Sensorineural Hearing Loss," J. Speech Hearing Research 19, 718-748.

Braida, L. D., Durlach, N. I., Lippmann, R. P., and Rabinowitz, W. M. (1978). "Matching Speech to Residual Auditory Function I. Review of Linear Amplification," submitted to J. Speech and Hearing Research.

Chaiklin, J. B. (1967). "Interaural Attenuation and Cross-Hearing in Air-Condition Audiometry," J. Auditory Research 1, 413-424.

Cuddy, D. R. (1972). "A Computer-Controlled Audio Processing System," unpublished M.S. Thesis Dissertation, Mass. Inst. of Technology.

deBruine-Altes, J. C. (1946). "The Symptom of Regression in Different Kinds of Deafness," Ph.D. Thesis Dissertation, University of Groningen, Holland.

DeGennaro, S. (1978). personal communication.

Dunn, K. D. and White, S. D. (1940). "Statistical Measurements on Conversational Speech," J. Acoust. Soc. Am 11, 278-288.

Egan, J. P. (1948). "Articulation Testing Methods," Laryngoscope 58, 955-991.

Fletcher, H. (1952). "The Perception of Speech Sounds by Deafened Persons," J. Acoust. Soc. Amer 24, 490-497.

Fletcher, H. and Steinberg, J. C. (1930). "Articulation Testing Methods," Bell System Tech. J. 8, 806-854.

French, N. R. and Steinberg, J.C. (1947). "Factors Governing the Intelligibility of Speech Sounds," J. Acoust. Soc. Am. 19, 90-119.

Hallpike, C. S. and Hood, J. D. (1959). "Observations upon the Neurological Mechanism of the Loudness Recruitment Phenomenon," Acta Otolaryngol. 50, 472-486.

Hecker, M.H.L. (1973). "A Study of the Relationship between Consonant- Vowel Ratios and Speaker Intelligibility," unpublished Ph.D. Thesis Dissertation, Stanford University.

Hellman, R. P. and Zwislocki, J. (1963). "Monaural Loudness Function at 1000 cps and Interaural Summation," J. Acoust. Soc. Am. 35, 856-865.

Hirsh, I. J., Davis, H., Silverman, S. R., Reynolds, E. G., Eldert, E., and Benson, R. W. (1952). "Development of Materials for Speech Audiometry," J. Speech Hearing Disorders 17, 321-337.

Homan, J. L. (1974). "Computer Control of Speech Compression," unpublished B.S. Thesis Dissertation, Mass. Institute of Technology.

Hood, J. D. and Poole, J. P. (1966). "Tolerable Limit of Loudness: Its Clinical and Physiological Significance," J. Acoust. Soc. Am. 40, 47-53.

IEEE (1969). "Recommended Practice for Speech Quality Measurements," IEEE Trans. Audio and Electro., AU-17, 225-247.

International Standards Organization (1961). "Normal Equal-Loudness Contours for Pure Tones and Normal Threshold of Hearing Under Free Field Listening Conditions," ISO R226.

International Standards Organization (1964). "Standard Reference Zero for the Calibration of Pure-Tone Audiometers," ISO R389.

Kalikow, D. N., Stevens, K. N. and Elliot, L. L. (1977). "Development of a Test of Speech Intelligibility in Noise using Sentence Materials with Controlled Word Predictability," J. Acoust. Soc. Am 61, 1337-1341.

Klatt, D. H. and Stevens, K. N. (1973). "On the Automatic Recognition of Continuous Speech: Implications from a Spectrogram-Reading Experiment," IEEE Transactions on Audio and Electroacoustics AU-21, 210-217.

Krasner, M.A. (1975). "A Simulator of the Cues for Auditory Localization," unpublished M.S.Thesis Dissertation, Mass. Institute of Technology.

Kryter, K. D. (1962). "Methods for the Calculation and Use of the Articulation Index," J. Acoust. Soc. Am. 34, 1689-1697.

Lippmann, R. P., Braida, L.D., and Durlach, N. I. (1978). "Matching Speech to Residual Auditory Function. II. Review of Amplitude Compression," submitted to J. Speech and Hearing Research.

Macrae, J. H. and Brigden, D. N. (1973). "Auditory Threshold Impairment and Everyday Speech Reception," Audiology 12, 272-290.

Martin, D. W., Murphy, R. L., and Meyer, A. (1956). "Articulation Reduction by Combined Distortions of Speech Waves," J. Acoust. Soc. Am. 28, 597-600.

Miller, G. A. and Nicely, P. E. (1955). "An Analysis of Perceptual Confusions Among Some English Consonants," J. Acoust. Soc. Am. 27, 338-352.

Miskolczy-Fodor, F. (1960). "Relation between Loudness and Duration of Tonal Pulses. III. Responses in Cases of Abnormal Loudness Function," J. Acoust. Soc. Am. 32, 486-492.

Morgan, D. E., Wilson, R. H., and Dirks, D. D. (1974). "Loudness Discomfort Level: Selected Methods and Stimuli," J. Acoust. Soc. Am. 56, 577-581.

Owens, E., Benedict, M., and Schubert, E. D. (1972). "Consonant Phonemic Errors Associated with Pure-Tone Configurations and Certain Kinds of Hearing Impairment," J. Speech Hearing Research 15, 308-322.

Pascoe, D. P. (1975). "Frequency Responses of Hearing Aids and Their Effects on the Speech Perception of Hearing-Impaired Subjects," The Annals of Otology, Rhinology and Laryngology, Supplement 23, Vol. 84.

Reed, C. M. and Bilger, R. C. (1973). "A Comparative Study of S/N and E/N ," J. Acoust. Soc. Am. 53, 1039-1044.

Reger, S. N. (1936). "Differences in Loudness Response of the Normal and Hard-of-Hearing Ear at Intensity Levels Slightly above Threshold ," Ann. Otol., Rhinol, Larynogol. 45, 1029-1036.

Sivian, L. J. and White, S. D. (1933). "On Minimum Audible Sound Fields," J. Acoust. Soc. Am. 4, 288-321.

Skinner, M. W. (1976). "Speech Intelligibility in Noise-Induced Hearing Loss: Effects of High-Frequency Compensation," unpublished Ph.D. Thesis Dissertation, Washington University.

Stevens, S. S. and Greenbaum, H. B. (1966). "Regression Effect in Psychophysical Judgement," Perception and Psychophysics 1, 439-446.

Studebaker, G. A. (1967). "Clinical Masking of the Non-Test Ear," J. Speech Hearing Dis. 32, 360-371.

Villchur, E. (1973). "Signal Processing to Improve Speech Intelligibility in Perceptive Deafness," J. Acoust. Soc. Am. 53, 1646-1657.

Wang, M. D., Reed, C. M., and Bilger, R. C. (1977). "A Comparison of the Effects of Filtering and Sensorineural Hearing Loss on Patterns of Consonant Confusions," submitted to J. Speech and Hearing Research.

Williams, C. E. and Hecker, M. H. L. (1968). "Relation between Intelligibility Scores for Four Test Methods and Three Types of Speech Distortion, J. Acoust. Am. 44, 1002-1006.

Yanick, P. (1976). "Effect of Signal Processing on Intelligibility of Speech in Noise for Persons with Sensorineural Hearing Loss," J. American Audiology Society 1, 229-238.

Zwicker, E., Flottorp, G., and Stevens, S. S. (1957). "Critical Band Width in Loudness Summation," J. Acoust. Soc. Am. 29, 548-557.

Zwislocki, J. (1953). "Acoustic Attenuation between the Ears," J. Acoust. Soc. Am. 25, 752-759.

APPENDIX 1. SUBJECT HISTORIES

E.D. FEMALE, AGE 37

Past History

She became aware of a hearing loss in her left ear 7 years ago when upon waking up the left ear felt blocked. The loss at that time was diagnosed as a unilateral sudden loss, possibly caused by a virus. She was not however ill at the time and has had no ear infections or ear aches. One year later the hearing in her right ear had also degenerated (primarily above 1000 Hz). Since then hearing in both ears has slowly been deteriorating. X-rays taken one year ago did not reveal any tumors on the auditory nerve. The loss is accompanied by constant tinnitus in both ears that has a buzzing quality.

Present Status

She wears a behind-the-ear hearing aid in the left ear and has used it for the last two years. She wears it daily and finds it very helpful. She feels that her loss has worsened over the last year, however clinical thresholds have been constant over the past 9 months.

Clinical Diagnosis

Bilateral Sensorineural sudden loss.

I.K. FEMALE, AGE 37

Past History

Evidence of a unilateral hearing loss in her right ear was first obtained in a kindergarten screening examination when she was 5 years old. Since then her loss has been stable and has not been accompanied by tinnitus.

Present Status

She has a flat unilateral loss that has been stable for many years and does not use a hearing aid.

Clinical Diagnosis

Unilateral sensorineural loss, congenital in origin.

F.M. MALE, AGE 47

Past History

He started being aware of hearing loss in his right ear accompanied by nausea and dizziness 9 years ago when the loss was diagnosed as Meniere's disease. Since then the loss has been variable and the attacks of dizziness have become rare and are seemingly prevented by taking antivert. The loss is accompanied by tinnitus which sounds like a steady low-level hissing.

Present Status

The last severe attack of dizziness was 2 years ago. He presently takes antivert when he feels an attack coming. His audiogram has become flat instead of sloping over the past year. Over the past 4 months his loss has been stable. Although hearing in his left ear is beginning to worsen, he gets along well without a hearing aid and doesn't plan to obtain one.

Clinical Diagnosis

Unilateral sensorineural loss caused by Meniere's disease.

E.S. MALE, AGE 60

Past History

His loss first became noticeable 2 years ago when he couldn't easily understand the voices of girls in high school classes that he teaches. Prior to that time he worked as a welder during World War II for 2 years and he worked in a print shop where he was exposed to loud noise for 38 years. The loss was diagnosed as noise-induced 2 years ago. At that time his hearing was measured and found to be a bilateral high-frequency loss and he was fitted with an in-the-ear hearing aid. The loss has been stable over the past 2 years and is accompanied by a steady hissing tinnitus.

Present Status

He has a high-frequency bilateral loss that has been stable over the last 2 years. He has constantly worn an in-the-ear hearing aid in his right ear and finds it helpful.

Clinical Diagnosis

Bilateral sensorineural loss caused by noise exposure.

H.S. MALE, AGE 49

Past History

Although his loss may have been present since birth, he first noticed it when he was a teenager and found it helpful to use the right ear when talking on the phone and to keep people on his right side when talking. The loss gives the impression of being constant since it was first noticed, was diagnosed as sensorineural and measured 3 years ago and has been stable since then. The loss is not strictly speaking unilateral because there is a moderate loss in the left ear and there is a mild loss in the right ear. The loss is not accompanied by tinnitus or dizziness.

Present Status

His loss has been stable over the past 3 years and he does not wear or feel he requires a hearing aid because of the mild loss in his right ear.

Clinical Diagnosis

"Unilateral" sensorineural loss which was possibly congenital.

Appendix 2. Pre-Experiment Clinical Measurements

The air and bone conduction thresholds presented in this section were measured using standard clinical procedures at the Massachusetts Eye and Ear Infirmary. All measurements are relative to ISO (1964) except bone conduction thresholds which are relative to HAIC (1966). Thresholds for FM changed significantly between the time clinical measurements were made and the initiation of the experimental program. The thresholds given in Appendix 3 which were measured immediately before the experimental program are more representative of his loss during the experiments than the clinical measurements. Only a limited number of bone conduction thresholds were measured because prior measurements on many of these subjects had already demonstrated the lack of an air-bone gap. Discrimination scores were obtained using recorded CID W-22 word lists (Ira Hirsh-talker) and SRT's were measured with the audiologist reciting spondees. Discrimination scores were obtained at a number of levels including each subject's MCL. Maximum scores were always obtained at the MCL and these scores are the scores that are reported. All subjects were screened using the Carhart Tone Decay Test and the results of all tests were negative.

Thresholds obtained after the experimental program were very similar to pre-experimental thresholds except as described in the text. (Post-experimental clinical data was not obtained for FM). SRT's were also very similar (average difference = 0.75 dB, maximum absolute difference = 10 dB). Discrimination scores however tended to increase. Average post-experiment scores were 74% words correct and average pre-experiment scores were 62% words correct. This increase may have been caused by the training which subjects received in taking speech tests (inluding practice with the CID W-22 word lists).

Table A2.1 Clinical Speech Reception Threshold

Speech Reception Threshold (dB re ISO, 1964)

Subject	Right Ear	Left Ear
ED	40	15
IK	65	10
FM	40	10
ES	40	10
HS	4	0

Table A2.2. Clinical Speech Discrimination (CID W-22)

	Speech Discrimination (CID W-22)			
	Right Ear		Left Ear	
Subject	Level (dB re ISO)	%Words Correct	Level (dB re ISO)	%Words Correct
ED	70	44	55	64
IK	75	36	45	100
FM	80	68	50	82
ES	65	80	55	88
HS	68	100	64	60

Table A2.3. Clinical Air Conduction Thresholds

		Air Conduction Threshold (dB re ISO, 1964)								
		Frequency (Hz)								
Subject	Ear	250	500	1000	1500	2000	3000	4000	6000	8000
ED	Right	15	30	50		80		80		100
ED	Left	10	10	45		55		75		95
IK	Right	50	55	60		55		60		60
IK	Left	10	10	5		15		10		15
FM	Right	35	30	45		60		70		70
FM	Left	20	10	15		15		50		35
ES	Right	5	-5	10	30	50	60	65	65	60
ES	Left	5	0	0	30	65	75	95	95	90
HS	Right	25	25	25		40		25		30
HS	Left	25	25	35		60		90	110+	90+

Table A2.4. Clinical Bone Conduction Thresholds

		Bone Conduction Threshold (dB re HAIC, 1966)				
				Frequency (Hz)		
Subject	Ear	250	500	1000	2000	4000
ED	Right		30			
ED	Left			45	75+	70+
IK	Right		55	50	50	55
IK	Left		0	20		
FM	Right		45	50	60	65+
FM	Left		5	10	15	45
ES	Right		5	0	45	60+
ES	Left			0	65	70+
HS	Right		20	25	35	25
HS	Left			55	70+	

Table A2.5. Clinical Alternate Binaural Loudness Balances

Subject	Frequency (Hz)	Right Ear HL(ISO, 1967)	Left Ear HL(ISO, 1967)
IK	1000	80	70
IK	1000	100	90
IK	2000	70	70
FM	1000	75	50
FM	1000	90	70
FM	1000	100	95

Appendix 3. Threshold and Loudness Discomfort Level Measurements (a).

Table A3.1. Individual Pre-experimental Threshold Measurements Made at MIT.

Frequency (kHz)	Threshold (Free-Field dB SPL)				
	Subject				
	ED	IK	FM	ES	HS
0.125	34.0	82.0	73.0	17.0	37.5
0.25	13.0	66.0	56.0	9.5	24.0
0.5	8.5	61.0	56.0	0.0	28.0
0.75	12.0	-	-	-	-
1.0	47.0	56.0	56.0	0.0	34.0
1.5	52.5	59.5	67.5	37.5	52.0
2.0	61.0	56.0	72.5	67.5	60.0
3.0	68.0	50.0	70.0	78.0	73.0
4.0	73.0	50.0	71.0	90.5	81.5
6.0	84.0	70.0	81.0	106.0	106.0
8.0	115.0	82.0	75.5	122.7	114.0+

a) All measurements reported in this and in the following appendices were made at the Communications Biophysics Group Laboratory at MIT. These measurements were made in the left ear of ED, the right ear of IK, the right ear of FM, the left ear of ES, and the left ear of HS.

Table A3.2. Individual Pre-Experimental Loudness Discomfort Levels Measured at MIT.

	Loudness Discomfort Level (Free-Field dB SPL)				
Frequency	Subject				
(kHz)	ED	IK	FM	ES	HS
0.125	-	-	113.0	108.0	108.5
0.25	89.0	113+	100.0	98.0	99.5
0.5	96.0	116.5	106.0	100.5	104.0
1.0	108.0	109.5	102.5	106.0	108.5
1.5	108.0	112.5	108.0	117.0	-
2.0	114.5	114.5	107.5	114.0	118.5
3.0	109.0	109.0	103.0	111.5	113.5
4.0	114.0	118+	103.5	121.0+	117.5
6.0	122.0+	122.0+	122.0+	122.0+	122.0+
8.0	124.0+	124.0+	124.0+	124.0+	124.0+

Appendix 4a. Procedures for Monaural Loudness Balances and Alternate Binaural Loudness Balances.

MLB's were measured for subjects with bilateral losses between frequencies seperated by octave or 1/2 octave intervals from 250 to 4000 Hz and then from 4000 Hz to 6000 Hz. Measurements were normally made at one to four levels that were approximately evenly spaced (in dB) between threshold and LDL. Fewer levels were used between frequencies with small threshold differences and at 6000 Hz where measurements were often difficult.

ABLB's were measured for subjects with unilateral losses at frequencies (500, 1000 and 4000 Hz for IK and 500 and 1000 Hz for FM) where the hearing loss in the better ear was less than 10 dB (re ISO, 1961) and interaural attenuation was large enough to isolate the two ears . ABLB's were performed at 3 to 4 levels between threshold and LDL. MLB's were also measured for subjects with unilateral losses in the impaired ear at two levels from 250 to 6000 Hz and one level at 125 and 8000 Hz between frequencies seperated by octave or 1/2 octave intervals (excluding 500-1000 Hz and 2000-4000 Hz for IK and 500-1000 Hz for FM).

All loudness balances were performed using the method of adjustment with a repetitive 300 msec tone burst and 100 msec silent interval automatically alternated between ears or frequencies. An individual match was made by having a subject initially adjust (by a bracketing procedure) the level of a tone at one frequency or in one ear to match the loudness of a tone of fixed level at another frequency or in the opposite ear. The subject then turned both tones off for 5-20 seconds. When the tones were turned on again, if the subject felt that the tones were equally loud, then the match was accepted. If not, then the procedure was repeated.

Subjective bias was minimized in the above balances by training subjects for 1-2 hours and making the angular-position to level transform of the knob used to adjust level of the variable tone 1) linear in dB 2) variable from match to match and 3) permit a minimum adjustment range of 80 phons. The effects of any residual bias and variance in the matches were minimized by performing MLB's between tones that were at most one octave apart, by performing all MLB's between two frequencies and all ABLB's at one frequency at one sitting, and by averaging over four non-sequential matches made with one of the tones fixed and the other variable and four matches made with the fixed and variable tones interchanged. Fixed levels for both tones were chosen before the matches and were usually close to the average of the 4 variable levels (average difference = 0.7dB, standard deviation = 4.9dB).

Final matches were taken to be the average of the fixed and variable levels. The average standard deviation of a single MLB, performed as above, was about 2 dB for all subjects and thus the standard deviation of the average level resulting from four balances was about 1 dB. The average standard deviation of one ABLB was about 2.9 dB in the better and 1.9 dB in the impaired ear and thus the resulting standard deviations of the average levels were about 1.5 dB and 1 dB. The average bias (fixed level minus four level average at one frequency or in one ear corrected by adding the difference between the fixed and four level average at the matching frequency or in the opposite ear) was -1.5 dB with a standard deviation of 5.5 dB. This bias was larger for the ABLB's and tended to increase at 6 and 8 kHz and at extremely high or low levels.

The above procedures were not followed exactly for subject HS because MLB's had been previously obtained in a preliminary study. For HS either MLB's were obtained in a balanced procedure as above with 1 instead of 4 matches or MLB's were obtained by holding the lower frequency tone fixed for two matches and taking the average of the variable levels as the matching level. This introduced more bias and variance into HS's data, however this is somewhat compensated for by the fact that MLB's were obtained at more levels and HS was highly trained in loudness balancing by the time these measurements were made.

Appendix 4b. Individual Monaural Loudness Balances and Alternate Binaural Loudness Balances

MLB and ABLB measurements are given in Tables 4.1-4.7. All levels are free- field sound pressure (dB SPL) as determined by the procedures described in Appendix 4a. Boxes indicate frequencies of tones common to each MLB. MLB's for HS are presented in Table 4.7. Levels with an asterisk in this table are averages obtained after 2 matches with the lower frequency tone fixed and all other levels were obtained as in the other tables but with 2 instead of 3 matches per pair of frequencies.

Table A4.1. ED - Monaural Loudness Balances

	Frequency (kHz) 0.25	0.5	1.0	2.0	4.0	6.0
Level (Free-Field dB SPL)	40.8	41.7				
	68.4	70.1				
		24.2	58.0			
		47.7	65.3			
		65.5	75.1			
		79.2	85.2			
			63.0	75.9		
			72.2	83.2		
			82	90.4		
				77.5	79.6	
				90.2	93.9	
					81.2	97.6

Table A4.2. IK - Monaural Loudness Balances

	Frequency (kHz)						
	0.125	0.25	0.5	1.0	2.0	4.0	8.0
Level	87.1	72.5		70.2	73.2	67.3	93.5
(Free-Field	99.5	91.2		87.7	89.7	80.9	108.8
dB SPL)		79	70.7				
		96.5	92.2				

Table A4.3. IK - Alternate Binaural Loudness Balances

	Frequency (kHz)					
	0.5	0.5	1.0	1.0	4.0	4.0
Ear	left	right	left	right	left	right
Threshold (a)	7.0	61.0	13.0	56.0	1.0	50.0
Level	38.2	78.2	25.2	66.5	36.8	70.2
(Free-Field	59.4	85.4	49.6	72.1	62.8	84.4
dB SPL)	79.9	97.3	67.0	79.1	83.6	102.6
			83.9	89.9		

a) Free-Field dB SPL

Table A4.4. FM - Monaural Loudness Balances

	Frequency (kHz) 0.125	0.25	0.5	1.0	2.0	4.0	6.0	8.0
	98.5	84.7		73.1	79.6			
		69.4	71.9	91.3	94.4			
Level		83.2	88.3		82.8	79.2		
(Free-Field					96.2	92.9		
dB SPL)						81.2	90.8	
						94.2	101.9	
							95.2	85.6

Table A4.5. FM - Alternate Binaural Loudness Balances

	Frequency (kHz)			
	0.5	0.5	1.0	1.0
Ear	left	right	left	right
Threshold (a)	4	56	-5.2	56
Level (Free-Field dB SPL)	33.9	76.0	24.5	73.4
	53.1	81.8	44.4	78.6
	72.0	86.5	69.4	84.3
	92.4	93.0	87.2	89.2

a) Free-Field dB SPL

Table A4.6. ES - Monaural Loudness Balances

	Frequency (kHz)							
	0.25	0.5	1.0	1.5	2.0	3.0	4.0	6.0
	30.6	28.9		45.4	78.9			
	52.3	51.6		63.1	85.9			
	74.8	80.3		76.3	96.8			
Level		25.6	28	88.2	103.6			
(Free-Field		51.4	48.7		80.3	82.1		
dB SPL)		76	80.3		91.8	92.3		
			25.1	53.1		85.0	101.4	
			47.5	65.8		92.1	107.5	
			66.7	81.6			102.9	112.5
			86.8	100.2				

Table A4.7. HS - Monaural Loudness Balances

	Frequency (kHz)					
	0.25	0.5	1.0	2.0	4.0	6.0
	40.3	41.9		71.0	88.0*	
	60.8	59.2		78.0	90.6*	
Level	75.5	76.8		84.0	93.2*	
(Free-Field	88.0	90.0		90.3	97.3	
dB SPL)		40.0	53.3*	98.8	100.3	
		50.0	61.5*		91.8	113.0
		60.0	66.4*		98	117.5
		70.0	72.7*			
		80.0	80.5*			
		90.0	95.0*			
			53	71.0*		
			60.2	73.7		
			66.0	79.0*		
			72.0	82.0		
			82.7	91.3		
			94.0	96.5*		
			101.5	99.1		

* See Appendix 4a

Appendix 5a. Methods used to Determine Equal Loudness Contours from Loudness Balances and to Relate these Contours to Normal Contours.

1) Determine equal loudness contours from loudness balances.

For subjects with bilateral losses equal loudness contours were calculated relative to levels at a reference frequency (250 or 500 Hz) at which the hearing loss was less than 13 dB (re ISO, 1951). These contours were composed of segments that spanned octave or 1/2 octave intervals. These segments were linear interpolations between the MLB's and threshold measurements from 250 Hz to 6000 Hz and were parallel to threshold measurements from 125 to 250 Hz and from 6000 to 8000 Hz.

For subjects with unilateral losses equal loudness contours were calculated relative to levels at the given reference frequencies and extended from these frequencies to frequencies connected by ABLB's and MLB's. Segments of these contours were linear interpolations of the threshold, MLB and ABLB measurements.

2) Relate equal loudness contours to normal.

All equal loudness contours were related to normal via one or more reference frequencies. Although thresholds at these frequencies differed little from "normal" threshold (the differences were -10 to 13 db re ISO, 1933), and were within the normal range of +-15 dB (Sivian and White, 1933), they were not exactly equal to the average "normal" threshold. In order to utilize previously measured "normal" thresholds specified in ISO(1961), the loudness of tones at the reference frequencies for impaired listeners had to be related to the loudness of these tones for normals. This was done as follows. If TN and TI represent the sound pressure levels (free-field dB SPL) corresponding to threshold in the normal and impaired ear, and if LN and LI represent the sound pressure levels (free-field dB SPL) of tones that are equally loud in the normal and impaired ear, then in the region TI LI TI+40, LN=TN+(LI-TI)(TI+40-TN)/40, and in the region LI TI+40, LN=LI. This relationship assumes linear recruitment in the impaired ear (for poorer thresholds in the impaired ear) or linear recruitment in the normal ear (for better thresholds in the impaired ear) that is complete 40 dB above threshold in the impaired ear. This assumption about the shape of the recruitment curve is roughly consistent with the model of recruitment proposed by Hallpike and Hood (1959) and with data presented for small losses by deBruine-Altes (1946) and Miskolczy-Fodor (1960).

Appendix 5b. Individual Phon Equal Loudness Contours

The phon level of an equal loudness contour refers to the level (free-field sound pressure -dB SPL) of a tone at 1000 Hz presented to a normal listener that is equally loud to a tone with frequency and level defined by the equal loudness contour and presented to the impaired listener.

All contours were calculated using procedures described in Appendix 5a. Contours that are above the level at which monaural or binaural loudness balances were measured were calculated by assuming that the loss was purely conductive at these high levels.

Table A5.1. ED - Phon Equal Loudness Contours, Thresholds, and Loudness Discomfort Levels.

	Level (Free-Field dB SPL)							
	Frequency (kHz)							
	.125	.25	.5*	1.0	2.0	4.0	6.0	8.0
Thresh	34	13	8.5	47	61	73	84	115
Phons								
10	37.9	16.9	12.7	50.2	64	74.2	86	116.9
20	44.7	23.7	20:9	55.8	69.2	76.3	89.4	120.3
30	51.6	30.6	29.3	59.6	72.7	77.7	91.8	122.7
40	58.7	37.7	37.9	62.3	75.2	78.7	93.4	124.3
50	66.7	45.7	46.8	65.0	77.5	79.7	95.1	126.0
60	75.9	54.9	56.2	70.0	81.5	84.1	100.5	131.4
70	85.5	64.5	66.0	75.6	85.7	88.8	105.2	136.1
80	95.4	74.4	76.1	82.9	91.3	95.0	111.4	142.3
90	105.9	84.9	86.6	92.6	101.0	104.7	121.1	152.0
100	116.7	95.7	97.4	103.4	111.8	115.5	131.9	162.8
LDL		89	96	108	114.5	114	122+	

* Reference frequency used to relate contours to normal (see text).

Table A5.2. IK - Phon Equal Loudness Contours, Thresholds, and Loudness Discomfort Levels

	Level (Free-Field dB SPL)						
	Frequency (kHz)						
	.125	.25	.5*	1.0	2.0	4.0*	8.0
Thresh	82.0	66.0	61.0	56.0	56.0	50.0	82.0
Phons							
10	84.8	69.6	63.7	60.1	61.0	53.1	84.0
20	89.3	75.8	68.3	66.6	68.8	58.3	87.7
30	92.1	80.9	73.1	68.6	71.2	63.3	90.9
40	95.4	84.9	77.9	70.5	73.5	68.4	94.6
50	97.1	87.5	81.1	72.5	75.3	73.2	100.2
60	98.8	90.1	84.4	76.3	78.9	78.4	106.0
70	102.4	94.1	89.3	81.0	83.4	83.6	111.4
80	107.7	99.4	95.2	87.4	89.4	91.1	119.0
90	116.6	108.3	104.0	96.0	98.0	99.1	126.9
100	127.4	119.1	114.8	106.0	108.0	107.5	135.4
LDL		113+	116.5	109.5	114.5	118.2+	

* Reference frequencies used to relate contours to normal (see text).

Table A5.3. FM - Phon Equal Loudness Contours, Thresholds, and Loudness Discomfort Levels

Level (Free-Field dB SPL)

	Frequency (kHz)							
	.125	.25	.5*	1.0*	2.0	4.0	6.0	8.0
Thresh	73.0	56.0	56.0	56.0	72.5	71.0	81.0	75.5
Phons								
10	75.6	59.0	59.5	60.4	74.2	72.5	82.4	76.4
20	80.2	64.0	65.5	68.1	77.5	75.1	84.9	78.1
30	84.8	69.3	71.7	74.4	80.7	77.6	87.3	79.9
40	88.7	73.7	77.0	77.5	83.1	79.6	89.2	81.3
50	90.8	76.1	79.9	79.9	85.1	81.6	91.1	82.6
60	92.9	78.4	82.6	82.2	87.0	83.5	92.8	83.8
70	94.7	80.4	85.1	84.5	88.9	85.4	94.4	85.0
80	96.8	82.8	87.9	87.2	91.1	87.7	96.3	86.7
90	99.8	86.0	91.1	92.0	95.1	91.8	99.8	90.2
100	106.7	92.9	98.0	102.0	105.1	101.8	109.5	99.9
LDL	113.0	100.0	106.0	102.5	107.5	103.5	122+	

* Reference frequencies used to relate contours to normal (see text).

Table A5.4. ES - Phon Equal Loudness Contours, Threshold, and Loudness Discomfort Levels.

	Level (Free-Field dB SPL)									
	Frequency (kHz)									
	.125	.25*	.5	1.0	1.5	2.0	3.0	4.0	6.0	8.0
Thresh	17.0	9.5	0.0	0.0	37.5	67.5	78.0	90.5	106.0	122.7
Phons										
10	21.7	14.4	6.4	7.0	41.9	73.8	80.0	93.7	107.7	124.4
20	29.8	22.5	17.7	19.5	49.5	80.5	82.3	97.2	109.5	126.2
30	38.3	31.0	29.2	30.9	56.4	83.2	84.7	101.0	111.5	128.2
40	46.9	39.6	38.3	38.2	60.5	84.9	86.1	102.4	112.3	129.0
50	55.9	48.6	47.7	45.7	64.8	87.3	88.3	104.3	113.9	130.6
60	64.9	57.6	58.3	57.7	74.2	95	95.5	110.9	120.5	137.2
70	74.2	66.9	70.3	72.9	87.3	103.1	103.6	119.0	128.6	145.3
80	83.9	76.6	82.1	86.4	99.9	115.3	115.8	131.2	140.8	157.5
90	94.1	86.8	92.3	96.6	110.0	125.4	125.9	141.3	150.9	167.6
100	104.8	97.5	103.0	107.3	120.7	136.1	136.6	152.0	161.6	178.3
LDL	108.0	98.0	100.5	106	117	114	111.5	121+		

*Reference frequency used to relate contours to normal (see text).

Table A5.5. HS - Phon Equal Loudness Contours, Threshold, and Loudness Discomfort Levels

	Level (Free-Field dB SPL)							
	Frequency (kHz)							
	.125	.25	.5*	1.0	2.0	4.0	6.0	8.0
Thresh	37.5	24.0	28.0	34.0	60.0	81.5	106.0	114.0
Phons								
10	39.9	27.4	30.9	38.6	62.7	83.5	107.1	115.1
20	45.9	33.4	36.0	46.9	67.5	86.1	108.9	116.9
30	52.0	39.5	41.2	54.3	71.4	88.1	110.4	118.4
40	58.3	45.8	46.6	58.7	73.1	88.8	110.9	118.9
50	64.9	52.4	52.1	62.5	75.8	89.8	111.6	119.6
60	71.7	59.2	57.8	65.3	78.4	90.8	112.3	120.3
70	79.4	66.9	66.5	70.6	81.3	92.0	113.1	121.1
80	89.1	76.6	78.0	78.9	88.0	95.8	115.4	123.9
90	99.3	86.8	88.7	93.1	96.1	99.3	118.8	126.8
100	110	97.5	99.5	104.5	102.1	103.6	123.1	131.1
LDL	108.5	99.5	104.0	108.5	118.5	113.5	122+	-

* Reference frequency used to relate contours to normal (see text).

Appendix 6. 10% Cumulative Levels of Speech Material

Cumulative levels presented in this appendix were measured with the 16-channel speech processing system described in the companion paper using the detectors of that system. These levels were obtained using 24 sec segments of cafeteria noise or continuous speech consisting of items (syllables and PBNS's) that had been segmented from male and female CVC and sentence lists and abutted. In each channel the detector output level was sampled every 1.4 msec and the 10% cumulative level or simply 10% level was the level in each channel that was exceeded by only 10% of the sampled levels for one 24 sec segment. The 90% level was defined correspondingly.

Table A6.1. 10% Cumulative Levels Measured Using Detectors in 16 Channels of Speech Processor for Speech materials of Experiment I.

Channel Center Frequency (kHz)	10% Cumulative Levels (dB re Overall RMS)		MATERIAL		
	MALE CVC's	FEMALE CVC's	MALE PBNS'S	FEMALE PBNS'S	OVERALL AVERAGE
0.16	2.5	0.0	0.0	-2.0	0.0
0.28	1.5	-5.0	1.0	-3.0	-1.5
0.4	-5.5	0.5	-6.0	-2.0	-3.5
0.5	-5.0	-7.0	-6.5	-3.5	-5.5
0.63	-10.5	-5.0	-9.0	-6.5	-8.0
0.8	-15.5	-7.5	-9.5	-7.5	-10.0
1.0	-17.5	-12.0	-15.0	-11.0	-14.0
1.25	-11.5	-15.0	-11.0	-12.0	-12.0
1.6	-10.5	-17.0	-12.5	-13.5	-13.0
2.0	-14.5	-15.5	-13.5	-16.0	-15.0
2.5	-15.5	-11.0	-19.5	-17.5	-15.5
3.2	-18.5	-13.5	-20.0	-22.0	-18.5
4.0	-16.5	-16.0	-19.5	-23.0	-18.5
5.0	-16.0	-13.0	-18.0	-19.5	-16.5
6.4	-19.0	-11.5	-23.0	-19.0	-18.0
8.0	-28.0	-17.5	-31.0	-26.0	-26.0

Table A6.2. 10% and 90% Cumulative Levels Measured Using Detectors in 16 Channels of Speech Processor for Cafeteria Noise.

Channel Center Frequency (kHz)	Cumulative Levels (dB re Overall RMS) 10%	90%
0.16	-2	-9.5
0.28	-2	-10.5
0.4	-5	-15
0.5	-4	-12
0.63	-4.5	-14
0.8	-6.5	-16.5
1.0	-11.5	-21
1.25	-10.5	-20
1.6	-9.5	-19
2	-12.5	-21
2.5	-13	-21.5
3.2	-15	-23.5
4.0	-17	-25.5
5.0	-21	-31
6.4	-24	-34.5
8.0	-31	-39

Appendix 7. Dunn and White 10% Speech Levels

Table A7.1. Dunn and White 10% Speech Levels in Critical Bands for Speech with an Overall Level of 65 dB SPL.

Center Frequency (kHz)	Critical Bandwidth(a) (Hz)	10% Cumulative RMS Levels(b) (Free-Field dB SPL)
0.16	90	58.5
0.25	95	59
0.4	105	59.5
0.5	110	59
0.63	130	57.5
0.8	150	54
1.0	160	51.5
1.25	200	50
1.6	240	50
2.0	300	48
2.5	400	44.5
3.2	500	42
4.0	700	39
5.0	900	36
6.3	1200	34
8.0	1600	33.5

a) From Zwicker (1957).
b) Obtained by averaging Dunn and White's (1940.) 10% cumulative RMS levels measured in 1/8 second intervals for male and female talkers. These were converted to critical bandwidth measurements using the procedures suggested by Dunn and White and then normalized to an overall speech level of 65 dB SPL.

Appendix 8. Most Comfortable Presentation Levels Chosen for Speech in Experiment I.

Table A8.1. ED - Relative Presentation Levels in Experiment I and Overall Average Output 10% Levels at 500 Hz for each System.

Presentation Level (dB re average over 8 conditions for each system)

Condition			System					
Talker	Mat.	Env.	ORTHO	10%	MA	OMCL	COMCL	EL(a)
Male	PBNS's	Quiet	2.8	4.0	-0.1	1.6	-1.4	1.3
Male	PBNS's	Noise	-1.2	1.0	0.9	0.6	-2.4	-8.7
Male	CVC's	Quiet	0.8	-3.0	0.9	-1.4	0.6	1.3
Male	CVC's	Noise	0.8	-1.0	1.9	3.6	1.6	1.3
Female	PBNS's	Quiet	-0.2	0.0	-0.1	-4.4	1.6	1.3
Female	PBNS's	Noise	-1.2	3.0	0.9	-3.6	0.6	1.3
Female	CVC's	Quiet	2.8	-3.0	-3.1	-2.4	-0.4	1.3
Female	CVC's	Noise	4.2	-1	-1.1	-1.4	-0.4	1.3
Average 10% Output Level for 8 Cond's at 500 Hz (Free-Field dB SPL)			85.2	57.0	37.1	54.4	61.0	49 5

a) These are relative input levels. The overall average free-field RMS input level for the EL system averaged over all conditions was 53.7 dB SPL.

Table A8.2. IK - Relative Presentation in Experiment I and Overall Average Output 10% Levels at 500 Hz for each System.

Presentation Level
(dB re average over 8 conditions for each system)

Talker	Condition Mat.	Env.	SYSTEM ORTHO	10%	MA	OMCL	COMCL	EL(a)
Male	PBNS's	Quiet	4.5	2.0	3.3	1.2	0.8	6.7
Male	PBNS's	Noise	-	-	-	-	-	-
Male	CVC's	Quiet	3.5	4.0	4.3	4.2	4.8	16.7
Male	CVC's	Noise	-0.5	-2.0	-2.7	1.2	-2.2	-3.3
Female	PBNS's	Quiet	0.5	4.0	3.3	3.2	-0.2	-3.3
Female	PBNS's	Noise	-	-	-	-	-	-
Female	CVC's	Quiet	-6.5	-5.0	-5.7	-8.8	-4.2	-13.3
Female	CVC's	Noise	-1.5	-3.0	-2.7	-0.8	-0.8	-3.3
Average 10% Output Level for 8 Cond's. at 500 Hz (Free-Field dB SPL)			82.4	77.0	82.7	72.8	78.8	82.2

a) These are relative input levels. The overall average free-field RMS input level of the EL system averaged over all conditions was 58.3 dB SPL.

Table A8.3. FM - Relative Presentation Levels in Experiment I and Overall Average Output 10% Levels at 500 Hz for each System.

Presentation Level
(dB re average over 8 conditions for each system)

	Condition		SYSTEM					
Talker	Mat.	Env.	ORTHO	10%	MA	OMCL	COMCL	EL(a)
Male	PBNS's	Quiet	2.5	2.2	1.7	1.5	-1.5	0
Male	PBNS's	Noise	-	-	-	-	-	-
Male	CVC's	Quiet	-4.5	-0.8	0.7	-1.5	-0.5	0
Male	CVC's	Noise	-5.5	-3.8	0.7	-1.5	-2.5	0
Female	PBNS's	Quiet	3.5	1.2	-1.3	-1.5	2.5	0
Female	PBNS's	Noise	-	-	-	-	-	-
Female	CVC's	Quiet	2.5	1.2	0.7	3.5	3.5	0
Female	CVC's	Noise	1.5	0.2	-2.3	-0.5	-1.5	0
Average 10% Output Level for 8 Cond's. at 500 Hz (Free-Field dB SPL)			79.5	74.5	74.8	74.3	75.7	77.5

a) These are relative input levels. The overall average free-field RMS input level of the EL system averaged over all conditions was 45 dB SPL.

Table A8.4. ES - Relative Presentation Levels in Experiment I and Overall Average Output 10% Levels at 500 Hz for each System.

Presentation Level
(dB re average over 8 conditions for each system)

Condition			SYSTEM					
Talker	Mat	Env.	ORTHO	10%	MA	OMCL	COMCL	EL(a)
Male	PBNS's	Quiet	7.1	5.7	8.0	4.9	1	+5
Male	PBNS's	Noise	1.1	0.7	2.0	2.9	2	+5
Male	CVC's	Quiet	1.1	0.7	-1.0	3.9	-3	-5
Male	CVC's	Noise	-5.9	-3.3	-2.0	-7.1	-9	-5
Female	PBNS's	Quiet	2.1	0.7	-1.0	5.9	7	+5
Female	PBNS's	Noise	0.1	-2.3	-2.0	-5.1	7	+5
Female	CVC's	Quiet	3.1	-2.3	1.0	-5.1	0	-5
Female	CVC's	Noise	-8.9	-0.3	-5.0	-0.1	-5	-5
Average 10% Output Level for 8 Cond's. at 500 Hz (Free-Field dB SPL)			74.9	62.2	38	70.1	73	56.3

a) These are relative input levels. The overall average free-frield RMS input level of the EL system averaged over all conditions was 60 dB SPL.

Table A8.5. HS - Relative Presentation Levels for the Different Conditions of Experim
ent I
and overall average Output 10% Levels at 500 Hz for each System.

Presentation Level
(dB re average over 8 conditions for each system)

Talker	Condition Mat	Env.	ORTHO	10%	MA	SYSTEM OMCL	COMCL	EL(a)
Male	PBNS's	Quiet	1.5	-0.4	-1.1	1.5	5.0	-1.3
Male	PBNS's	Noise	1.5	0.6	3.9	7.5	-4.0	-1.3
Male	CVC's	Quiet	-6.5	-0.4	-1.1	1.5	-1.0	+8.7
Male	CVC's	Noise	-0.5	5.6	3.9	-1.5	0.0	+8.7
Female	PBNS's	Quiet	3.5	-3.4	-5.1	1.5	-1.0	+8.7
Female	PBNS's	Noise	-0.5	2.6	3.9	-0.5	-1.0	-1.3
Female	CVC's	Quiet	1.5	-4.4	-6.1	-6.5	3.0	-11.3
Female	CVC's	Noise	-0.5	-0.4	1.9	-3.5	-1.0	-11.3
Average 10% output level for 8 cond's. at 500 Hz (free-field dB SPL)			73.5	59.4	53.1	63.5	65.3	61.2

a) These are relative input levels.
The overall average free-field RMS input level of the EL system averaged over all conditions was 66.3 dB SPL.

Appendix 9. 10% Output Levels in Experiment I
For Individual Subjects

Table A9.1. ED - 10% Output Levels of Systems in Experiment I.

10% Output Level (Free-Field dB SPL)*

Frequency (kHz)	SYSTEM					
	ORTHO	10%	MA	OMCL	COMCL	EL
0.16	91	59	41	60	66	66
0.28	89	57	39	58	65	51
0.4	87	56	38	56	63	50
0.5	85	57	37	54	61	40
0.63	83	59	37	52	60	54
0.8	81	61	37	75	72	58
1.0	77	61	72	71	69	63
1.25	79	6	74	73	71	67
1.6	78	73	77	78	77	72
2.0	76	75	85(87)	76	76	76
2.5	75	76	84(90)	75	75	77
3.2	72	75	81(90)	73	78	78
4.0	72	77	81(97)	73	78	79
5.0	74	91	83(111)	75	80	86
6.4	73	102	82(103)	74	79	93
8.0	65	96(123)	74(113)	66	73	95(119)

* Where the levels were equipment limited the required values are given in parenthesis

Table A9.2. IK - 10% Output Levels of Systems in Experiment I

	10% Output Level (Free-Field dB SPL)*					
	SYSTEM					
Frequency (kHz)	ORTHO	10%	MA	OMCL	COMCL	EL
0.16	88	88	88	78	84	97
0.28	86	86	87	77	83	89
0.4	84	81	85	75	81	85
0.5	82	77	83	73	79	82
0.63	80	73	79	70	77	78
0.8	78	70	76	82	86	74
1.0	74	64	71	79	83	70
1.25	77	68	75	80	85	71
1.6	75	69	77	80	82	72
2.0	73	69	73	78	80	73
2.5	72	69	71	78	80	72
3.2	69	68	69	76	75	71
4.0	69	70	69	76	75	71
5.0	71	79	77	78	77	76
6.4	70	86	74	68	69	87

* No levels were equipment limited.

Table A9.3. FM - 10% Output Levels of Systems in Experiment I.

10% Output Level (Free-Field dB SPL)*

	SYSTEM					
Frequency (kHz)	ORTHO	10%	MA	OMCL	COMCL	EL
0.16	85	84	75	80	81	84
0.28	84	75	64	79	79	75
0.4	82	75	70	77	77	76
0.5	80	75	74	75	76	78
0.63	77	74	72	72	74	77
0.8	75	73	70	82	86	76
1.0	71	71	67	78	82	75
1.25	73	76	76	80	84	77
1.6	72	77	81	81	81	79
2.0	70	77	86	79	79	81
2.5	70	79	85	79	79	80
3.2	67	78	83	78	81	79
4.0	67	80	85	78	80	79
5.0	69	85	88	80	82	83
6.4	67	92	88	78	82	87
8.0	59	75	63	70	75	76

* No levels were equipment limited.

Table A9.4. ES - 10% Output Levels of Systems in Experiment I.

10% Output Level (Free-Field dB SPL)*

Frequency (kHz)	SYSTEM ORTHO	10%	MA	OMCL	COMCL	EL
0.16	80	66	47	76	79	60
0.28	79	64	47	74	77	59
0.4	77	63	42	72	75	57
0.5	75	62	38	70	73	56
0.63	72	57	36	68	71	52
0.8	70	53	34	84	83	47
1.0	66	47	32	80	79	43
1.25	68	59	54	82	81	53
1.6	67	71	71	87	82	64
2.0	65	90	97(101)	85	80	86
2.5	65	92	94(108)	84	81	87
3.2	62	97	89(112)	89	87	88
4.0	62	101(110)	88(131)	89	87	99(105)
5.0	64	102(119)	90(133)	91	88	100(108)
6.4	62	103(127)	90(132)	90	88	98(113)
8.0	54	97(133)	84(131)	82	81	90(127)

* Where the levels were equipment limited the required values are given in parenthesis.

Table A9.5. HS - 10% Output Levels of Systems in Experiment I.

10% Output Level (Free-Field dB SPL)*

Frequency (kHz)	SYSTEM ORTHO	10%	MA	OMCL	COMCL	EL
0.16	79	65	51	67	71	74
0.28	78	63	49	66	69	65
0.4	76	61	51	64	67	63
0.5	74	59	53	64	65	61
0.63	71	60	53	65	66	62
0.8	69	62	53	68	77	63
1.0	65	62	52	69	73	64
1.25	67	69	65	75	75	68
1.6	66	74	75	77	83	72
2.0	64	80	80	78	81	77
2.5	64	85	89	83	81	81
3.2	61	89	95	86	87	86
4.0	61	96	98(103)	91	86	90
5.0	63	106(112)	100(113)	95	88	101
6.4	61	105(123)	99(118)	94	88	103(112)
8.0	53	97(118)	91(104)	86	81	97(114)

* Where the levels were equipment limited the required values are given in parenthesis.

Appendix 10. Normalized Functional Gain of Systems in Experiment I For Individual Subjects.

Table A10.1. ED - Normalized Functional Gain of Systems in Experiment I.

Functional Gain (dB re gain at 500 Hz)*

Frequency (kHz)	SYSTEM ORTHO	10%	MA	OMCL	COMCL	EL
0.16	0	-4	-2	0	0	11
0.28	0	-4	-2	0	0	-3
0.4	0	-3	-1	0	0	-2
0.5	0	0	0	0	0	0
0.63	0	4	2	0	1	7
0.8	0	8	4	25	16	14
1.0	0	12	43	25	16	22
1.25	0	17	44	25	16	26
1.6	0	23	47	31	24	31
2.0	0	27	58(60	31	24	36
2.5	0	29	58(64)	31	24	38
3.2	0	31	58(69)	32	30	40
4.0	0	33	58(74)	32	30	43
5.0	0	45	58(86)	32	30	49
6.4	0	57	58(79)	32	31	56
8.0	0	59(87)	58(97)	32	32	66(90)

* Where the gain was equipment limited the required values are given in parenthesis.

Table A10.2. IK - Normalized Functional Gain of Systems in Experiment I.

Functional Gain (dB re gain at 500 Hz)*

			SYSTE			
MS Frequency (kHz)	ORTHO	10%	MA	OMCL	COMCL	EL
0.16	0	5	0	0	0	9
0.28	0	5	0	0	0	3
0.4	0	2	0	0	0	1
0.5	0	0	0	0	0	0
0.63	0	-2	-1	0	1	-1
0.8	0	-3	-2	14	12	-2
1.0	0	-5	-3	14	12	-3
1.25	0	-3	-1	14	12	-2
1.6	0	-1	1	15	11	-1
2.0	0	1	0	15	11	1
2.5	0	2	-2	15	11	1
3.2	0	4	-1	16	10	11
4.0	0	6	-1	16	9	12
5.0	0	133	5	16	9	10
6.4	0	21	11	16	10	18
8.0	0	29	12	16	10	26

* No gain levels were equipment limited.

Table A10.3. FM - Normalized Functional Gain of Systems in Experiment I.

Functional Gain (dB re gain at 500 Hz)*

Frequency (kHz)	SYSTEM					
	ORTHO	10%	MA	OMCL	COMCL	EL
0.16	0	4	-5	0	0	5
0.28	0	-4	-14	0	0	3
0.4	0	-2	-6	0	0	2
0.5	0	0	0	0	0	
0.63	0	2	0	0	1	2
0.8	0	3	0	12	15	4
1.0	0	5	1	12	15	6
1.25	0	8	8	12	15	8
1.6	0	10	14	14	13	11
2.0	0	12	21	14	13	13
2.5	0	14	21	14	13	13
3.2	0	16	22	16	18	13
4.0	0	18	24	16	18	14
5.0	0	21	25	16	18	18
6.4	0	30	26	16	19	22
8.0	0	21	10	16	20	20

* No gain levels were equipment limited.

Table A10.4. ES - Normalized Functional Gain of Systems in Experiment I.

Functional Gain (dB re gain at 500 Hz)*

Frequency (kHz)	SYSTEM					
	ORTHO	10%	MA	OMCL	COMCL	EL
0.16	0	-2	3	0	0	-2
0.28	0	-2	5	0	0	-2
0.4	0	-1	2	0	0	-1
0.5	0	0	0	0	0	0
0.63	0	-3	0	0	0	-2
0.8	0	-5	1	18	14	0
1.0	0	-7	2	18	14	-5
1.5	0	3	22	18	14	5
1.6	0	16	40	24	16	15
2.0	0	37	68(72)	24	17	40
2.5	0	40	66(80)	24	18	42
3.2	0	48	64(87)	32	27	44
4.0	0	52(61)	63(105)	32	27	55(62)
5.0	0	52(68)	63(106)	32	26	54(64)
6.4	0	53(77)	64(107)	32	27	54(71)
8.0	0	55(91)	66(113)	32	28	54(91)

* Where the gain was equipment limited the required values are given in parenthesis.

Table A10.5. HS - Normalized Functional Gain of Systems Used in Experiment I.

Functional Gain (dB re gain at 500 Hz)*

				SYSTEMS		
Frequency (kHz)	ORTHO	10%	MA	OMCL	COMCL	EL
0.16	0	0	-8	-2	0	8
0.28	0	0	-8	-2	-1	0
0.4	0	0	-4	-2	0	0
0.5	0	0	0	0	0	0
0.63	0	3	2	4	1	3
0.8	0	7	4	9	16	5
1.0	0	11	7	14	16	8
1.25	0	16	18	18	16	14
1.6	0	22	29	21	25	19
2.0	0	28	36	24	25	25
2.5	0	36	46	29	25	31
3.2	0	43	55	35	34	37
4.0	0	50	58(63)	40	34	42
5.0	0	58(64)	58(71)	42	34	52
6.4	0	58(76)	58(77)	43	35	56(64)
8.0	0	58(79)	58(71)	43	36	58(74)

* Where the gain was equipment limited the required values are given in parenthesis.

Appendix 11. Compression Ratios for the EL and COMCL Systems in Experiment I.

Table A11.1. Average Compression Ratio in the Region Above Compression Threshold and Below 10% Input Speech Level*.

	COMCL-System					EL-System				
Freq.		Subject					Subject			
(kHz)	ED	IK	FM	ES	HS	ED	IK	FM	ES	HS
0.16	1	1	1	1	1	1.0	5.3	1.6	0.9	1.2
0.28	1	1	1	1	1	1.0	3.9	1.7	1.0	1.3
0.4	1	1	1	1	1	1.0	3.9	1.7	1.0	1.5
0.5	1.1	1.1	1.1	1	1.1	1.0	4	1.7	1.0	1.7
0.63	1.5	1.5	1.5	1	1.5	1.5	4.2	1.6	1.0	2.0
0.8	1.8	1.8	1.8	1	1.8	2.1	4.3	1.4	1.0	2.3
1.0	2.0	2	2	1	2.0	2.6	4.5	1.3	1.0	2.6
1.25	2.3	2.3	2.3	1.1	2.3	3.0	4.5	1.9	1.4	3.1
1.6	2.6	2.3	2.6	1.2	2.6	3.4	4.4	2.6	1.7	3.5
2.0	3	2.3	3	1.6	3	3.7	4.4	3.2	6.5	4.0
2.5	3	2.1	3	2	3	4.8	3.6	3.6	5.6	5.0
3.2	3	1.9	3	2	3	5.8	2.7	3.9	4.7	6.0
4.0	3	1.8	3	2	3	6.9	1.9	4.3	1.0(3.5)	7.1
5.0	3	1.8	3	2	3	5.5	2	4.3	1.0(4.8)	1.0(8.
4)6.4	3	1.9	3	2	3	4.1	2.1	4.2	1.0(6.0)	1.9(9.
8)8.0	3	1.9	33	2	3	1.0(3.9)	2.2	5.7	1.0(5.6)	1.0(9.
2)										

* Where the compression ratios were equipment limited the required values are given in parenthesis.

Appendix 12. Individual Compression Curves of the COMCL and EL Systems in Experiment I.

The compression curves of the COMCL system were of the form shown in Figure A12.1a and those of the EL system were of the form shown in Figure A12.1b. Input levels in these figures are relative to the 10% input speech levels averaged over all materials in Experiment I. In practice the overall RMS input level of each type of material (e.g., anechoic-female-CVC's, reverberant-male-PBNS's) was adjusted to a fixed level at the input to the 16-channel speech processor and the 10% level of individual materials varied slightly from the average levels in a manner that can be determined from Table 6.1.

Compression curves for the COMCL system (Figure A12.1a) in each channel of the 16-channel speech processing system were linear of slope one at input levels below the compression threshold (CT) and then were linear with slope 1/CR above CT. Compression action was active from CT to the upper limit of the detectors in the 16-channel system which was reached at UL. Above UL the signal in each channel was clipped. The values of UL and CT were identical for all subjects and are given in Table A12.1. The values of CR varied over subjects and are given in Appendix 11. Only the

10% output speech levels which are designated 10%(OUT) in Figure A12.1a, and the 10% input speech levels (relative to the overall speech RMS input level and averaged over all Experiment I materials) are required in addition to CT, CR, and UL to completely specify the compression curves of the COMCL system. The 10% input speech levels were identical for all subjects and are presented in Table A6.1. The 10% output speech levels varied from subject to subject and are presented in Appendix 9.

Compression curves for the EL system (Figure A12.1b) in each channel of the 16-channel speech processing system were linear of slope one at input levels below the compression threshold (CT), then were linear with slope 1/CR1 up to an input level B1, and then were linear with slope 1/CR2 above B1. Compressor action was active from CT to the upper limit of the detectors in the 16-channel system which was reached at UL. Above UL the signal in each channel was clipped. The values of UL and CT were identical for all subjects and are given in Table A12.2. The values of CR1, CR2, and B1 varied over subjects and over the effective overall RMS input levels chosen by each subject and reported in Appendix 8. (In practice the RMS input speech level was held fixed and compression characteristics were varied to produce an effective change in this level.) The values of CR1, CR2, and B1 for all effective input levels chosen by each subject are presented in Tables A12.3-A12.7. Also presented in these tables are values of the 10% output speech levels, denoted by 10%(OUT) in Figure A12.1b, which, along with the 10% input speech levels (relative to the overall speech RMS input level and averaged over all Experiment I materials) are required, in addition to CT, CR1, B1, CR2, and UL to completely specify the compression curves of the COMCL system. The required values of the 10% input speech levels were identical for all subjects and are presented in Table A6.1.

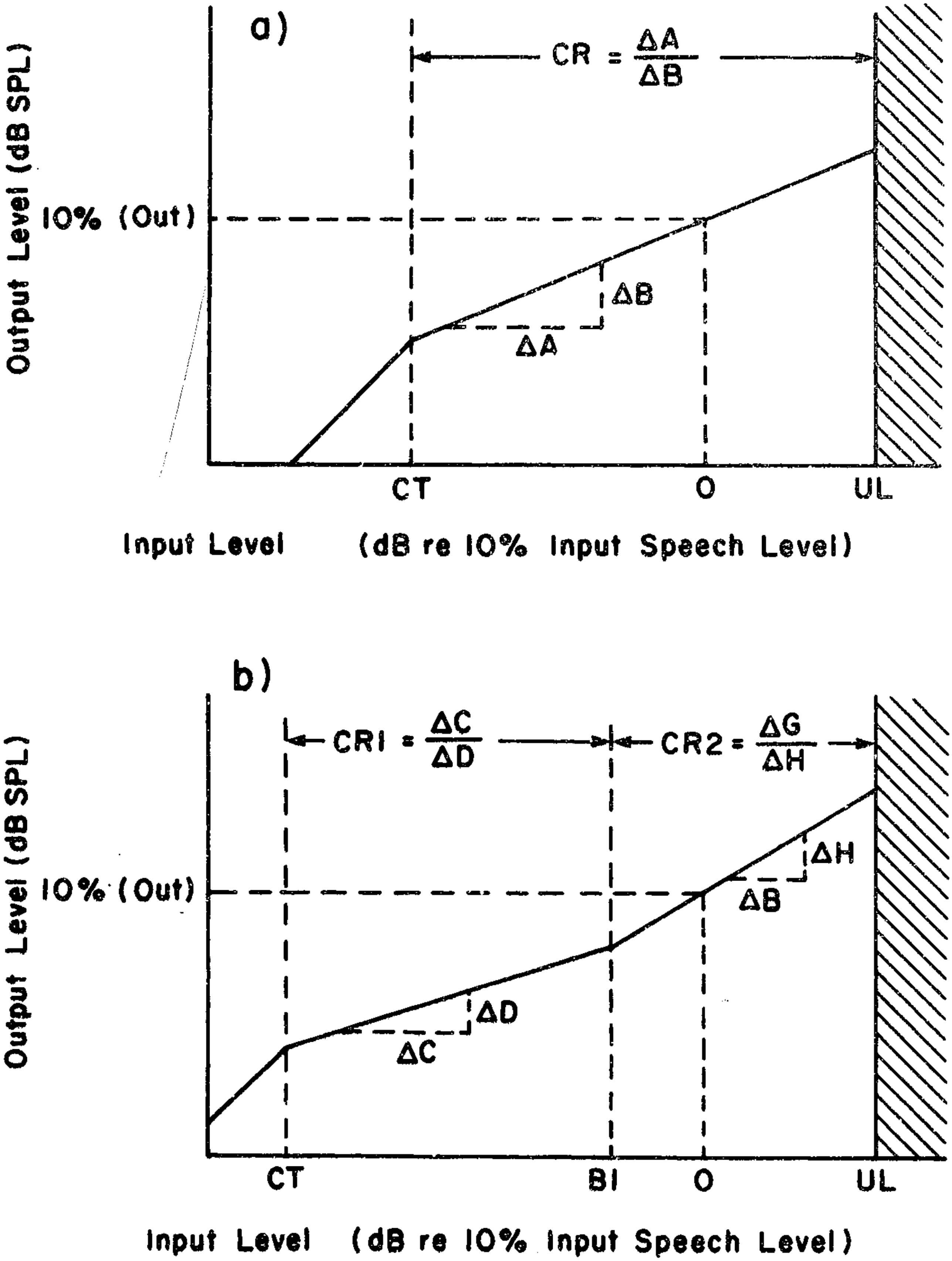

a)
CR = ΔA/ΔB
Output Level (dB SPL)
10% (Out)
ΔB
ΔA
CT
O
UL
Input Level (dB re 10% Input Speech Level)
b)
CR1 = ΔC/ΔD
CR2 = ΔG/ΔH
Output Level (dB SPL)
10% (Out)
ΔH
ΔB
ΔD
ΔC
CT
BI
O
UL
Input Level (dB re 10% Input Speech Level)

FIGURE A12.1

Table A12.1. Compression threshold (CT) and upper limit of compression (UL) for all COMCL systems.

Channel Center Frequency (kHz)	CT Compression Threshold (dB re 10% input speech level)	UL Upper Limit of Compression (dB re 10% input speech level)
0.16	-18	8
0.28	-17	10
0.4	-17	11
0.5	-17	13
0.63	-16	14
0.8	-16	14
1.0	-16	17
1.25	-16	12
1.6	-17	11
2.0	-17	13
2.5	-17	13
3.2	-15	16
4.0	-16	16
5.0	-16	14
6.4	-14	16
8.0	-12	24

Table A12.2. Compression threshold (CT) and upper limit of compression (UL) for all EL systems.

Channel Center Frequency (kHz)	CT Compression Threshold (dB re 10% input speech level)	UL Upper Limit of Compression (dB re 10% input speech level)
0.1	-27	8
0.28	-26	10
0.4	-26	11
0.5	-27	13
0.63	-26	14
0.8	-25	14
1.0	-25	17
1.25	-25	12
1.6	-26	11
2.0	-26	13
2.5	-26	13
3.2	-25	16
4.0	-26	16
5.0	-26	14
6.4	-24	16
8.0	-22	24

Table A12.3. ED - Compression ratios, breakpoints, and 10% output levels of the EL compression system (1,2,3)

	Effective RMS Input Speech Level (Free-field dB SPL)							
	45				55			
Freq. (kHz)	C1	C2	B1	10%(OUT)	C1	C2	B1	10%(OUT)
0.125	1	1	-	57	1	1	-	67
0.25	1	1	-	42	1	1	-	52
0.5	1	1	-	39	1	1	-	50
1.0	1.8	3.1	-6	60	1.8	3.1	-16	64
2.0	2.1	4.9	-5	74	2.1	4.9	-15	76
4.0	6.9	2.1	16	78	6.9	2.1	6	80
6.0	4.1	2	23	90	4.1	2	13	93
8.0	1(3.9)	1(3.9)	-	85(118)	1(3.9)	1(1.9)	-	95(120)

1) See Figure A12.1 for definition of terms and units
2) Values at frequencies not included in the table should be linearly interpolated (log frequency) from the values given.
3) Where the levels were equipment limited the required values are given in parentheses.

Table A12.4. IK - Compression ratios, breakpoints, and 10% output levels of EL compression system (1)

Effective RMS Input Speech Level(Free-field dB SPL)

	45				55			
Freq. (kHz)	C1	C2	B1	10%(OUT)	C1	C2	B1	10%(OUT)
0.125	1.0	5.3	-24	96	5.3	5.3	0	97
0.25	3.9	3.9	1	86	3.9	1.1	6	89
0.5	4.2	2.0	10	79	4.2	2.0	0	83
1.0	1.5	5	-9	69	1.5	5.0	-19	71
2.0	1.2	4.5	-12	72	1.2	4.5	-22	74
4.0	1.9	1.9	18	65	1.9	1.9	18	70
8.0	1.0	2.2	-4	83	1.0	2.2	-14	88

Effective RMS Input Speech Level (Free-field dB SPL)

	65				75			
0.125	5.3	1.0	10	99	5.3	1.0	0	101
0.25	3.9	1.1	6	92	3.9	1.1	-3.5	97
0.5	4.2	2.0	-9	88	2.0	1.0	10	93
1.0	5.0	1.2	15	74	5.0	1.2	5	75
2.0	4.5	1.0	16	77	4.5	1.0	6	80
4.0	1.9	1.0	19	76	1.9	1.0	9	81
8.0	2.2	2.2	26	92	2.2	1.2	21	97

(1) See notes accompanying Table A12.3.

Table A12.5. FM - Compression ratios, breakpoints, and 10% output levels of EL compression system (1).

Effective RMS Input
(Free-field dB SPL)

45

Freq. (kHz)	C1	C2	B1	10%(OUT)
0.125	1.6	4.7	3	84
0.25	1.6	4.3	-3	74
0.5	1.4	3.8	-2	77
1.0	1.3	4.3	-1	75
2.0	3.2	4.8	0	81
4.0	4.3	5.0	18	78
6.0	4.2	5.8	13	86
8.0	5.7	8.0	31	76

1) See notes accompanying Table A12.3.

Table A12.6. ES - Compression ratios, breakpoints, and 10% output levels of the EL compression system (1)

	Effective RMS Input Speech Level (Free-field dB SPL)							
			55			65		
Freq. (kHz)	C1	C2	B1	10% (OUT)	C1	C2	B1	10% (OUT)
0.125	.9	.9	0	54	.9	.9	0	65
0.25	1	1	-	53	1	1	-13	63
0.5	.8	.9	-19	52	.9	.9	5	63
1.0	1	.6	19	38	1	.6	9	48
1.5	1.7	.9	14	61	1.7	.9	2	67
2.0	6.5	.9	12	85	6.5	.9	2	88
3.0	4.7	.9	12	87	4.7	.9	2	88
4.0	1(3.5)	.9	12	94(103)	1(3.5)	.9	2	104(107)
6.0	1(6)	1	17	93(112)	1(6)	1	7	103(114)
8.0	1(5.6)	1(5.6)	26	85(126)	1(5.6)	1	23	95(128)

1) See notes accompany Table A12.3.

Table A12.7. HS - Compression ratios, breakpoints, and 10% output levels of the EL compression system (1).

Effective RMS Input Speech Level (Free-field dB SPL)

	55				65			
Freq. (kHz)	C1	C2	B1	10%(OUT)	C1	C2	B1	10%(OUT)
0.125	1.2	.9	5	64	1.2	.9	5	72
0.25	1.3	1.0	10	56	1.3	1	0	64
0.5	1.7	1.0	8	54	1.7	1		
-1	61							
1.0	1.3	3.4	-5	60	1.3	3.4	-15	63
2.0	2.3	4.0	-17	74	4.0	1.2	18	76
4.0	7.1	7.1	18	89	7.1	2.7	15	90
6.0	1(9.8)	1(9.8)	18	103(111)				
8.0	1(9.2)	1(9.2)	26	97(112)	1(9.2)	1(9.2)	26	97(113)

Effective RMS Input Speech Level (Free-field dB SPL)

	75			
0.125	1.2	.9	-5	82
0.25	1.3	1	-9	74
0.5	1.7	1	-11	71
1.0	3.4	.9	8	66
2.0	4	1.2	8	79
4.0	7.1	2.7	5	92
6.0	1(9.8)	1(3.4)	15	103(112)
8.0	1(9.2)	1(4.2)	26	97(113)

1) See notes accompanying Table A12.3.

Appendix 13. Percent Phoneme Correct Scores and Quality Judgements in Experiment 1

Table A13.1. ED - Percent Phoneme Correct Scores in Experiment I.

	CONDITION					SYSTEM		
Talker	Mat.	Env.	ORTHO	10%	MA	OMCL	COMCL	EL
Male	PBNS	Quiet	31	77	72	71	53	53
Male	PBNS	Noise	21	46	47	52	46	38
Male	CVC	Quiet	42	76	73	79	63	56
Male	CVC	Noise	40	73	70	72	55	57
Female	PBNS	Quiet	43	79	70	68	70	63
Female	PBNS	Noise	13	49	37	41	28	32
Female	CVC	Quiet	51	77	69	75	73	73
Female	CVC	Noise	35	75	71	70	61	56

Table A13.2. IK - Percent Phoneme Correct Scores in Experiment I.

	CONDITION		SYSTEM					
Talker	Mat.	Env.	ORTHO	10%	MA	OMCL	COMCL	EL
Male	PBNS	Quiet	32	36	35	41	28	25
Male	PBNS	Noise	24	9	16	19	18	9
Male	CVC	Quiet	64	63	68	64	54	49
Male	CVC	Noise	-	-	-	-	-	-
Female	PBNS	Quiet	50	40	49	49	56	33
Female	PBNS	Noise	-	-	-	-	-	-
Female	CVC	Quiet	71	68	73	74	67	63
Female	CVC	Noise	61	65	58	57	55	53

Table A13.3. FM - Percent Phoneme Correct Scores in Experiment I.

	CONDITION				SYSTEM			
Talker	Mat.	Env.	ORTHO	10%	MA	OMCL	COMCL	EL
Male	PBNS	Quiet	5	31	26	24	25	9
Male	PBNS	Noise	-	-	-	-	-	-
Male	CVC	Quiet	44	67	70	73	55	59
Male	CVC	Noise	19	38	59	40	23	36
Female	PBNS	Quiet	14	30	27	31	38	18
Female	PBNS	Noise	-	-	-	-	-	-
Female	CVC	Quiet	41	69	65	73	61	51
Female	CVC	Noise	33	52	61	56	48	44

Table A13.4. ES - Percent Phoneme Correct Scores in Experiment I.

	CONDITION		SYSTEM					
Talker	Mat.	Env.	ORTHO	10%	MA	OMCL	COMCL	EL
Male	PBNS	Quiet	61	78	85	79	75	80
Male	PBNS	Noise	20	48	43	44	41	46
Male	CVC	Quiet	53	70	75	75	75	64
Male	CVC	Noise	27	63	57	52	44	51
Female	PBNS	Quiet	60	81	82	80	79	75
Female	PBNS	Noise	32	51	44	41	34	43
Female	CVC	Quiet	54	65	74	78	73	68
Female	CVC	Noise	33	54	56	54	32	43

Table A13.5. Percent Phoneme Correct Scores in Experiment I.

	CONDITION		SYSTEM					
Talker	Mat.	Env.	ORTHO	10%	MA	OMCL	COMCL	EL
Male	PBNS	Quiet	56	88	80	86	82	78
Male	PBNS	Noise	0	64	62	54	62	52
Male	CVC	Quiet	64	81	83	85	73	69
Male	CVC	Noise	42	73	72	78	60	49
Female	PBNS	Quiet	57	81	77	83	76	80
Female	PBNS	Noise	1	56	54	49	46	44
Female	CVC	Quiet	53	80	76	72	78	78
Female	CVC	Noise	37	71	72	71	57	52

Table A13.6. Conversion of numerical values of quality judgements presented in Tables A13.7-A13.11 to scale values.

Scale Value	Numerical Value
Excellent	14
Very Good-Excellent	13
Very Good	12
Good-Very Good	11
Good	10
Fair-Good	9
Fair	8
Moderate-Fair	7
Moderate	6
Poor-Moderate	5
Poor	4
Very Poor-Poor	3
Very Poor	2
Terrible-Very Poor	1
Terrible	0

Table A13.7. ED - Quality Judgements in Experiment I*

	CONDITION		SYSTEM					
Talker	Mat.	Env.	ORTHO	10%	MA	OMCL	COMCL	EL
Male	PBNS	Quiet	4	10	8	10	10	6
Male	PBNS	Noise	8	11	8	12	7	6
Male	CVC	Quiet	11	9	10	13	10	11
Male	CVC	Noise	0	13	6	11	6	5
Female	PBNS	Quiet	10	10	9	11	8	7
Female	PBNS	Noise	2	7	8	11	8	8
Female	CVC	Quiet	1	11	8	11	12	11
Female	CVC	Noise	0	13	10	11	8	9

* See Table A13.6 for meaning of values given.

Table A13.8. IK - Quality Judgements in Experiment I*

	CONDITION				SYSTEM			
Talker	Mat.	Env.	ORTHO	10%	MA	OMCL	COMCL	EL
Male	PBNS	Quiet	8	6	4	6	2	0
Male	PBNS	Noise	-	-	-	-	-	-
Male	CVC	Quiet	10	10	10	10	10	8
Male	CVC	Noise	10	8	6	6	0	8
Female	PBNS	Quiet	10	10	5	8	10	4
Female	PBNS	Noise	-	-	-	-	-	-
Female	CVC	Quiet	10	8	8	12	8	8
Female	CVC	Noise	10	8	8	6	2	4

* See Table A13.6 for meaning of values given.

Table A13.9. FM - Quality Judgements in Experiment I*

	CONDITION		SYSTEM					
Talker	Mat.	Env.	ORTHO	10%	MA	OMCL	COMCL	EL
Male	PBNS	Quiet	0	0	0	0	0	0
Male	PBNS	Noise	-	-	-	-	-	-
Male	CVC	Quiet	8	8	8	8	10	8
Male	CVC	Noise	4	6	4	4	2	4
Female	PBNS	Quiet	0	1	1	1	1	0
Female	PBNS	Noise	-	-	-	-	-	-
Female	CVC	Quiet	8	10	10	10	8	8
Female	CVC	Noise	4	6	6	6	4	4

* See Tale A13.6 for meaning of values given.

Table A13.10. ES - Qualtiy Judgements in Experiment I*

	CONDITION				SYSTEM			
Talker	Mat.	Env.	ORTHO	10%	MA	OMCL	COMCL	EL
Male	PBNS	Quiet	10	12	8	10	10	12
Male	PBNS	Noise	0	8	4	6	6	6
Male	CVC	Quiet	10	8	10	10	8	8
Male	CVC	Noise	0	4	8	4	0	0
Female	PBNS	Quiet	4	10	10	12	14	8
Female	PBNS	Noise	0	2	1	2	1	0
Female	CVC	Quiet	8	9	10	14	10	8
Female	CVC	Noise	4	4	6	4	4	4

* See Table A13.6 for meaning of values given.

Table A13.11. HS - Quality Judgements in Experiment I*

	CONDITION		SYSTEM					
Talker	Mat.	Env.	ORTHO	10%	MA	OMCL	COMCL	EL
Male	PBNS	Quiet	4	8	4	12	8	6
Male	PBNS	Noise	0	10	4	4	6	4
Male	CVC	Quiet	9	10	12	14	12	12
Male	CVC	Noise	4	10	10	10	6	8
Female	PBNS	Quiet	4	8	8	8	6	5
Female	PBNS	Noise	0	10	8	10	8	10
Female	CVC	Quiet	8	11	8	12	14	10
Female	CVC	Noise	4	8	8	12	6	6

* See Table A13.6 for meaning of values given.

Table A13.12. Scores for Sloping Loss Group
Averaged over all Systems.

Mat.	Talker	Env.	Ini.Con	Vowel	Fin.Con	Phonemes
PBNS	Male	Q/A	74	75	71	73
PBNS	Male	N/R	45	51	35	44
PBNS	Female	Q/A	75	76	64	72
PBNS	Female	N/R	43	46	26	38
CVC	Male	Q/A	66	80	63	70
CVC	Male	N/R	54	79	39	57
CVC	Female	Q/A	67	87	57	70
CVC	Female	N/R	56	76	33	55

Table A13.13. Scores for Flat Loss Group Averaged over all Systems.

Mat.	Talker	Env.	Ini.Con	Vowel	Fin.Con	Phonemes
PBNS	Male	Q/A	32	21	26	26
PBNS	Male	N/R	-	-	-	-
PBNS	Female	Q/A	46	33	34	38
PBNS	Female	N/R	-	-	-	-
CVC	Male	Q/A	65	52	65	61
CVC	Male	N/R	47	38	42	42
CVC	Female	Q/A	68	65	62	65
CVC	Female	N/R	57	59	44	53

Table A13.14. Scores for Normals in Quiet (1).

Mat.	Talker	Env.	Ini.Con	Vowel	Fin.Con	Phonemes
PBNS	Male	Q/A	98	96	97	98
PBNS	Male	N/R	89	85	74	83
PBNS	Female	Q/A	98	99	98	98
PBNS	Female	N/R	77	81	63	75
CVC	Male	Q/A	99	100	98	99
CVC	Male	N/R	97	100	91	96
CVC	Female	Q/A	98	99	96	98
CVC	Female	N/R	95	99	85	93

1. The overall speech presentation level was roughly 75 dB SPL for the PBNS lists and 65 dB SPL for the CVC lists.

Appendix 14. Consonant Confusion Matrices

Table A14.1. Flat Loss Subject Group, ORTHO System-
Consonant confusion matrix for CVC's in quiet.

STIM	RESPONSE																
	P	K	T	CH	B	G	D	DZ	F	TH	S	SH	H	V	DH	Z	ZH	TOT
P	11	5	2				1		3	1				2				25
K	1	17	7															25
T			25															25
CH			1	23				1										25
B					14		3		1			1	5	1				25
G			1		4	12	3		2					3				25
D	2				2	1	18		1					1				25
DZ				3			1	21										25
F	4		1						16	2		1		1				25
TH	1	1	2				1		13	4			1	2				25
S			3						1		20	1						25
SH									2		2	21						25
H									2				9	1				12
V					4	1	2						1	17				25
DH					2			1	1					14	7			25
Z							1						8	2	12	2		25
ZH																	13	13
TOTAL	19	23	42	26	26	14	29	24	42	7	22	23	12	54	10	12	15	400

Table A14.2. Flat Loss Subject Group, OMCL System-
Consonant confusion matrix for CVC's in quiet.

STIM.	RESPONSE																	
	P	K	T	CH	B	G	D	DZ	F	TH	S	SH	H	V	DH	Z	ZH	TOT
P	19	4							2									25
K	2	19	4															25
T			24	1														25
CH				25														25
B	3				9	2	6		3					2				25
G		1	1		1	21	1											25
D			2				23											25
DZ			1	5			1	18										25
F			2		1				20	2								25
TH					2				14	9								25
S		1									24							25
SH									3			22						25
H	1												11					12
V					4	1			1				1	18				25
DH						1	4		1					15	4			25
Z									1						y	23		25
ZH												2				1	12	13
TOTAL	25	25	34	31	17	25	35	18	45	11	24	24	12	35	5	24	12	400

Table A14.3. Flat Loss Subject Group, COMCL System-
Consonant confusion matrix for CVC's in quiet.

STIM.	RESPONSE																
	P	K	T	CH	B	G	D	DZ	F	TH	S	SH	H	V	DH	Z	ZH	TOT
P	13	2	8		1	1												25
K	3	12	9	1														25
T	2	1	20	1			1											25
CH			8	16				1										25
B			1		17		2						1	4				25
G					4	14	5	1						1				25
D			1		2	3	19											25
DZ			3	4			3	15										25
F									20			4		1				25
TH			1						15	6	1		1	1				25
S									5	1	16	3						25
SH									9		1	15						25
H									3	1			8					12
V					2	1			2					18	1	1		25
DH					1	1		1		1			2	18		1	1	26
Z														6		17	2	25
ZH														1			12	13
TOTAL	18	15	51	22	27	20	30	18	54	9	18	22	12	50	1	19	15	400

Table A14.4. Flat Loss Subject Group, EL System-
Consonant confusion matrix for CVC's in quiet.

STIM.	RESPONSE																
	P	K	T	CH	B	G	D	DZ	F	TH	S	SH	H	V	DH	Z	ZH	TOT
P	15	4	4		2													25
K	6	14	5															25
T	2	3	19					1										25
CH	1		6	16				1										25
B					17	1	5							2				25
G					5	13	6							1				25
D					7	3	15											25
DZ						3	4	18										25
F									21	3		1						25
TH									19	5				1				25
S									6	1	16	1				1		25
SH									11			12	1					25
H									2	1			9					12
V					1	1								22	1			25
DH						2								22	1			25
Z						1		1						13		8	2	25
ZH														6	1	1	5	13
TOTAL	24	21	34	16	32	24	30	22	59	10	16	14	10	67	3	10	7	400

Table A14.5 Sloping Loss Subject Group, ORTHO System-
Consonant confusion matrix for CVC's in quiet.

STIM.	RESPONSE																	
	P	K	T	CH	B	G	D	DZ	F	TH	S	SH	H	V	DH	Z	ZH	Tot
P	19	6	8	2			1			1								37
K	5	23	8				1											37
T	9	11	16	1														37
CH		2	12	21						1	1							37
B					20	3	11							2			1	37
G			1		4	22	9							1				37
D					9	8	20											37
DZ			1		1	3	8	20						3			1	37
F			3						22	2	6		2	1	1			37
TH		1							15	12	7	1	1					37
S			2						14	4	15	1	1					37
SH				2					4	1	6	22	2					37
H													19					19
V					1	2	5	1					1	20	4	2	1	37
DH					2	1	3							19	6	5	1	37
Z						1	3							17	1	13	2	37
ZH							1				1			3	1	3	9	18
Total	33	43	53	26	37	41	61	21	55	21	36	24	26	66	13	23	15	592

Table A14.6. Sloping Loss Subjject Group, OMCL System-
Consonant confusion matrix for CVC's in quiet.

STIM	RESPONSE																
	P	K	T	CH	B	G	D	DZ	F	TH	S	SH	H	V	DH	Z	ZH	Tot
P	31	2	2		1					1								37
K	3	31	3															37
T	2	9	26															37
CH	1		1	35														37
B					25	1	7							4				37
G						34	2									1		37
D						11	25							1				37
DZ								35							1		1	37
F	1								30	1	3	1	1					37
TH									28	5	3	1						37
S									11	1	24		1					37
SH			1						1			35						37
H													19					19
V							1	3						30		3		37
DH					1									24	6	6		37
Z							1							16		15	5	37
ZH																1	17	18
TO	38	42	33	35	27	46	36	38	70	8	30	37	21	75	7	25	23	592

Table A14.7. Sloping Loss Subject Group, COMCL System-
Consonant confusion matrix for CVC's in quiet.

STIM.	RESPONSE																
	P	K	T	CH	B	G	D	DZ	F	TH	S	SH	H	V	DH	Z	ZH	TOT
P	27	3	7															37
K	4	22	10	1														37
T		10	27															37
CH			4	31						2								37
B					26	1	7							3				37
G						33	3						1					37
D					1	8	28											37
DZ							2	35										37
F									23	4	5	2	3					37
TH									18	5	12	2						37
S									6	5	21	3	2					37
SH									3	2	1	31						37
H													19					19
V					1		4	1						29			2	37
DH					1									27	4	5		37
Z								1	1					8	6	20	1	37
ZH														1		3	14	18
TOTAL	31	35	48	32	29	42	44	37	51	18	39	38	25	68	10	28	17	592

Table A14.8. Sloping Loss Subject Group, EL System-
Consonant confusion matrix for CVC's in quiet.

STIM.	P	K	T	CH	B	G	D	DZ	F	TH	S	SH	H	V	DH	Z	ZH	TOTAL
	RESPONSE																	
P	26	4	7															37
K	3	14	16	4														37
T	1	4	32															37
CH		5	15	14				2		1								37
B					27		9							1				37
G			1			23	6	6						1				37
D						3	32	2										37
DZ						1	4	32										37
F		1	1						25	4	4	1	1					37
TH			1	1					20	6	6	2	1					37
S									4	4	20	9						37
SH			1	1					3	2	7	24						37
H													19					19
V									1					31	1	4		37
DH							3			1				21	4	6	2	37
Z								1			1		1	4	3	24	3	37
ZH														3		2	13	18
TOTAL	30	28	74	20	27	27	54	43	53	18	38	35	22	61	8	36	18	592

Appendix 15. Most Comfortable Levels in Experiment II

Table A15.1. Relative presentation levels in Experiment II.

Relative overall RMS Presentation Level (dB re overall RMS levels in Experiment I averaged over all materials for each subject and system).

Speech Test

Sub.	System	CID W-22	Harv. Sent.	SPIN Quiet	SPIN Noise
ED	ORTHO	1.1	1.3	0.8	5.8
ED	OMCL	-4.1	-5.9	-2.4	-4.4
ED	COMCL	-2.1	-3.9	-0.4	2.6
IK	ORTHO	-3.4	1.8	1.3	2.3
IK	OMCL	-3.8	-1.6	-0.1	2.9
IK	COMCL	-1.2	-4.0	-4.5	2.5
FM	ORTHO	1.8	7	5.5	3.5
FM	OMCL	5.8	6	5.5	5.5
FM	COMCL	3.8	2	0.5	3.5
ES	ORTHO	5.6	10.6	10.1	7.1
ES	OMCL	1.4	6.4	4.9	7.9
ES	COMCL	3.5	7.5	5	5
HS	ORTHO	-0	3.8	5.5	2.5
HS	OMCL	-3	-1.5	3.5	3.5
HS	COMCL	3.5	6.5	5	3

Appendix 16. Word and Phoneme Scores and Quality Judgements in Experiment II.

Table A16.1. Percent phoneme correct scores in Experiment II.

		Speech Test					
Sub.	System	CID W-22	Harv. Sent.	SPIN-PL Quiet	SPIN-PL Noise	SPIN-PH Quiet	SPIN-PH Noise
ED	ORTHO	76	91	59	45	77	68
ED	OMCL	87	99	87	68	100	99
ED	COMCL	92	97	94	71	100	91
IK	ORTHO	80	86	68	41	87	71
IK	OMCL	80	81	67	34	87	71
IK	COMCL	71	70	73	49	93	53
FM	ORTHO	32	32	44	30	41	31
FM	OMCL	66	63	63	40	85	70
FM	COMCL	73	64	65	31	82	59
ES	ORTHO	82	95	89	51	100	79
ES	OMCL	94	100	95	71	99	95
ES	COMCL	95	100	97	69	100	97
HS	ORTHO	84	96	79	65	95	89
HS	OMCL	97	99	97	77	100	100
HS	COMCL	95	99	98	59	97	88

Table A16.2. Percent word correct scores in Experiment II.

		Speech Test					
Sub.	System	CID W-22	Harv. Sent.	SPIN-PL Quiet	SPIN-PL Noise	SPIN-PH Quiet	SPIN-PH Noise
ED	ORTHO	50	85	28	16	68	56
ED	OMCL	66	98	64	40	100	96
ED	COMCL	92	97	84	40	100	88
IK	ORTHO	66	83	44	16	80	48
IK	OMCL	62	76	36	8	80	64
IK	COMCL	54	64	40	16	88	32
FM	ORTHO	20	27	20	8	36	12
FM	OMCL	62	76	36	8	80	64
FM	COMCL	46	54	32	12	76	44
ES	ORTHO	60	90	76	8	100	64
ES	OMCL	86	100	76	8	100	64
ES	COMCL	92	98	88	28	100	96
HS	ORTHO	68	94	40	28	92	84
HS	OMCL	90	99	96	52	100	100
HS	COMCL	86	98	92	32	96	88

Table A16.3. Quality judgements in Experiment II*

				Speech Test	
Sub.	System	CID W-22	Harv. Sent.	SPIN Quiet	SPIN Noise
ED	ORTHO	9	10	7	5
ED	OMCL	10	11	12	11
ED	COMCL	10	12	13	10
IK	ORTHO	9	8	10	4
IK	OMCL	10	10	8	0
IK	COMCL	10	8	10	2
FM	ORTHO	-	0	6	4
FM	OMCL	-	2	6	4
FM	COMCL	-	4	10	4
ES	ORTHO	8	8	10	2
ES	OMCL	14	12	14	6
ES	COMCL	12	14	14	6
HS	ORTHO	6	11	9	6
HS	OMCL	14	14	14	14
HS	COMCL	14	12	14	2

* See Table A13.6 for meaning of values given.

Appendix 17. Phoneme Scores and Most Comfortable Levels in Experiment III.

Table A17.1. Most comfortable levels in Experiment III.

Subject	System	MCL (dB re MCL in Experiment I for each subject with female CVC's in quiet*)
ED	OMCL	-3
ED	COMCL	-2
IK	ORTHO	7
IK	COMCL	3
ES	OMCL	14
ES	COMCL	4
HS	OMCL	2
HS	COMCL	-3

*See Appendix 8.

Table A17.2. Percent Phoneme Correct Scores in Experiment III.

Subject	System	Level(dB re MCL*) 0	-8	-16	-24
ED	OMCL	79	73	45	51
ED	COMCL	79	80	83	57
IK	ORTHO	73	72	53	43
IK	COMCL	77	79	71	62
ES	OMCL	81	81	77	71
ES	COMCL	72	75	80	82
HS	OMCL	77	75	67	47
HS	COMCL	72	77	73	60

*See Table A17.1

PART 3

FREE-FIELD CALIBRATION OF A NEW CIRCUMAURAL EARPHONE

R. P. Lippmann

Research Laboratory of Electronics
Massachusetts Institute of Technology
Cambridge, Massachusetts 02139

ABSTRACT

A new circumaural earphone was calibrated relative to free-field using probe-microphone measurements at low frequencies and threshold measurements at high frequencies. The earphone was found to have a relatively flat response from 0.3 to 6 kHz. The intersubject standard deviation of this calibration had an average value of roughly 1.2 dB below 1 kHz and roughly 2.1 dB from 1 to 6 kHz when variation caused by measurement errors was factored out. These results indicate that for many applications the earphone system can be used as a substitute for free-field sound presentation.

INTRODUCTION

The free-field or real-ear calibration of an earphone requires measurement of the free-field sound pressure level of a plane progressive wave at a subject's head-center position (without the subject present) that produces an eardrum sound pressure equal to that produced by the earphone with a given applied voltage. This type of calibration is important because it allows conventional psychoacoustic measurements of threshold, equal loudness contours, loudness discomfort level, etc. performed using earphones to be compared to measurements made in an anechoic chamber using a free-field sound presentation reference condition (e.g., Sivian and White, 1933; Robinson and Dadson, 1956; Morgan and Dirks, 1974). It also allows the frequency-gain characteristic of a speech processing system that uses a calibrated earphone to be specified relative to the important orthotelephonic reference condition (e.g., see French and Steinberg, 1947; Pascoe, 1975; Killion, 1976). Free-field calibration of an earphone is meaningful and useful to others, however, only if its intra- and intersubject variability is small.

The goal of minimizing the intra- and intersubject variability of the free-field calibration of any sound source (including earphones) places constraints on the design of that source. In order to minimize intrasubject variation it is requried that the sound source be capable of being positioned and repositioned with little variation in sound transmission to the eardrum. In order to minimize intersubject variation it is required that the acoustic path between the sound source and eardrum include, unaltered, those physical structures which significantly affect free-field sound transmission and vary between subjects. This is necessary because the intersubject standard deviation of free-field sound transmission to the eardrum which presumably is caused by variations in the geometry of the external ear is fairly large (Shaw [1974] reports that it is 1 dB or less below 500 Hz and rises to 5 dB above 5 kHz). Insert hearing-aid receivers coupled to the ear through tubing and an earmold do not meet the second of the above requirements because they bypass the normal free-field transmission path except for a small part of the ear canal (e.g., Killion, 1976). Earphone systems utilizing MX41/AR cushions do not meet the second requirement because they alter the shape of the entrance to the ear canal and concha and change their transmission characteristics (e.g., Villchur, 1969). Such systems also do not meet the first requirement because of acoustic leaks which cause large variations in low-frequency transmission characteristics (Villchur, 1970; Shaw, 1966).

Shaw and Thiessen (1962) and Shaw (1966) suggested and demonstrated that circumaural earphones could meet the above requirements better than most presently available supraaural earphones. Shaw and Thiessen (1962) also described a circumaural earphone designed to meet both of the above requirements. Villchur (1970) designed a circumaural earphone mounting for the standard TDH-39 driver. He demonstrated that the free-field calibration of the earphones had intra- and intersubject variation that was acceptable for most subjective measurements and was smaller than the variation obtained with an MX41/AR cushion mounted on a TDH-39 driver, especially below 1 kHz. However, the usefulness of this calibration is limited because the reference condition is different from that which is most commonly used to specify free-field or orthotelephonic reference responses.[1] Also, above 1 kHz calibration and measurement of intersubject variation was based on only three subjects. The purpose of the research described in this paper was to calibrate a circumaural earphone that was similar in design to that described by Villchur and to determine the frequency region over which the calibration is meaningful by measuring the intra- and intersubject standard deviation of that calibration.

I. SUBJECTS AND EQUIPMENT

Eleven subjects (7m, 4f) with normal hearing ranging in age from 18 to 29 years took part in these experiments. All measurements were made in an anechoic chamber in which no noise that could affect free-field thresholds was perceptable. A 2-inch radius loudspeaker was used as a free-field sound source and an individualized dental bite plate was used to fix each subject's head position for field measurements made using this loudspeaker. A probe microphone similar to that described by Villchur and Killion (1975) but with no tubing on the microphone port was used to make probe-microphone measurements.

A calibrated B&K 1/2-inch condenser microphone was used to measure all free-field and coupler sound pressure levels. For the field and probe-microphone measurements pure tones were generated and measured using an HP 302A distortion analyzer. All frequencies were set to better than ±0.2% using the HP5302A counter.

The earphone system was calibrated before and after the experiments on an extended NBS-9A coupler (Charan et al, 1965) and the TDH-39 driver was calibrated on an ASA Type 1 coupler. Free-field sound pressure was measured periodically during the two weeks of experiments.

II. METHOD

The earphone used in this study consisted of a recently purchased TDH-39 driver in a mounting similar to that described by Villchur (1970).[2] The free-field reference source was a loudspeaker located approximately 1 meter in front of each subject's head-center position. The relationship between sound pressure produced by this loudspeaker at the subject's head-center position (without the subject present) and voltage on the loudspeaker was determined by simple objective measurements. Free-field calibration then involved measuring voltage on the loudspeaker and voltage on the TDH-39 driver when both loudspeaker and earphone produced the same eardrum sound pressure. At low frequencies (100 Hz to 1 kHz), where the sound field under the earphone is relatively uniform (Shaw, 1974b) and eardrum sound pressure can be accurately estimated by placing the port of a probe microphone in the concha and using procedures similar to those suggested by Villchur and Killian (1975), we based our calibration on probe tube measurements. At high frequencies (1-8 kHz), where the earcanal sound pressure is very nonuniform (Shaw, 1974b) and accurate estimation of eardrum sound pressure by means of a probe microphone requires that the probe be placed near the eardrum (a difficult and possibly dangerous procedure), we based our calibration on psychophysical threshold measurements. Calibration based on these procedures may be used to reference all earphone above-threshold measurements from 125 Hz to 8 kHz and all earphone threshold measurements above 500 Hz to free-field sound pressure. This calibration may not, however, apply exactly to earphone threshold measurements made in quiet below 500 Hz. These measurements may be 1-6 dB higher than thresholds measured in a free-field because of earphone-induced physiological noise (Villchur, 1970; Shaw, 1974b).

Thresholds were measured monaurally with the opposite ear sealed with plugs[3] that provided at least 20 dB of attenuation at 500 Hz. Threshold was defined as the level at which 75% correct was obtained in a 2IFC experiment with feedback in which one interval contained a 500-msec tone pulse with 25 msec rise and fall times and the other contained silence. Thresholds were measured manually by an adaptive procedure, designed to measure the 75% correct level, which used 1 to 5 dB step sizes with 2 to 8 trials per step.

Field and earphone threshold measurements at one frequency were made sequentially in the same session. In order to estimate the intrasubject variability of thresholds, field and earphone thresholds of three subjects were measured twice at 4 kHz and 8 kHz. All probe-microphone measurements were made in one session. Each series of measurements involved 1) taping the microphone in position, 2) making 9 field measurements, 3) making 9 earphone measurements, and 4) repeating at least five field measurements. If the repeated measurements were within ±1 dB of the initial measurements and if a visual check indicated that the probe-microphone hadn't changed position during the measurements, then the initial measurements were accepted. Otherwise, the above procedures were repeated. The additional field measurements were used to estimate the intrasubject variability of field probe-microphone measurements. In order to estimate the intrasubject variability of earphone probe-microphone measurements, repeat earphone measurements were made with four subjects at five frequencies after the repeat field measurements.

Each subject participated in two to three sessions lasting from two to three hours each. At all frequencies data was obtained from at least eight subjects. Both measurement procedures (probe-microphone and threshold) were used at 500, 750, and 1 kHz as a double check.

III. RESULTS AND DISCUSSION

A. Probe Microphone and Threshold Measurements

Average probe-microphone and threshold measurements are presented in Figures 1 and 2. Figure 1 shows field measurements made using the loudspeaker as a sound source and Figure 2 shows measurements made using the earphone as a sound source. In both figures the probe-microphone measurements are not corrected for the low-frequency roll-off of the probe microphone-preamplifier system. The average field threshold measurements in Figure 1 are similar to the monaural field thresholds measured by Sivian and White in 1933 (the average difference from 1 kHz to 8 kHz is less than 1 dB). Also, the average earphone thresholds in Figure 2 (when referenced to flat-plate coupler measurements) are similar to thresholds measured by Michael (1976) except they are lower by about 6 dB over the frequency range tested. These differences are probably due to the different procedures used in the two studies.[4]

In order to compare the two procedures used to calibrate the earphone system, both threshold and probe-microphone measurements were made at 500 and 750 Hz using four subjects and at 1 kHz using eight subjects. Earphone calibration was then determined using both the threshold and probe microphone measurements. The calibration differencies were generally small (1.7, -1.3 and 2 dB at 500, 750, and 1000 Hz, respectively). Final calibration was based on probe-microphone measurements at frequencies up to and including 800 Hz and on threshold measurements at frequencies above 800 Hz.

The intrasubject standard deviation of the probe-tube measurements below 1 kHz was 0.35 dB for the earphone source (repeated placement of the earphones on the subject's head) and 0.3 dB for the loudspeaker source (repeated placement of the subject in the sound field using the bite plate). Thus, the intrasubject standard duration for the earphone calibration below 1 kHz (based on two probe-tube measurements -- one with the earphone and one with the loudspeaker) was approximately 0.45 dB. The intrasubject standard duration of the threshold measurements at both 4 and 8 kHz was 2.0 dB for both the earphone and loudspeaker sources. Thus the standard deviation of the earphone calibration at these frequencies (based on two threshold measurements -- one with the earphone and one with the loudspeaker) was roughly 2.8 dB.

The intersubject standard deviation of the probe-microphone measurements in Figures 1 and 2 averaged 1 dB for field measurements and 0.8 dB for earphone measurements. The intersubject standard deviation of the thresholds in Figures 1 and 2 averaged 6.0 dB for field measurements and 5.6 dB for earphone measurements.

B. <u>Free-Field Calibration</u>

Calibration of the earphone system presented in terms of the equivalent free-field sound pressure produced by a constant voltage on the earphone is shown in Figure 3a. This curve is obtained by subtracting the curve in Figure 1a from the curve in Figure 2a and adding a constant, and by subtracting the curve in Figure 2b from the curve in Figure 1b and adding a different constant. The constants were chosen such that the reference earphone voltage for the calibration curve is 1 volt. Points on the curves in Figures 1, 2, and 3a have been connected with straight lines for clarity; however, these lines may not provide accurate interpolations above 5-6 kHz.

The relative flatness of the calibration curve from 300 Hz to 6 kHz indicates that over this frequency region the earphones present a coupling situation similar to free-field conditions. The falloff of about 15 dB between 300 Hz and 100 Hz is caused by acoustic leaks under the earphone cushion and also by an effective increase in volume at low frequencies created by a controlled leak from the front to the rear cavity of the earphone mounting. To a first approximation, this earphone system can be modified to have a flat response relative to free field by the addition of a filter with a 12-dB-per-octave falling slope between 100 Hz and 300 Hz. The intersubject standard deviation of the calibration curve is given in Figure 3b.

These standard deviation values can be used to estimate the <u>repeatability</u> and <u>validity</u> of the calibration curve. By repeatability we mean how much curves measured on similar groups of subjects differ and by validity we mean whether or not sound transmission between earphone and eardrum includes, unaltered, those physical structures which significantly affect free-field sound transmission and vary between subjects. Repeatability depends on number of subjects and on measurement procedures and their associated errors, while validity depends only on the earphone.

The repeatability of the calibration curve is determined from the measured intersubject standard deviation of the calibration curve and the number of subjects used to arrive at the average curve. The average of the standard deviation values below 1 kHz is approximately 1.2 dB. This value is small because of the reliable seal provided by the fluid-filled cushion, the controlled leak from the front to the rear cavity of the earphone mounting (Villchur, 1970), and the fact that at low frequencies variations in dimensions of the external ear have little effect on sound transmission to the eardrum. At and above 1 kHz the intersubject standard deviation tends to increase with frequency and has an average value of about 4 dB. Since measurements were made on at least eight subjects at each frequency, the standard deviation of the curve in Figure 3a has an average value of about 0.4 dB below 1 kHz and 1.4 dB from 1 to 8 kHz.

The question of the validity of the calibration curve is addressed by examining the "real" intersubject variation of the calibration curve caused by differences between free-field and earphone sound transmission to the eardrum and not by intrasubject variation in probe-microphone or threshold measurements. The "real" intersubject standard deviation can be calculated from the measured standard deviation if we assume that the "real" and measurement errors are independent.[5] The "real" intersubject standard deviation below 1 kHz is then found to have an average value of 1.2 dB. This small value clearly demonstrates the validity of free-field calibration below 1 kHz. From 1 kHz to

8 kHz the "real" intersubject standard deviation varies from 0 dB at 2 and 4 kHz to 5.4 dB at 3 kHz and 5.9 dB at 8 kHz and has an average value of about 2.7 dB. The variability in the estimated "real" standard deviation of the calibration curve makes it difficult to determine the validity of the calibration curve at frequencies from 1 to 8 kHz. Another measure which gives some insight into this question is the correlation between field and earphone threshold measurements. This correlation is presented in Table I where it can be seen that it remains high, up to 4-6 kHz and then falls to 0.1 at 8 kHz. The high correlation values are most likely caused by earphone and free-field sound transmission to the eardrum being affected by the same physical differences between subjects. The low correlation of 0.1 at 8 kHz indicates that at this frequency sound transmission from the earphone and in free field is determined by different factors. Free-field calibration at 8 kHz is still useful, but it has a "real" intersubject standard deviation of about 6 dB. At 6 kHz the correlation is 0.53 and the "real" intersubject standard deviation of the calibration curve falls to about 2.7 dB. Taken together, these values demonstrate the validity of calibration at 6 kHz. Below 6 kHz the correlation values are high or the standard deviations low except at 3 kHz. Here two of the 10 subjects tested accounted for about 80% of the variance of the calibration curve and if their threshold measurements are excluded, the correlation rises to 0.91. The correlation combined with the "real" intersubject standard deviation of the calibration curve thus indicate that free-field

sound transmission is similar to sound transmission with the earphone and thus that free-field calibration is valid at frequencies up to and including 6 kHz.

Another calibration curve of the earphone system is given in Figure 4. This curve represents the equivalent free-field sound pressure produced by the earphone system relative to the sound pressure produced on an extended NBS-9A flat-plate coupler. It was derived from coupler measurements and the curve in Figure 3a. The dip in this curve between 2 and 8 kHz is a coupler artifact that can also be observed as a peak in the coupler response curves presented by Charan et al (1965). The flatness of the curve below 500 Hz demonstrates that the seal between the fluid-filled cushions and the flat-plate coupler is maintained when the earphones are worn on the head. The intersubject standard deviation of this curve is equal to the standard deviation of the curve in Figure 3a.

FOOTNOTES

1 The loudspeaker used was "located 25 cm from the subject's right ear and turned about 30° toward the front" and the reference pressures to which eardrum pressures were compared were measured with a "1/4-inch microphone lying against a molded rubber inset made to fit the pinna cavity and to form a plane joining the edges" (see Villchur, 1969, especially Figure 4).

2 This mounting was a prototype of the Telephonics Model 556 Headset and was provided by Telephonics. It may not, however, be representative of any production version of the 556 Headset.

3 These were designed and provided by A. W. Mills and described in U.S. Patent #3,737,929.

4 Michael (1976) used an open response ascending technique. This study used a descending technique with marked stimulus and response intervals and feedback. Robinson and Watson (1972) and Green (1972) report that differences in thresholds of 1-5 dB may be caused by adding feedback, marking stimulus intervals, or by using ascending instead of descending procedures.

5. $\sigma(\text{real}) = [\sigma^2 (\text{measured}) - 2\,\sigma^2 (\text{measurement})]^{1/2}$

In this equation σ(real) is the real intersubject standard deviation, σ^2 (measurement) is the intrasubject variance of a probe-microphone measurement (100-800 Hz) or a threshold measurement (1-3 kHz). When the difference inside the square-root brackets is negative, then σ(real) is set equal to zero.

TABLE I. Correlation between free-field and earphone thresholds.

Frequency (kHz)	.5	1	1.5	2	3	4	6	8
	.99	.91	.89	.95	.65	.94	.53	.1

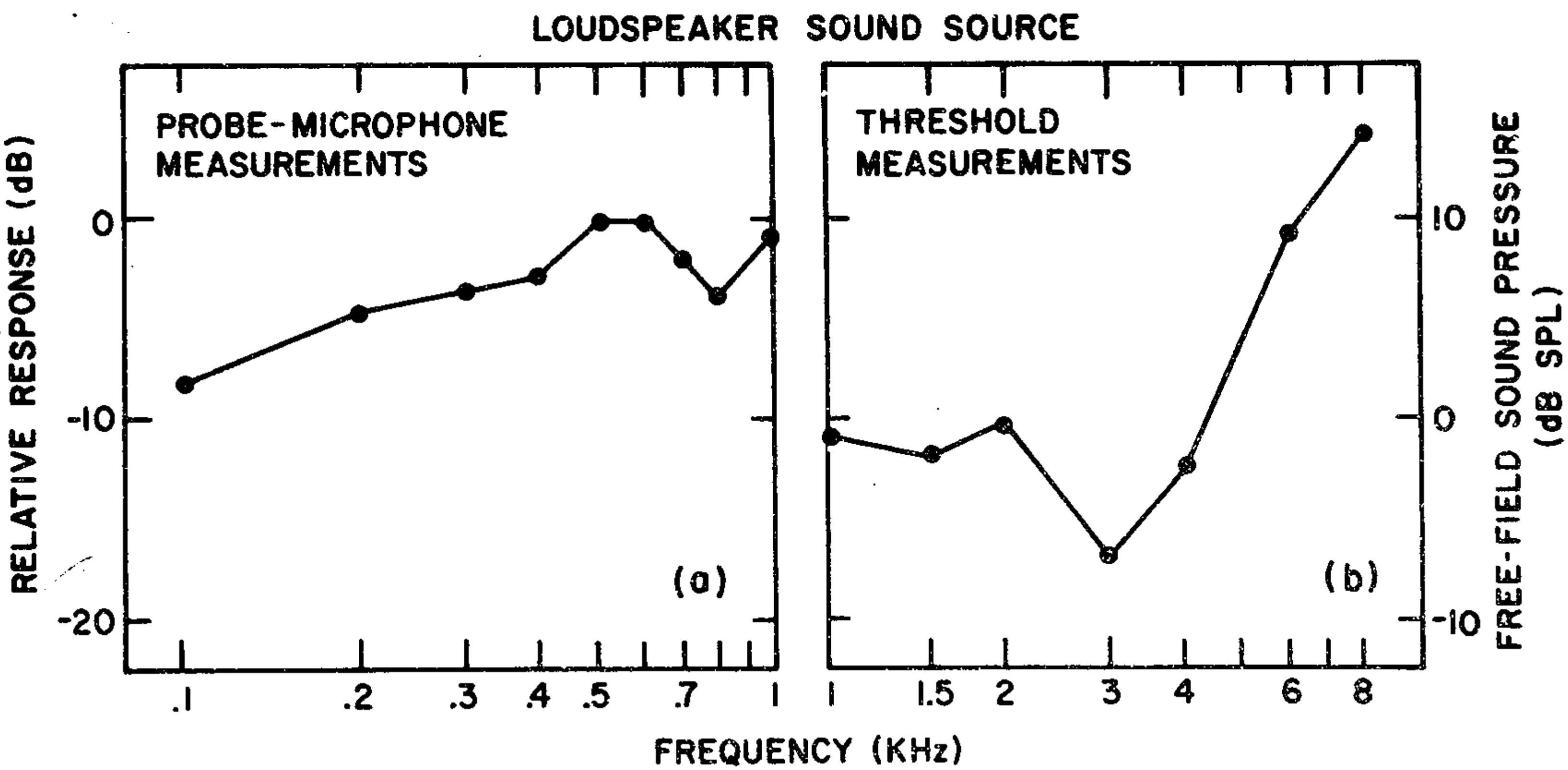
FIGURE CAPTIONS

Figure 1. Field measurements made using the loudspeaker as a sound source. Graph (a) presents probe-microphone response in dB for a fixed free-field sound pressure of 65 dB SPL at head-center position relative to the maximum response over the frequency range tested. Graph (b) presents free-field sound pressure level (dB SPL) at head-center position required for psychophysical threshold.

Figure 2. Measurements made using the earphone as a sound source. Graph (a) presents probe-microphone response in dB for a constant 100 mv on the 300Ω TDH-39 driver in the circumaural mounting relative to the same zero as in Figure 1a. Graph (b) presents the voltage on the TDH-39 driver (dB re 1 μv) required for psychophysical threshold.

Figure 3. Calibration of earphone relative to free-field transmission. Graph (a) presents the free-field sound pressure required to produce the same pressure at the eardrum as produced by 1 volt across the earphone driver. Graph (b) presents the intersubject standard deviation corresponding to the average results shown in Graph (a).

Figure 4 Equivalent free-field sound pressure produced by the circumaural earphones relative to sound pressure on an extended NBS-9A flat plate coupler for a fixed voltage on the earphone driver.



LOUDSPEAKER SOUND SOURCE
PROBE-MICROPHONE MEASUREMENTS
THRESHOLD MEASUREMENTS
RELATIVE RESPONSE (dB)
0
-10
-20
(a)
(b)
.1
.2
.3
.4
.5
.7
1
1.5
2
3
4
6
8
FREQUENCY (KHz)
10
0
-10
FREE-FIELD SOUND PRESSURE (dB SPL)

FIGURE 1

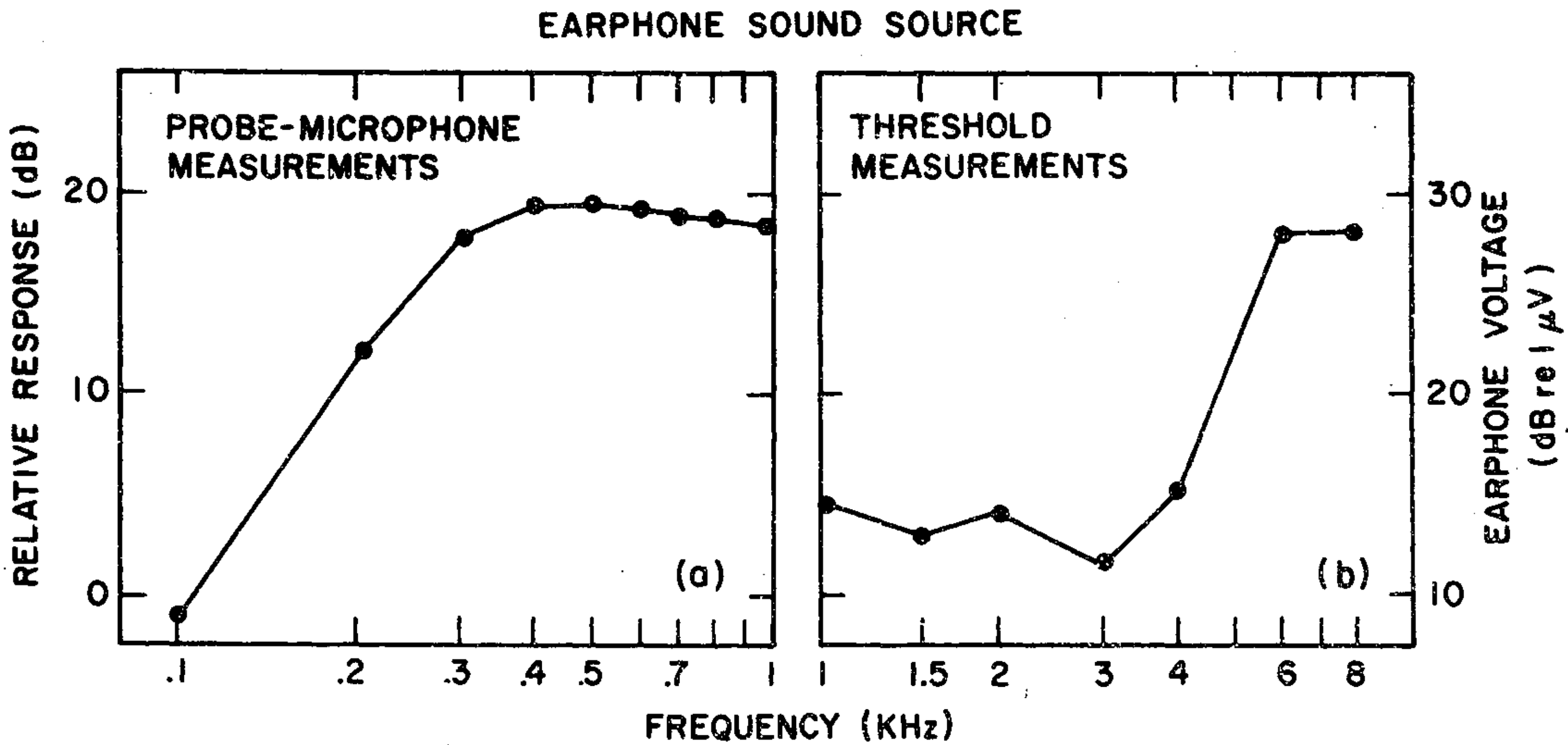

EARPHONE SOUND SOURCE
PROBE-MICROPHONE MEASUREMENTS
THRESHOLD MEASUREMENTS
RELATIVE RESPONSE (dB)
0
10
20
.1
.2
.3
.4
.5
.7
1
(a)
1
1.5
2
3
4
6
8
(b)
10
20
30
EARPHONE VOLTAGE
(dB re 1 μV)
FREQUENCY (KHz)

FIGURE 2

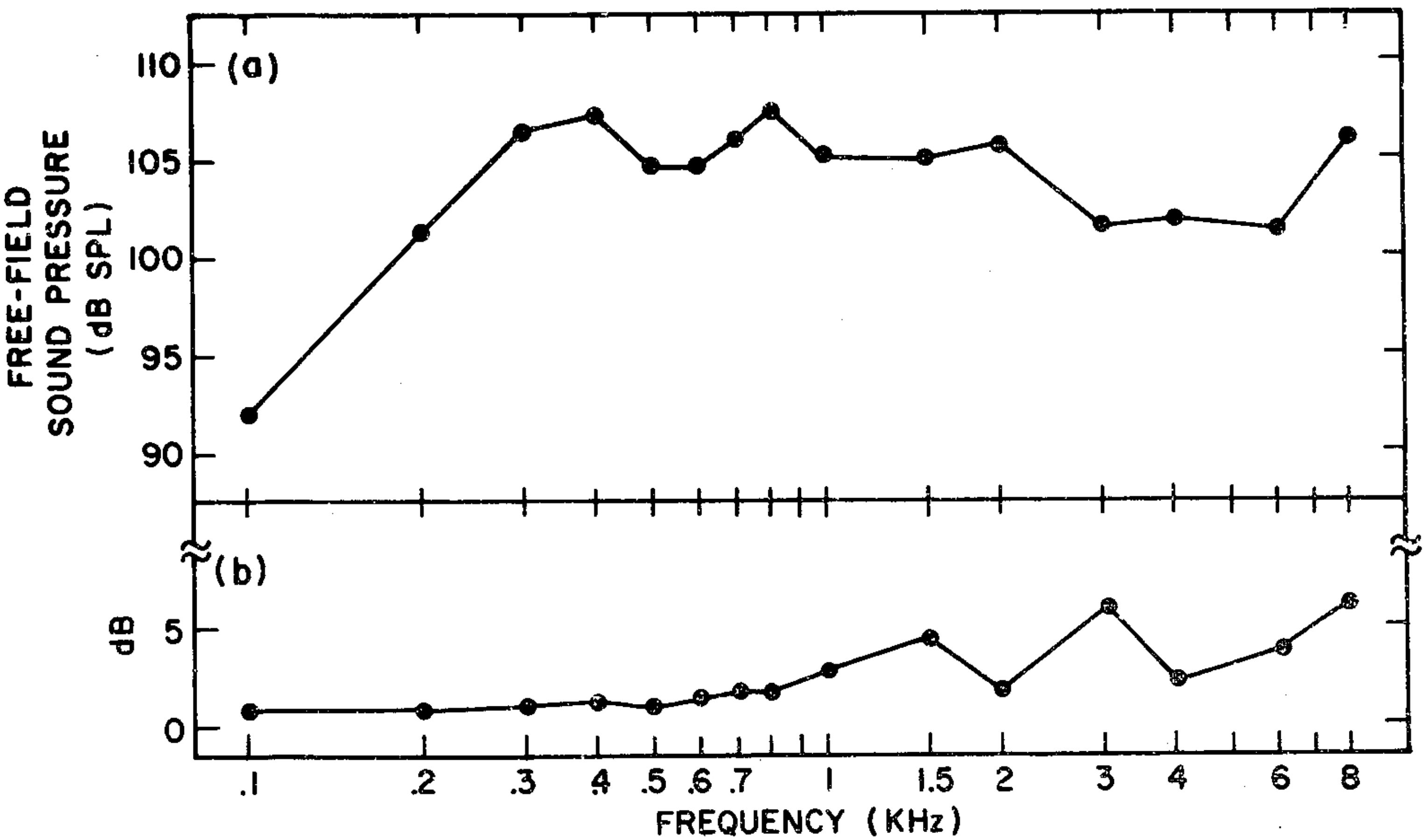

FREE-FIELD
SOUND PRESSURE
(dB SPL)
110
105
100
95
90
(a)
(b)
dB
5
0
.1
.2
.3
.4
.5
.6
.7
1
1.5
2
3
4
6
8
FREQUENCY (KHz)

FIGURE 3

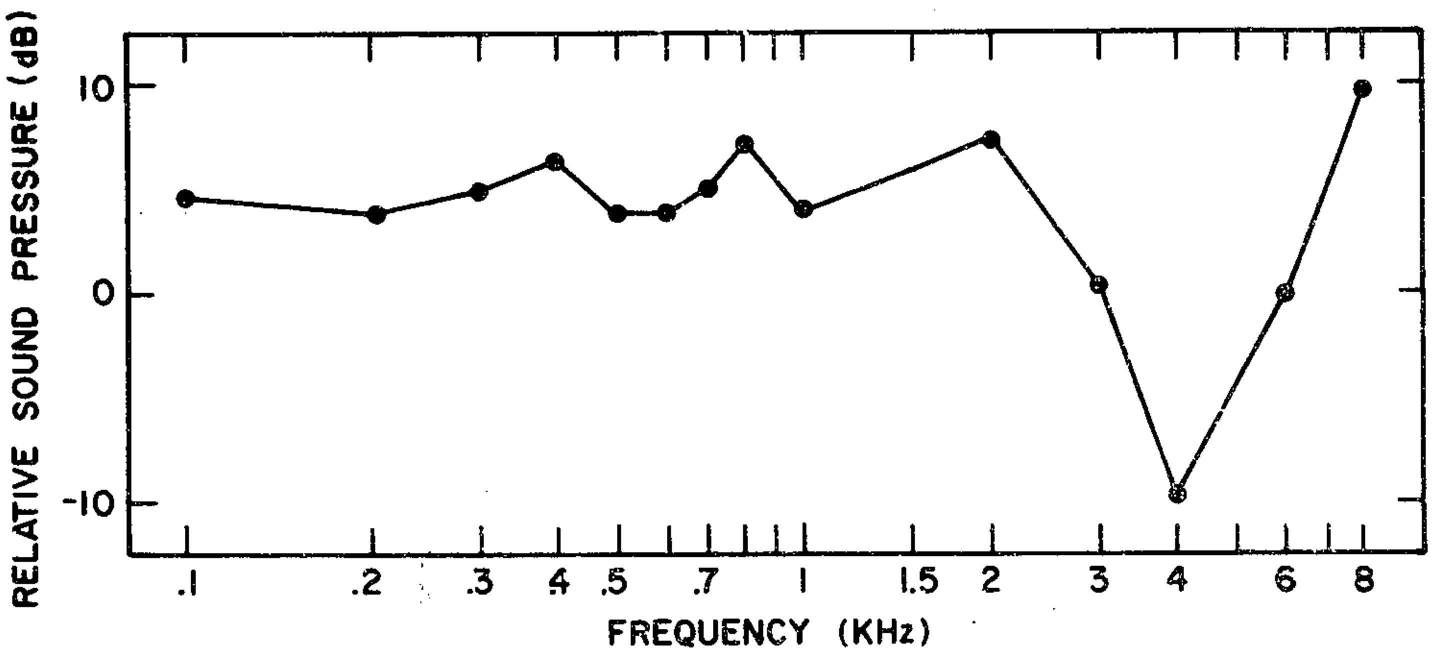

RELATIVE SOUND PRESSURE (dB)
10
0
-10
.1
.2
.3
.4
.5
.7
1
1.5
2
3
4
6
8
FREQUENCY (KHz)

FIGURE 4

REFERENCES

Charan, K.K., Cox, J.R. and Niemoeller, A.F. (1965). "Evaluation of New Couplers for Circumaural Earphones," J. Acoust. Soc. Am. 38, 945-955.

French, N.R. and Steinberg, T.C. (1947). "Factors Governing the Intelligibility of Speech Sounds," J. Acoust. Soc. Am. 19, 90-119.

Green, D.S. (1972). "Pure Tone Air Conduction Thresholds," in Handbook of Clinical Audiology, edited by J. Katz (Williams and Wilkins Co., Baltimore), Chapter 5.

Killion, M.C. (1976). "An Experimental Wideband Hearing Aid," J. Acoust. Soc. Am. 59, 562(A).

Michael, P.L. and Bienvenue, G.R. (1976). "Calibration Data for a Circumaural Headset Designed for Hearing Testing," J. Acoust. Soc. Am. 60, 944-950.

Morgan, D.E. and Dirks, D.D. (1974). "Loudness Discomfort Level under Earphone and in the Free Field: The effects of Calibration Methods," J. Acoust. Soc. Am. 56, 172-178.

Pascoe, D.P. (1975). "Frequency Response of Hearing Aids and Their Effects on the Speech Perception of Hearing Impaired Subjects," Annals of Otol. Rhinol. Laryngol., 84, Supp. 23.

Robinson, D.E. and Watson, C.S. (1972). "Psychophysical Methods in Modern Psychoacoustics," in Foundations of Modern Auditory Theory, edited by J.V. Tobias (Academic Press, New York), Vol. 2.

Robinson, D.W. and Dadson, R.S. (1956). "A Re-determination of the Equal-loudness Relations for Pure Tones," British J. of Applied Physics 7, 166-181.

Shaw, E.A.G. and Thiessen, G.J. (1962). "Acoustics of Circum-aural Earphones," J. Acoust. Soc. Am. 34, 1233-1246.

Shaw, E.A.G. (1966). "Earcanal Pressure Generated by Circum-aural and Supraaural Earphones," J. Acoust. Soc. Am. 39, 471-479.

Shaw, E.A.G. (1974a). "Transformation of Sound Pressure Level from the Free Field to the Eardrum in the Horizontal Plane," J. Acoust. Soc. Am. 56, 1848-1861.

Shaw, E.A.G. (1974b). "The External Ear," in Handbook of Sensory Physiology, edited by W.D. Keidel and W.D. Neff (Sprinter-Verlag, Berlin), Vol. V/1, Chapter 14.

Sivian, L.J. and White, S.D. (1933). "On Minimum Audible Sound Fields," J. Acoust. Soc. Am. 4, 288-321.

Villchur, E. (1969). "Free-field Calibration of Earphones," J. Acoust. Soc. Am. 46, 1527-1534.

Villchur, E. (1970). "Audiometer-Earphone Mounting to Improve Intersubject and Cushion-Fit Reliability," J. Acoust. Soc. Am. 48, 1387-1396.

Villchur, E. and Killion, M.C. (1975). "Probe-tube Microphone Assembly," J. Acoust. Soc. Am. 57, 238-240.