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COMPRESSION IN HEARING AIDS:

AN ANALYSIS, A REVIEW AND SOME RECOMMENDATIONS

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CONTENTS

The topic of amplitude compression in hearing aids is bedevilled by terminological confusion. A plethora of terms is used to describe compression amplifiers and the same terms are sometimes used by different authors in different ways. A recent survey by Schweitzer (1979) indicated that no less than 17 different terms were used by hearing aid manufacturers when referring to the compression characteristics of their hearing aids. The following discussion describes and defines the terms we shall use in relation to compression amplification as it applies in hearing aids.

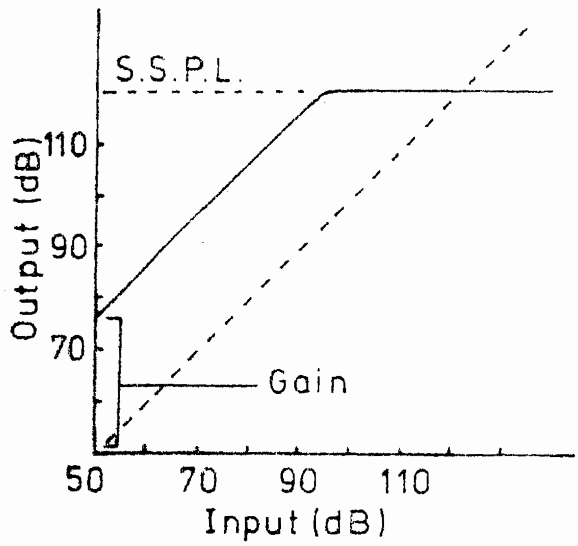
I. THE NATURE OF COMPRESSION AMPLIFIERS

What are the essential characteristics of an amplifier which allows it to be classified as a compression amplifier? The defining characteristics may be divided into two types: static characteristics and dynamic characteristics.

A. Static Characteristics

Amplifiers may be classified as linear or non-linear on the basis of their input/output transfer function. This is shown by a graph which plots the level of the output of the amplifier as a function of the level of the input to the amplifier. The input/output (I/O) curve of a linear amplifier is shown in figure 1.

Fig. 1: Transfer function for a linear amplifier, with a gain of 27 dB.



A number of features of this graph should be noted. Firstly, the curve is displaced from zero along the vertical axis by an amount equal to the gain of the amplifier. Secondly, the curve is linear and has a slope of unity until the output approaches the saturated sound pressure level (S.S.P.L.). For inputs above the input level corresponding to an output at the S.S.P.L., there is no further increase in output at that frequency. Unity slope simply means that any increase in input level results in an equal decibel increase in output. Manipulation of a volume control will cause the curve in the linear region to move up or down the vertical axis, but at any particular volume control setting the amount of gain applied to the signal does not depend on the input level (providing the S.S.P.L. point is not exceeded).

A compression amplifier is one which has an I/O characteristic with a slope of less than unity over any portion of its useful operating range.

Several such amplifier I/O characteristics are shown in figure 2. An amplifier with an I/O characteristic with a slope of greater than unity is known as an expansion amplifier.

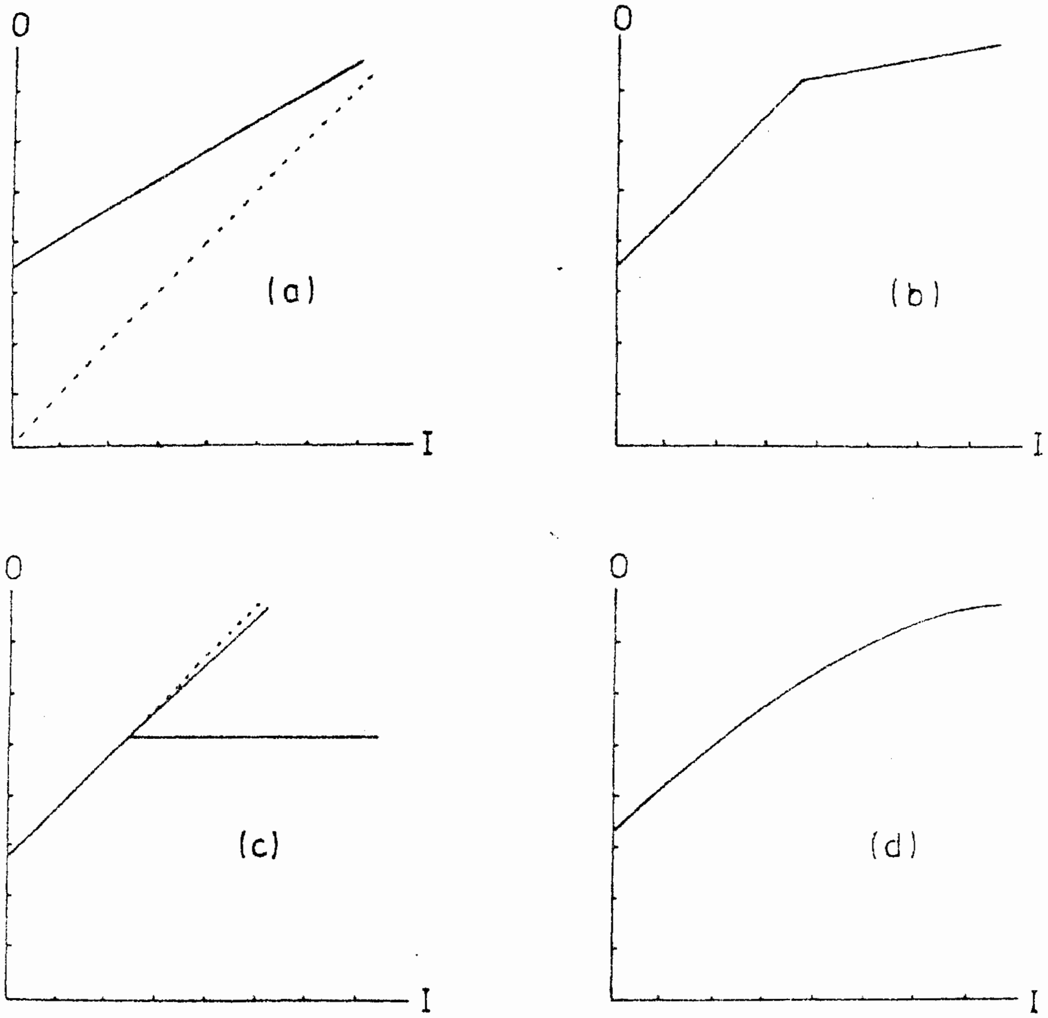
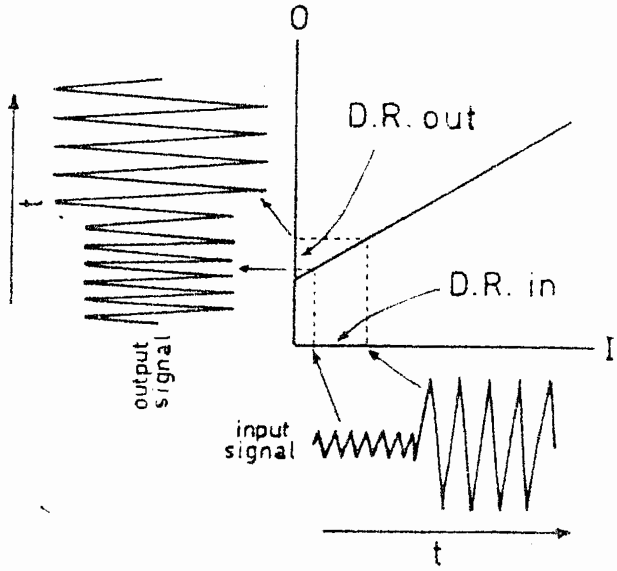


Fig. 2: Transfer functions for a variety of compression amplifiers.

In figure 2(a) the slope is less than unity over the entire operating range of the amplifier whereas in figure 2(b) the slope of the curve only becomes less than unity when a certain threshold input level is exceeded. In the latter case the amplifier would act as a linear device for inputs below the threshold level and as a compression amplifier for inputs above this level. Note too, that the slope of the curve can vary from just under unity to zero (as shown in figure 2(c)). In both these cases the compression part of the curve is a straight line when plotted in dB units. It is possible however, to have virtually any shaped curve, for example, as shown in figure 2(d). It should be noted that all compression and expansion amplifiers are strictly classified as non-linear devices even though some of them have I/O curves which are straight lines when plotted on a logarithmic scale (i.e. in dB units).

The important characteristic of a compression amplifier is that the amount of gain applied to the signal varies automatically as a function of the input level. As the level of the input increases, the amount of gain applied to the signal decreases and vice versa. Thus, the dynamic range of an input varying in level is reduced (compressed) at the output, as shown schematically in figure 3.

Fig. 3: Reduction of dynamic range as a result of compression amplification.



Several terms are used to describe the I/O curve of a compression amplifier. These are illustrated in figure 4.

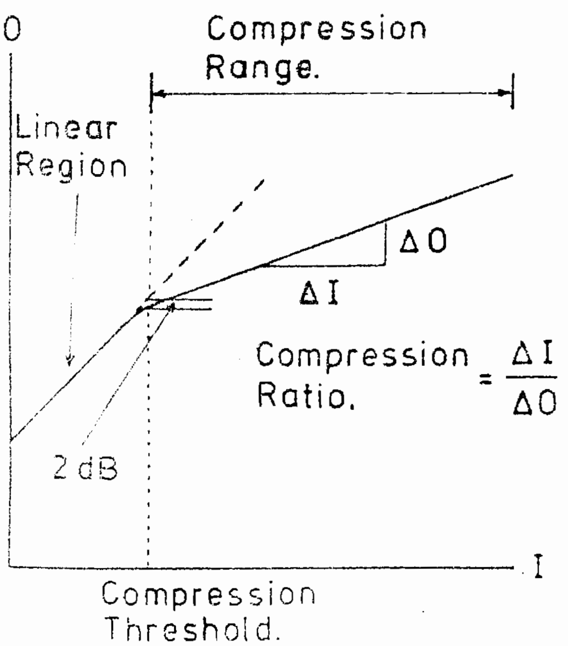


Fig. 4: Definition of the characteristics of the transfer function of a compression amplifier.

1. <u>The compression threshold</u> is defined as the minimum input level required to actuate the compressor function of the amplifier. Since there is not usually a precise "knee point" in the curve a further definition is necessary. The I.E.C. defines the compression threshold as the input level which gives a reduction of gain of 2 dB ($\pm$0.5 dB) with respect to the gain in the linear mode. (I.E.C. standard 118-2,1979).

2. <u>The compression range</u> is defined as the range of input levels over which the compression function operates.

3. <u>The compression ratio</u> is the inverse of the slope of the I/O curve. That is, any small change in input level divided by the resulting change in output level. The compression ratio is always greater than 1 for compression amplifiers. A compression ratio of 1 indicate unity slope, i.e., a linear amplifier, whilst a compression ratio of less than 1 indicates an expansion amplifier.

$$C.R. = \frac{\Delta I}{\Delta O}, \quad \text{where } \Delta I = \text{the change in input level, and}$$

$$\Delta O = \text{the resulting change in output level.}$$

For a compression amplifier whose I/O curve is a straight line, a single compression ratio will describe the slope. However, for an amplifier whose I/O function is curved, no single compression ration can adequately describe the function.

Some authors use terms such as "logarithmic compression", "semi-logarithmic compression" and "curvilinear compression" to refer to the specific shape of the I/O curve.

The I/O curves depicted in this section are usually measured with a pure tone input signal at some particular frequency.

For a given hearing aid, different I/O curves will be measured at different frequencies if the aid has anything other than a flat frequency response. The way in which the curves change with frequency depends on the location of the frequency dependent devices in the circuit. Some examples will be given later in this report, and it will be shown that only the position of the I/O curve with respect to the I/O axis is affected by a change of measurement frequency; the curve shape remaining unchanged. Although the I/O curve can also be measured with a broad band signal, the position of the curve will depend on the spectral composition of the signal.

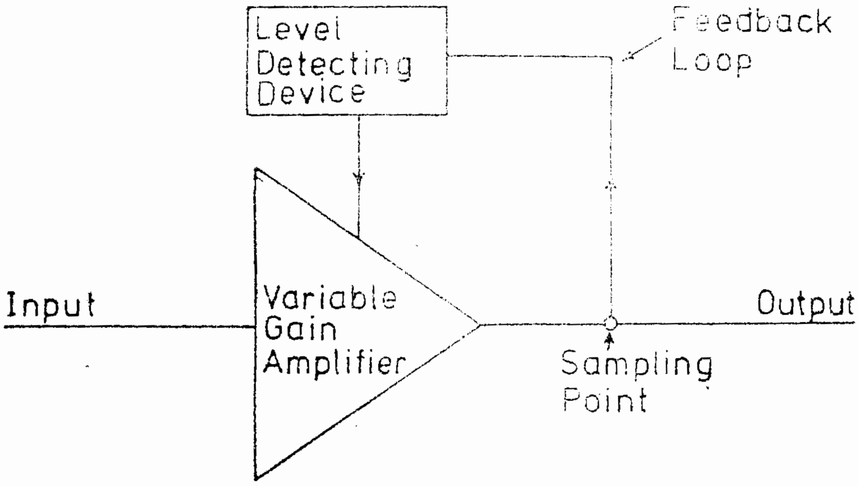
B. <u>Dynamic Characteristics</u>

The term dynamic characteristics refers to the way in which the amplifier responds to variations in signal level over time. Non-linear amplification can be achieved technically in a number of different ways. These may be divided into two classes; systems which react instantaneously to changes in input level and those which take a specified period of time to establish a stable output level following a change in input level. In this paper the term "compression" will be restricted to those systems falling into the latter category.

Despite their common usage, instantaneous limiters (such as peak clipping) have severe limitations for use in hearing aids because of the large amount of distortion they generate when performing their limiting function. (Peak clipping systems have probably continued to be used because of the difficulties associated with the design and manufacture of wearable compression systems.)

Amplification is normally achieved in several stages. The input is amplified in the first stage and the output of this stage is then used as the input for the second stage, and so on, until the required degree of amplification is achieved. Compression amplifiers operate by means of a feedback circuit. That is, the signal level in the circuit is sensed at a point somewhere in the amplification process by a level detecting device, and the output of this device is then fed back around one or more amplifier stages to control the gain provided by these stages for subsequent inputs. The controlling circuit is known as a feedback loop. Such a system is shown schematically in figure 5. (It is also possible for the control signal to be fed forward to following amplifier stages.)

Fig. 5: Block diagram of the feedback loop of a compression amplifier.

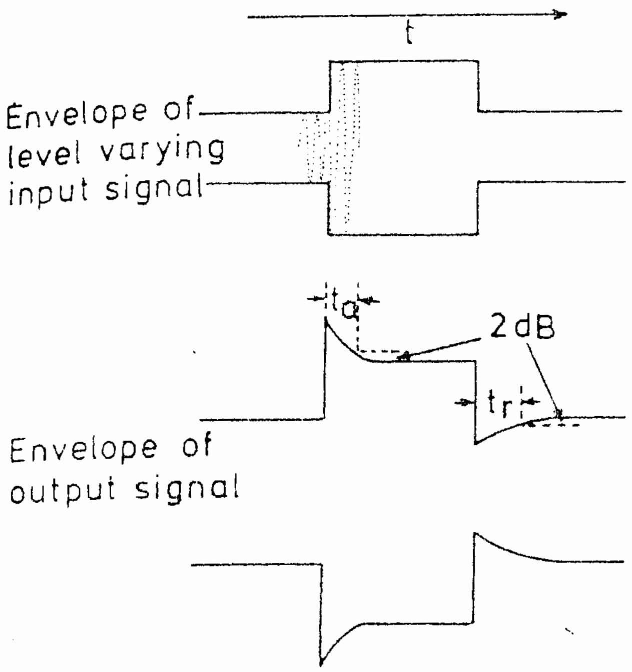


When the input level of these variable gain amplifiers is increased, there immediately follows a brief period of time during which the gain is progressively decreasing, until it finally stabilizes at the value which is appropriate for the new input level. Similarly, when the input level decreases, the system takes some time to adjust the gain upwards to its new stable level. The characteristic times which the amplifier takes to adjust to level changes are referred to as the attack and recovery times. Those times are determined by the time constants of the components in the feedback loop.

1. <u>Attack time</u> (t_a) is defined as the time interval between the moment when ~~the the input~~ signal is increased abruptly by a stated amount (usually 25 or 40 dB) and the moment when the output sound pressure level from the aid stabilizes at the new steady state level within ±2 dB.

2. <u>Recovery time</u> (t_r) is defined as the time interval between the moment when the input signal is decreased abruptly by a stated amount and the moment when the output sound pressure level from the aid stabilizes at the new steady state level within ±2 dB.

I.E.C. standard 118-2 (1979) contains a comprehensive description of the methods to be used to measure t_a and t_r, as well as other characteristics of compression amplifiers. Attack and recovery functions are depicted schematically in figure 6.



Fig, 6: Definition of attack and recovery of a compression amplifier,

The attack and recovery times employed have a marked influence on the way in which a compression amplifier operates. For the aforementioned static characteristics of the amplifier to apply, t_a and t_r must be considerably shorter than the time intervals between the level changes which are to be controlled. If this is not the case, and long time constants are employed, the compressor will adjust the gain as a function of the long term average level of the input. The length of time over which the "long term average" is computed by the aid is, of course, dependent on t_a and t_r.

C. Summary

It is concluded that a compression amplifier is one which:

1. Has an I/O characteristic which has a less than unity slope over all or part of its operating range, and;

2. Utilizes a form of feedback to achieve its effect and therefore;

3. Introduces time delays into the process of adjusting gain in response to changing input levels.

II. CLASSIFICATION OF COMPRESSION SYSTEMS

Compression amplifiers can be divided into those that have time constants sufficiently long that the gain does not change significantly during a single syllable or word, and those that have time constants sufficiently short that the gain does change significantly during a syllable or word. Compression amplifiers can also be divided into those having high compression thresholds and those having low compression thresholds. By a high compression threshold we mean one that is exceeded only occasionally by the signal of interest or by other background noise. Conversely, a low threshold is one which results in the input signal activating the compression function for a large proportion of the time. To complete the classification scheme, we also need to specify the size of the compression ratio for signals above the compression threshold. Although a continuum of values could be used here also, a useful division is to categorise the CR as large if it is greater than 5 or small if it is less than 5. The above classifications provide eight possible combinations, as shown in Table 1.

	LOW COMPRESSION THRESHOLD		HIGH COMPRESSION THRESHOLD	
	LOW C.R.	HIGH C.R.	LOW C.R.	HIGH C.R.
Short Time Constants	(1) Syllabic Compression	(2) (Syllabic Compression)	(3) —	(4) Compression Limiting
Long Time Constants	(5) (Automatic Volume Control)	(6) Automatic Volume Control	(7) —	(8) —

Table 1.

Classification Scheme for Compression Hearing Aids.

A. Syllabic Compression

This type of compression system is characterised by short time constants and a low compression threshold so that virtually all signals of interest to the listener will be undergoing some compression. Such systems are referred to as syllabic compressors because they reduce the level differences between successive speech syllables (Braida et al, 1979). The term "dynamic range compression" (DRC) is sometimes applied to such systems but this can be misleading as all compression systems reduce the dynamic range of the input signal. Syllabic compression systems normally have a low compression ratio (cagegory 1), so that some level differences between syllables are retained. However systems falling into category 2 will also be referred to as syllabic compression systems. Such systems might be appropriate for a listener with an extremely small dynamic range (between threshold and discomfort level), as all syllables will be presented at the same level. (For such a listener however, a suitably designed category 2 system will be indistinguishable from a suitably designed system falling into category 4.)

B. Compression Limiting

Category 4 systems with short time constants, a high compression threshold and a high compression ratio are usually referred to as compression limiting systems. These systems limit the maximum output of the aid while operating as a linear amplifier for commonly occurring inputs which fall below the compression threshold. The term "high level compression" is also applied to this type of system.

C. Automatic Volume Control (A.V.C.)

Category 6 systems with long time constants, a low compression threshold and a high compression ratio will adjust the gain of the amplifier as a function of the long term average level of the input. A.V.C. systems thus maintain a relatively constant average output level in the presence of an input fluctuating slowly in level. Short term level fluctuations in the input (e.g. from one word to the next) remain essentially unchanged. These systems are also referred to as automatic gain control (A.G.C.) systems.

The exact break point between long and short time constants must necessarily be somewhat arbitrary but the Swedish authorities regard compression systems with release times in excess of 150 msecs as A.V.C. systems (Johannson & Lindblad). Similarly, a clearcut distinction cannot be made between high and low compression thresholds, but an A.V.C. system would have its compression threshold set so that the compression function was operative over the major portion of the useful input range.

As well as category 6 systems, which adjust the gain control so that the average output level stays at or close to a predetermined value, systems falling into category 5 will also be referred to as A.V.C. or A.G.C. systems. For these systems however, the average output level will vary with the average input level, but by a reduced amount.

D. Miscellaneous

Systems falling into categories 3, 7 and 8 have so far not been mentioned. These systems appear to be of no practical value for the following reasons. Category 3 systems provide no compression to the speech signal of interest and so can only function to reduce the output for especially intense inputs. This they do, but not as well as do category 4 systems. Systems falling into categories 7 or 8 provide no protection from intense sounds if the attack time as well as the recovery time is made long. If the attack time is made short, a large gain reduction will result when intense sounds are encountered. Unfortunately, the reduction will persist for a long time after the intense sound ceases, and so cause any following syllables to be perceived at a very low level, if at all.

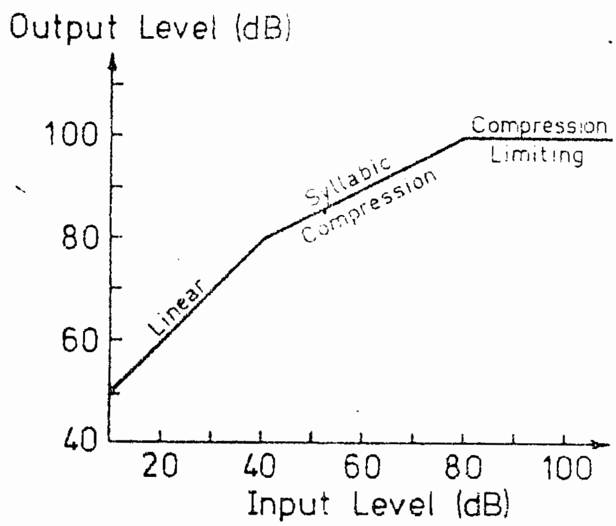
E. Time Constants

As the choice of the two time constants can be made independently, there are really four possible combinations of the time constants (short-short, short-long, long-short, and long-long). In practice however, not all of these combinations are useful so the classification scheme has not been complicated in this way. "Short time constant" systems (1, 2 & 4) require that both attack and release times be short for proper functioning. "Long time constant" systems (5 & 6) require that the release time be long and that the attack time be of at least moderate length.

F. Limitations of Classification System

The principle limitation of the system is that each of the three defining parameters can have a value drawn from any point in a continuum. The boundaries between the categories are thus somewhat arbitrary. A lesser limitation is that a particular piece of hardware may function as one type of system for one input signal, but as a different type for another. (For example, a change from category 2 to category 4 when going from a very quiet environment to a very noisy one.) That is, the classification is signal dependent as well as device dependent. Clearly, an aid can also move between categories 2 and 4 if its gain is changed by a large amount by altering the volume control. Also, a particular device may function simulatenously as more than one type of compression system. It is quite conceivable that an aid could usefully contain an A.V.C. system (6), followed by a fast acting system which worked as a syllabic compressor for low level signals (1) and as a compression limiter for high level signals (4). The I/O curve for the fast acting section of such a hearing aid is shown in figure 8.

Fig. 8: Input/Output curve of a compression system performing different functions for different input levels.



A further complication for any classification scheme is the fact that any of the three defining parameters can in principle have values which vary as a function of frequency.

Some classification scheme is required so that results obtained with different compression systems can be sensibly discussed. Despite the above limitations, we feel that the present scheme is useful for this purpose.

III. EFFECT OF COMPRESSION ON AID RESPONSE.

A. Effect on Output Level: Input and Output Control

The operation of a hearing aid compression amplifier is influenced by the position of the manual volume control within the electronic circuit. The volume control may be positioned after the sampling point of the feedback loop or it may be positioned before it. The former arrangement is known as input controlled compression because the compressor function is controlled only by the level of the signal entering the circuit, and is unaffected by the manual volume control setting. The latter arrangement is known as output controlled compression. The two arrangements are shown schematically in figure 9 together with I/O curves showing the effects of altering the setting of the manual gain control in each case.

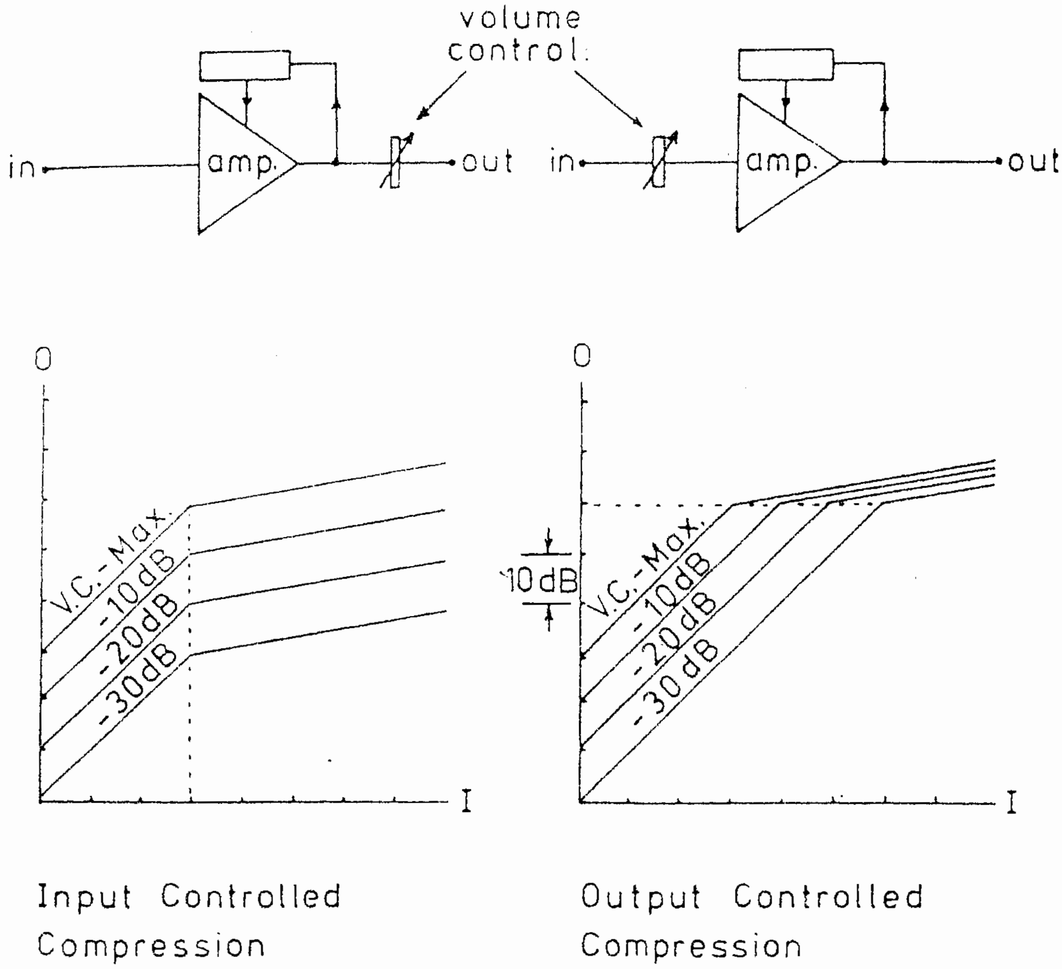


Fig. 9: Circuit arrangement of input and output controlled compression amplifiers and effects of varying the manual volume control in each case.

It will be noted that manually reducing the gain of the input controlled compression amplifier maintains a constant compression threshold but alters the <u>output</u> level at which the compression function becomes operative. Thus, changing the volume control setting also effectively alters the maximum output of the amplifier. On the other hand, manually reducing the gain of the output controlled compression amplifier raises the compression threshold but the output level at which compression starts remains constant. Thus, the maximum output level is independent of the volume control setting.

B. <u>Effects on Signal Spectrum</u>

The effect of a compression hearing aid on the spectrum of a signal depends both on the position of any frequency dependent devices within the aid and on the way the response is measured. There are only three functionally different positions which the frequency dependent circuits can occupy: before the point from which the signal is fed back to the compression amplifier, after it, or in the feedback loop itself.

1. Prefiltering

Initially, we will assume that all frequency dependent devices in the aid occur prior to the feedback point. The system block diagram is shown in figure 10(a). An assumed filter characteristic is shown in figure 10(b), and the I/O curve for the amplifier section in figure 10(c). As an example, we have chosen an I/O characteristic having a linear region, a 2:1 compression region, and a region of compression limiting.

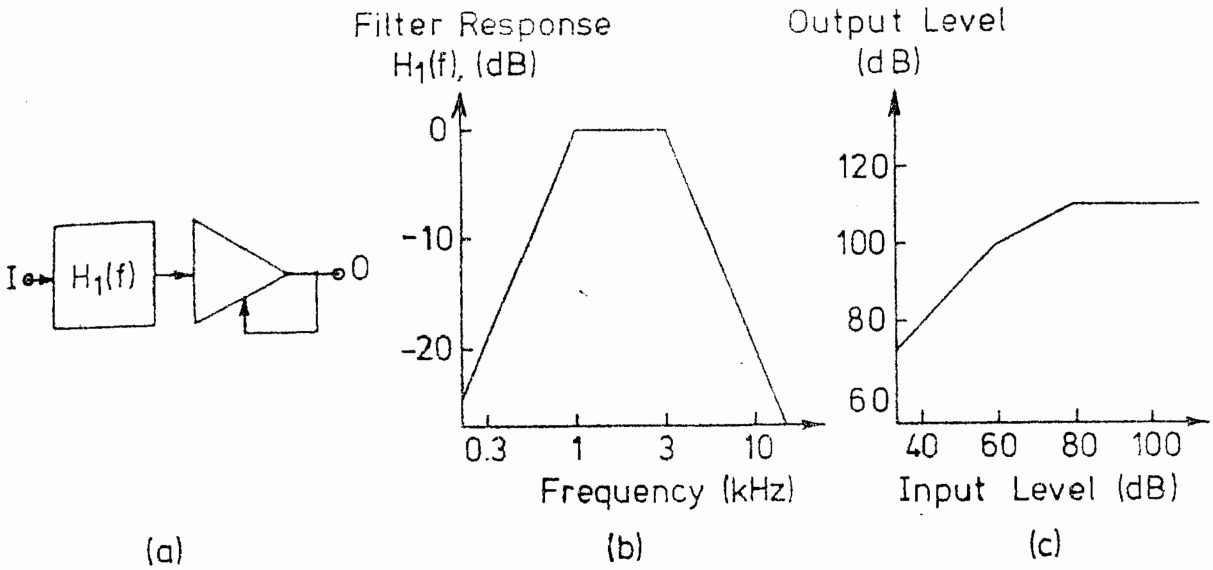


Fig. 10: Prefiltering compression system: (a) block diagram; (b) Response of linear filter; (c) Input/Output curve of compression amplifier.

Now let us test this device in the conventional manner by applying a swept-frequency pure tone of constant amplitude to the input. After each sweep, the input amplitude is increased by 10 dB and the process repeated. The resulting family of output curves is shown in figure 11(a).

The diagram shows that the low and high frequency slopes decrease as the signal moves into the 2:1 compression region and that the response becomes completely flat (over the frequency range shown) for the highest level inputs.

Instead of applying a swept pure tone as the analysis signal, let us now apply a white noise signal and measure the output with a spectrum analyzer. For any particular input level, the aid will have one, and only one, gain. The relationships between the various spectral components in the input will therefore be retained in the output. The resulting family of curves will thus be as shown in figure 11(b). Which of these two measurement methods is the most relevant for speech input to a hearing aid? Neither graph by itself fully describes the aid's operation. The first is applicable when different spectral components are applied sequentially; the second when they are applied simultaneously. Both of these are of relevance to hearing aid use because the signal of interest is nearly always a broad band signal (simultaneous components) but one which has a spectrum which can be quite different from one syllable, or even phoneme, to the next (sequential inputs). Although the response curves shown in

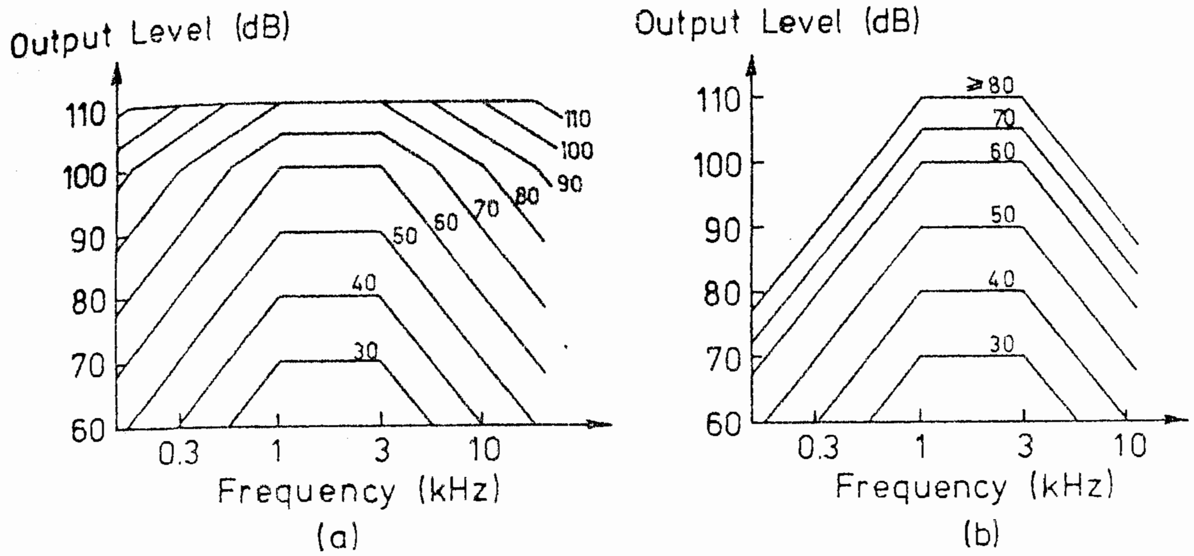


Fig. 11: Frequency response of system defined in figure 10: (a) Response for swept pure tone input; (b) Response for white noise input (parameters show input level).

this section have been derived theoretically by considering the separate effects of the filter and the compressor or the signal they have been confirmed experimentally by us using a Franz PDM Compressor type EMT 156, a Krone-Kite filter type 3202 to simulate H(f), a B & K oscillator and level recorder to make the swept pure tone measurements, and a Grason Staedler white noise generator type 1285 and Spectral Dynamics type SD 360 spectrum analyzer to make the broadband signal measurements.

Apart from a conventional frequency response characteristic, there are two other useful graphical ways of showing the response to different input frequencies of compression hearing aids. The first of these, the I/O curve, has been described earlier in this report. Figure 12 shows the I/O curves for several frequencies (measured sequentially) for the system described in figure 1 (a). Different input frequencies simply cause a lateral shift in the I/O curve in exactly the same way as when a volume control precedes the compression amplifier. A third type of curve that is useful for understanding the aid's operation is the curve which shows how the compression threshold changes with frequency. The curve for our pre-filter hearing aid is shown in figure 12(b), and it can be seen that for this filter, mid-band signals will have a lower compression threshold than low or high frequency signals where the aid gain is less.

When a broadband signal is input to the device, the position of the I/O curve for each spectral component depends on the spectral shape of the signal, so no general curves can be drawn. For a similar reason a compression threshold curve cannot be drawn for a broadband input signal.

2. Post filtering

The effect on the signal spectrum is somewhat simpler when the frequency dependent devices are located after the compression feedback point. Figure 13 in the same way as before shows the block diagram and systems properties.

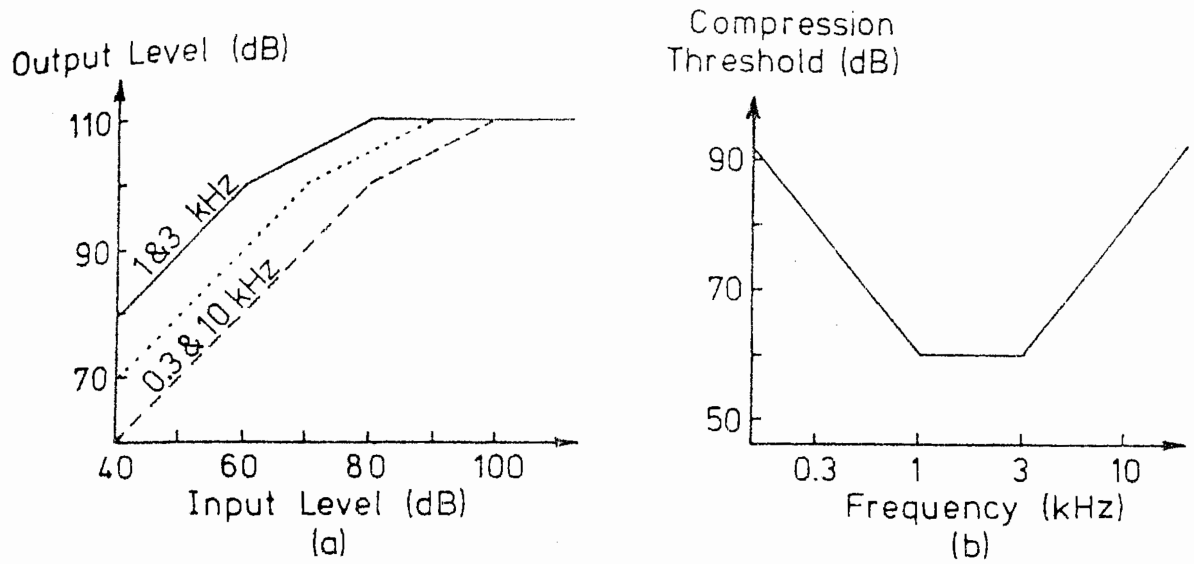


Fig. 12: Characteristics of system defined in figure 10: (a) Input/Output curves for different frequencies; (b) Compression threshold curve.

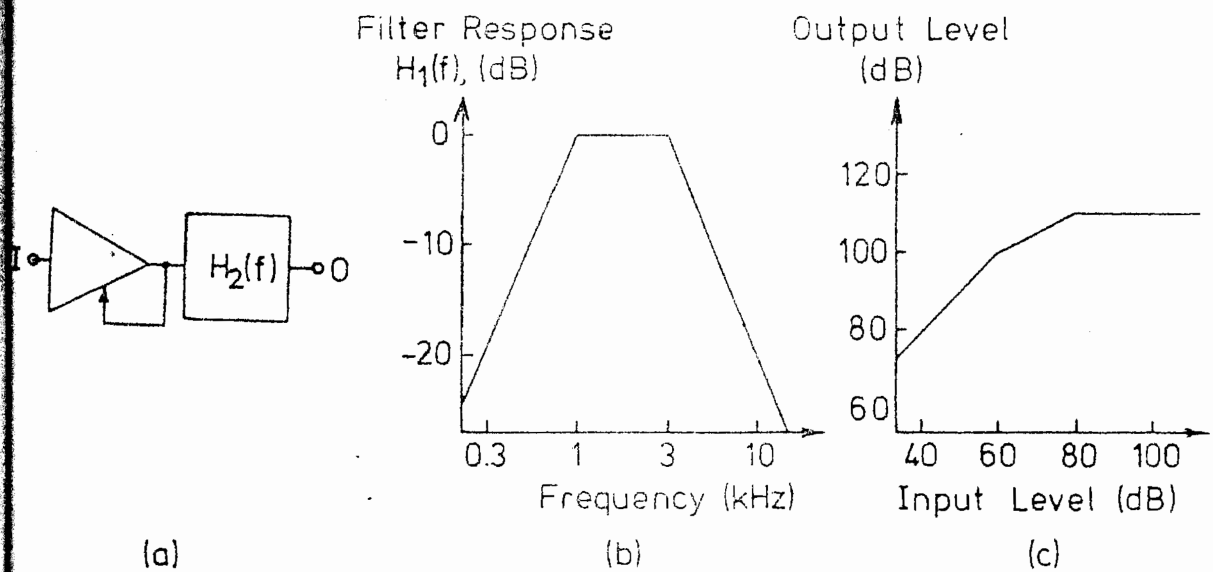


Fig. 13: Postfiltering compression system: (a) Block diagram; (b) Response of linear filter; (c) Input/Output curve of compression amplifier.

Now however, the family of output curves resulting from inputs spaced 10 dB apart in intensity is the same no matter whether sequential input signals (e.g. swept pure tones) or simultaneous components (e.g. white noise) are used. The curves that would be obtained are similar to those already shown in figure 11(b).

I/O curves and compression threshold curves, however, are quite different from the prefiltered case. Figure 14(a) shows the I/O curves and it can be seen that the curve shifts vertically for different frequencies. The compression threshold curve, shown in figure 14(b) indicates that all frequencies now have the same compression threshold; the threshold being independent of the aid's frequency response, $H_2(f)$.

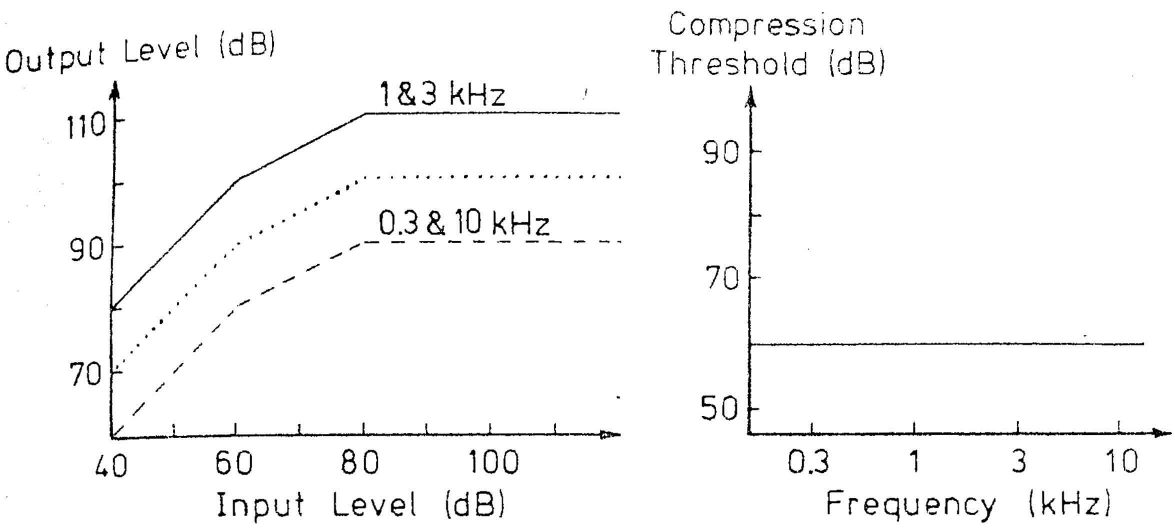


Fig. 14: Characteristics of system defined in figure 13: (a) Input/Output curves for different frequencies; (b) Compression threshold curve.

3. Feedback Loop Filtering

The third functionally different position which frequency dependent devices can occupy is within the feedback loop, but before the a.c. to d.c. converter. The system block diagram is shown in figure 15(a), and an example of the frequency dependence of G(f) in figure 15(b). A typical I/O characteristic for the amplifier (without the filter) is shown in figure 15(c), and is unchanged from before.

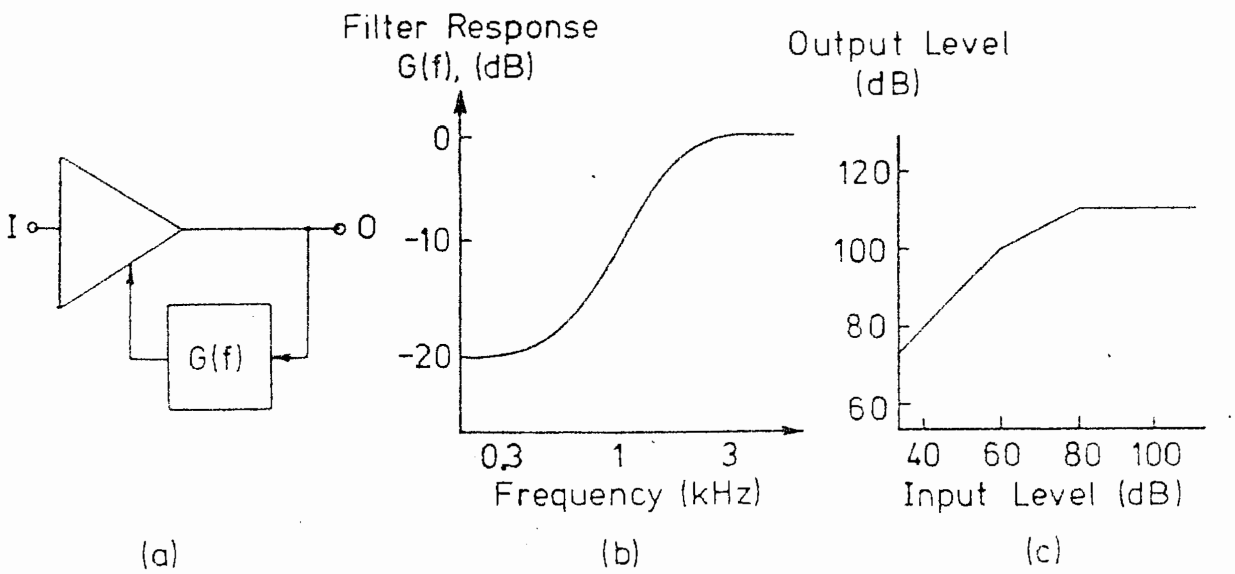


Fig. 15: Feedback loop filtering compression system: (a) Block diagram; (b) Response of linear filter; (c) Input/Output curve of compression amplifier.

For this system, it is clear that for low level inputs, where the amplifier is operating in its linear region, the gain of the system is independent of frequency. By contrast, higher level signals will most easily cause a gain reduction in the compression amplifier when they fall within the pass band of G(f). The resulting frequency response curves are shown in figure 16(a). I/O curves for inputs of different frequencies are shown in figure 16(b), and the compression threshold is shown as a function of frequency in figure 16(c). Both the maximum possible output (M.P.O.) curve (the top curve in figure 17(a) and the compression threshold curve are equal to the inverse of the G(f) curve. These graphs are all for sequential input signals (e.g. swept pure tones); it is clear that simultaneous input signals are always passed through the system of figure 15(a) with the relationships between the spectral components unchanged.

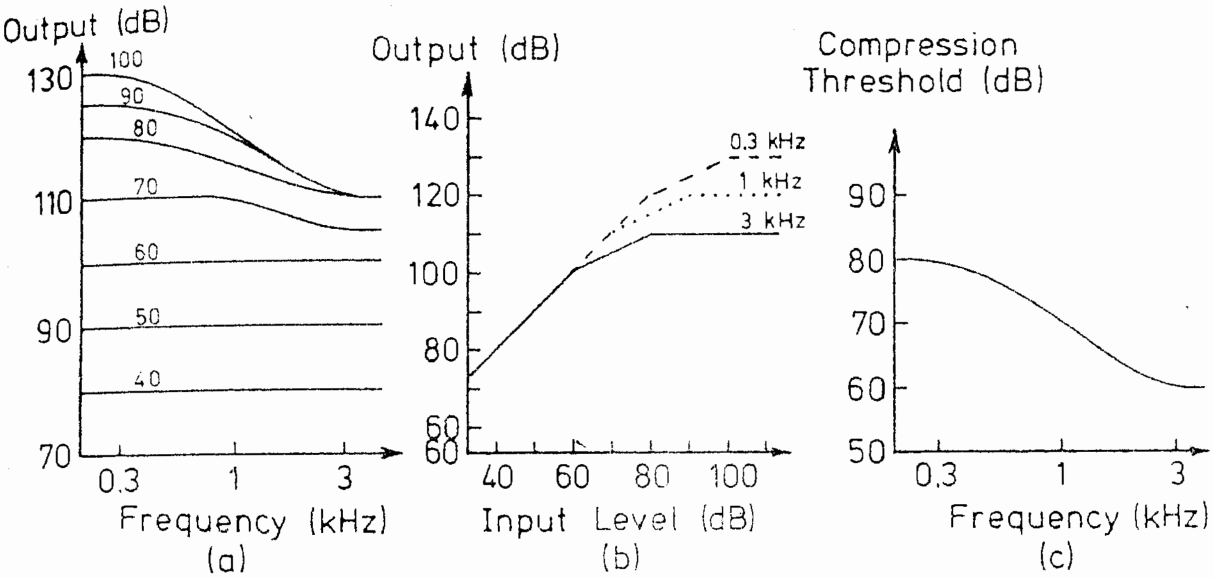


Fig. 16: Characteristics of feedback loop filtering system defined in figure 15: (a) Frequency response (parameters show input level); (b) Input/Output curves for different frequencies; (c) Compression threshold curve.

4. Combined Systems

Practical hearing aids will usually have some frequency dependent devices before the compression feedback point (microphone, fixed or variable tone controls), some after the feedback point (fixed or variable tone controls, earphone/tubing/earmould), and may or may not have some within the feedback loop. The combined system will thus appear as in figure 17. The variation of the aid's response (for sequential input signals) as the input level is changed will thus be complex, but predictable once the frequency dependence of the components within each of these three sections of the circuit is known.

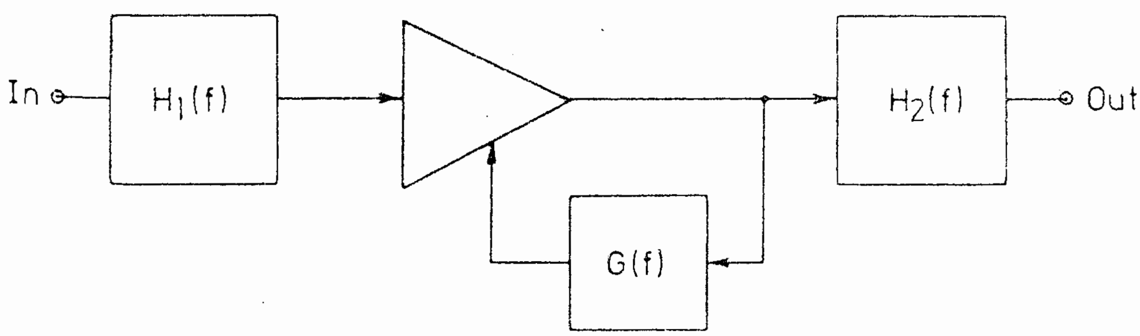


Fig. 17: Block diagram showing the three possible locations of frequency dependent devices.

The results of the previous 3 sections may be combined and summarised as follows. We are assuming that the compression amplifier performs compression limiting for high level input signals. Firstly, for sequential input signals:

(i) The frequency response for low level input signals (linear amplifier region) is equal to $H_1(f)\ H_2(f)$;

(ii) The frequency response for the highest level input signals will be equal to $H_2(f)/G(f)$, and independent of $H_1(f)$ provided that the input dynamic range of the compression amplifier in the limiting region is larger than the variation of $H_1(f)$ across the frequency range of interest. Otherwise, $H_1(f)$ will also influence the response;

(iii) The M.P.O. curve for the aid is also equal to $H_2(f)/G(f)$ under the same conditions as in the previous paragraph;

(iv) The compression threshold curve is equal to $1/(H_1(f)G(f))$;

(v) The I/O curve for any particular frequency, f, is shifted with respect to some reference frequency, fr, in the following way: a vertical shift upwards equal to $H_2(f)\ G(fr)/(H_2(fr)G(f))$, and a horizontal shift to the right equal to $H_1(fr)\ G(fr)/(H_1(f)\ G(f))$.

[Note: We are expressing the transfer functions $H_1(f)$, $H_2(f)$, and G(f) in linear terms. If they are expressed in logarithmic units, the multiplications and divisions reported here become additions and subtractions.]

Secondly, for simultaneous (broadband) input signals:

(i) The frequency response for input signals at any level within the dynamic range of the device is equal to $H_1(f)\ H_2(f)$.

(ii) The M.P.O. for any spectral component occurs when a high level signal containing only that component is input to the device. This situation thus becomes identical to the sequential input signal measurement method. In general, when other spectral components are added to a pure tone, the M.P.O. at the original pure tone frequency decreases, just as for a conventional linear amplifier in an overload condition.

(iii) A compression threshold curve can not be measured for a broadband signal.

(iv) The position of the I/O curves for each spectral component depends upon the overall spectral shape,

C. Effect on Tone Control Variations

As the above discussion has shown, a tone control will have different effects depending upon whether it is located before or after the feedback point. Tone control circuits located after the feedback point will have the same effect as they do for a linear amplifier both for sequential and simultaneous input signals. The effect of varying the tone control will thus be just the same as for linear amplifiers. Tone controls located before the feedback point will treat different sequential inputs (of varying intensity and spectral composition) differently. The effect of tone control variations for such signals will thus be different from that which is measured for a linear amplifier.

IV. RATIONALE FOR THE USE OF COMPRESSION IN HEARING AID AMPLIFIERS

A. Compression Limiting

There are very important reasons for limiting the maximum possible output of a hearing aid. These have been adequately dealt with elsewhere and will not be discussed here. It is sufficient to say that the purpose of output limiting is to prevent the output sound pressure level of the aid from becoming uncomfortable or dangerous for the user.

All amplifiers have an inherent output limitation. The maximum electrical current possible in the circuit is determined by the battery voltage and by the resistances of the earphone, the battery and the residual resistance of the transistors. This current limit and the battery voltage together impose an absolute maximum output power limit on the amplifier. In practice an additional limiting resistor is often wired in parallel with the earphone to limit the maximum output to below the absolute maximum. This additional resistor is often variable in order to provide the fitter with the facility to vary the maximum output of the aid. This form of output limiting is referred to as peak clipping. It should also be noted that the maximum output of a hearing aid may sometimes be governed by the mechanical limitations of the earphone, rather than by the electrical limitations of the amplifier.

Many hearing aids today still rely on peak clipping to limit the maximum output of the aid. There is ample evidence that the high levels of distortion generated by peak clipping severely degrade both the intelligibility and the perceived quality of the reproduced sound.

By contrast, fast acting compression, with a high compression threshold and ratio, limits the maximum output of the aid without causing massive increases in distortion (see Lybarger, 1974, p.68). However, particular designs and aids may not achieve the reduction of distortion which is theoretically possible.

B. Syllabic Compression

Sensori-neural hearing loss is characterised by a reduced dynamic range of hearing. That is, the threshold of audibility is considerably higher than normal, but the loudness discomfort level (L.D.L.) may be the same as for normal ears or raised by a much smaller amount than the threshold. The reduced D.R. is accompanied by "recruitment", that is an abnormally rapid growth of loudness with increasing sensation level. The audiologist is faced with the problem of fitting the amplified speech signal, or as much of it as possible, within this reduced range above threshold but below L.D.L. It frequently becomes impossible to raise all of the speech signal to audibility because the clients' dynamic range of hearing is less than the dynamic range of the signal. In extreme cases, the situation may arise where any signal amplified above threshold to any usable degree is intolerably loud for the client. Edgardh (1952) was among the first to suggest that speech intelligibility could be improved for patients with reduced dynamic ranges if compression were used to equalize the levels of vowels and consonants of speech thus permitting more of the signal to be audible without violating the L.D.L. Since that time compressors have been used to try to match the dynamic range of speech to that of the impaired ear. Several investigators have attempted to precisely compensate for recruitment by providing an I/O function which accurately mirrors the recruitment curve. The outcome of such studies will be discussed later.

C. Automatic Volume Control

The term A.V.C. is used to designate slow acting compressors which adjust the gain of the amplifier as a function of the long term average input level. In terms of the effect on short-term level variations in the signal they may be considered to be linear amplifiers.

A.V.C. compressors have long been used in tape recorders, and radio broadcasting and receiving equipment to maintain a relatively constant long term output level in the presence of a fluctuating input level. Some "auditory training" equipment for use with the deaf and hard of hearing also incorporate A.V.C. circuitry.

A major problem facing the user of a hearing aid is fluctuating overall signal levels. One of the major causes of these fluctuations is the changing distance between the signal source and the aid microphone. One advantage that radio transmitter aids have over conventional acoustic hearing aids is that they are capable of maintaining a fixed distance between the microphone and the signal source and hence the problems associated with a fluctuating input level are much reduced. The maintenance of a constant output level is particularly important for those persons with sensori-neural hearing loss who exhibit maximum speech intelligibility scores over a very narrow range of intensities and/or who exhibit a very narrow tolerance range for auditory inputs. The incorporation of A.V.C. facilities into hearing aids would certainly go some way towards alleviating the problems associated with overall level fluctuations. The alternative is for the client to continually operate the manual volume control. This latter approach is far from adequate or practicable in most everyday listening conditions.

D. Advantages of Compression which are Common to More Than One System

Compression amplifiers are superior to linear amplifiers in that they maintain the signal to noise ratios of high level signals in noise whereas linear amplifiers driven in saturation do not. Take as an example a signal imposed on noise so that both signal and noise occur simulatenously. When the signal causes a linear amplifier to peak clip there will be intermodulation distortion products formed between the signal and itself, the signal and the noise and the noise and itself. The first set of products are perceived as distortion and the other two as noise. As the input level is raised past the point at which the aid begins to distort heavily, the distortion at the output will increase in level but the signal will not. Of course, this does not mean that lower level signals and noise will continue to be amplified linearly when simultaneously present high level signals have begun to be clipped. In fact, once high level signals reach saturation the low level signals will probably be obscured by the resulting distortion products. On the other hand, a compression amplifier will reduce the gain provided for both signal and noise, without adding distortion products, so the signal to noise ratio is maintained. The above advantage would apply for both syllabic and limiting compressors, and for A.V.C. compressors if the signal is of sufficient duration.

Against any superiority of compression amplifiers in maintaining signal to noise ratios for simultaneous presentation of signal and noise must be weighed the disadvantage of compression associated with sequential presentation of signal and noise. During periods of signal absence, the lower level noise will receive increased gain and the difference between the signal level and the noise level will thus be reduced. This is most likely the cause of clients' complaints about the "Noisiness" of compression hearing aids. A partial solution to this problem is the incorporation into the aid of expansion for low level signals.

A second advantage which is common to both syllabic and limiting compression is that because of the protection afforded against the passage of very intense sounds, the aid user may choose to use a somewhat higher gain setting than he would with linear peak clipping amplification. The extent of this effect would be expected to increase as the compression ratio becomes higher. This could be a very important advantage for hearing impaired persons with high thresholds of audibility and a greatly reduced dynamic range of hearing.

V. EXPERIMENTAL STUDIES OF COMPRESSION

There are two general classes of studies relating to compression amplification in hearing aids. The first class of experiments utilise laboratory type equipment to simulate the effects of compression in hearing aids. The great advantage of this approach is that the parameters of the compressor can usually be controlled and systematically varied in order to observe their effects. Studies utilising laboratory equipment and/or involving the systematic variation of compressor characteristics will be referred to as "laboratory studies".

The second type of study involves the evaluation of actual hearing aids with compression amplifiers. This sort of study is needed to assess the usefulness of compression as it exists in a particular hearing aid. The weakness of this approach is that it is impossible to generalize the results beyond the particular aid tested in the particular test environment. There are so many factors which may vary in the aid [e.g. time constants compression ratio, compression threshold, type of compression (I/O), shape of I/O curve, frequency response, maximum output, distortion level] and in the test situation [type of test material, type of noise, signal to noise ratio, intensity of signal] that a negative result cannot be taken as evidence against compression per se, but only against the particular aid under test. We will call these studies "evaluative studies".

A. Laboratory Studies

Davis et al (1947) used a master hearing aid to investigate the relative effectiveness of peak clipping and compression to limit the hearing aid's output. Monosyllabic word lists were used to measure speech intelligibility for subjects with sensori-neural, conductive and mixed hearing impairment. The authors reported that with peak clipping the performance-intensity (P.I.) function deceased when input levels were increased beyond the clipping threshold. This occurred even if the signal was high pass filtered before clipping. On the other hand, with compression, intelligibility scores remained essentially unchanged when the input level was raised above compression threshold. The instrument used in this study had t_a = 1 ms , t_r = 200 ms , a compression ratio of 10 and "low distortion". It would be classified as a compression limiting system although the recovery time is larger than would be found in most such systems today.

Hudgins et al (1948) confirmed the findings of Davis et al using the same master hearing aid and demonstrated the feasibility of constructing a wearable hearing aid with similar compression characteristics.

Kretsinger and Young (1960) compared the monosyllable speech intelligibility scores of normal hearing subjects through amplifier systems with linear amplification, instantaneous clipping, and compression limiting. Attack time of the compressor was 0.15 ms with a release time of 22 ms and a compression ratio greater than 10. They reported that the intelligibility of the monosyllables imbedded in white noise (S/N ratio = + 3 dB) was markedly improved with compression as compared to linear amplification. "Instantaneous clipping" produced only a marginal improvement in intelligibility compared to the linear condition.

Vargo (1977) presented C.N.C. word lists to normal hearing and sensorineurally hearing impaired subjects under a linear and two compression amplification conditions (with C.R.'s of 2 and 5). For the compression conditions, t_a was equal to 100 ns and tr was equal to 50 ms. Since speech at a normal level activated the compressor, we would classify this as a syllabic compressor. After processing, the peak speech levels were equated and the signals were presented at 10, 20, and 30 dB relative to each group's average S.R.T. The resulting P.I. functions were almost identical for each amplifying system for both groups. Vargo thus concluded that the compression systems were of no benefit. However, we feel the results support the opposite conclusion. The results clearly show that, when sensation level is equated, amplitude compressed speech is no less intelligible than linearly amplified speech. That is, a reduction in the dynamic range of adjacent syllables does not decrease speech intelligibility. In situations more realistic than that used by Vargo, compression systems have two effects on the speech signal. Firstly, this same effect of a reduction in the short term dynamic range occurs, and as Vargo has shown, causes neither an increase nor a decrease in speech intelligibility. Secondly the overall level of the speech is caused to vary less widely from time to time by the action of the compressor. It is thus possible to have input signals which vary over a large amplitude range all result in output signals at or near the S.L. which is associated with optimum speech intelligibility for each individual subject. If this advantage is not offset by any other disadvantages (such as a possible reduced intelligibility of compressed speech), then syllabic compression systems provide a net advantage for speech intelligibility.

Several controlled studies of the effects of attack and release times on the performance of compression amplifiers have been carried out.

Lynn and Carhart (1963) carried out a systematic investigation of the effects of t_a and t_r on speech intelligibility through a compression amplifier. They presented monosyllabic and spandaic word lists through a compressor and systematically varied t_a and t_r. The speech reception threshold (S.R.T.) and intelligibility scores were compared with those obtained with a linear amplifier; t_a's ranged from 6 ms to 85 ms while t_r's ranged from 30 ms to 1200 ms . The compression ratio was greater than 5 and in terms of our classification system the system was operating as a syllabic compressor for the shorter time constant conditions and as an A.V.C. system for the longer time constant conditions. Subjects were divided into three groups on the basis of the aetiology of their hearing loss, the groups being characterised by otoschlerosis, presbyacusis and labyrinthine hydrops. The authors reported that varying the time constants influenced both S.R.T.'s and intelligibility scores and that the effect of compression varied across groups. The otoschlerotic group did not show any improvement with compression but the other groups exhibited slightly improved speech discrimination for the compression condition when shorter attack and release times were employed. All groups exhibited some loss of intelligibility with the longest attack/release times (70/400 ms , 85/1200 ms). The failure to find advantages for compression with the otoschlerotic group can be accounted for in terms of the very high speech discrimination scores obtained by this group with linear amplification. The authors also point out that the reduced speech scores obtained for the long t_a/t_r conditions may simply be due to the reduced presentation level of the material which results under these conditions.

Schweitzer and Causey (1977) investigated the question of the influence of release time on the performance of compression hearing aids. They took 6 hearing aids of the one model and had them modified so that t_r varied from aid to aid. The t_r's used were 7, 15, 35, 70, 150 and 320 msecs. The t_a remained constant at 10 ms. Once again, in terms of our classification, the aids with the four shortest t_r's were performing as syllabic compressors while the aid with the longest t_r would be classified as having A.V.C. Monosyllabic word lists were recorded through the aids mounted on a Kemar mannikin. The six recordings thus obtained were then equalised for intensity. These recorded and equalised lists were presented to groups of normal hearing and sensori-neurally hearing impaired subjects and their articulation scores compared. The normal hearing groups had broad band noise mixed into the earphone at 0 dB signal to noise ratio in order to make the list more difficult. It was concluded that recovery time did have an effect on intelligibility. A recovery time of 70 ms produced the best intelligibility, while the longest (320 ms) and shortest (7 and 15 ms) values of t_r led to a marked decline in intelligibility. The effect was greater for the normal hearing groups than for the hearing impaired group. This may have been caused by the higher scores obtained by the hearing impaired group due to the absence of a masking noise when they were tested. In th case reduced intelligibility found with the 320 msec t_r cannot be explained in terms of level differences. The drop in intelligibility for very short t_r's is predictable (see later discussion).

Johansson (1973) also briefly reported experiments where t_a and t_r were varied systematically to examine their effect on the perception of C.V.C. nonsense syllables. The t_a's were varied between 0.05 and 5 ms whilst t_r's varied between 10 and 1000 ms. Subjects with normal hearing and with sensori-neural hearing impairment were used. For the normal hearing subjects, noise was mixed with the signal after it was compressed. The signal to noise ratio was 5 dB. Johansson reported that vowel intelligibility was unaffected by t_r but that final consonant intelligibility dropped when t_r exceeded about 150 ms. This is not an unexpected result since the final consonant would occur during the period when the compressor's gain level was recovering after the offset of the high intensity vowel. This finding clearly illustrates why it is not desirable to have short attack time in an A.V.C. system. If t_a was also long the gain of the system would not be so drastically reduced during the occurrence of the vowel and consequently the effect of the gain reduction on the following consonant would not be so dramatic.

B. Evaluative Studies

There are numerous studies which could be discussed here. However, discussion will be restricted to those studies carried out in the last decade as earlier work was carried out with instruments which do not have the performance capabilities of more modern instruments. Rintelman (1972) contains an excellent review of the earlier work.

Blegvad (1974) had 42 patients with sensori-neural hearing losses wear two otherwise equivalent B.T.E. aids, one with and one without compression, for a period of two months and then asked them to make value judgements about the aids. The t_a of the compression aid was 40 ms and t_r was 200 ms. The compression was output controlled with the ouput at compression threshold equal to 110 dB S.P.L. and a compression ratio of 3.3. Unfortunately the gain selected by each subject was not reported, so the compression threshold appropriate to each subject cannot be determined. Together with the intermediate value of the time constants, the uncertainty about the compression threshold makes classification difficult. Blegvad reported that only 13 (31%) of his patients expressed a preference for the compression aid. This result is to be expected as we have argued earlier that the combination of long t_a/t_r's with high compression thresholds is inappropriate. Ludvigsen and Nielsen (1975) compared a "Stereophonic" body worn aid with compression with a monaural body worn aid without compression. Obviously this experiment runs the risk of confounding binaural advantage with the possible advantage afforded by the compression. However, because of the physical configuration of the particular "binaural" aid it is unlikely to produce any true binaural advantage. This contention is supported by the authors who state that the results showed no effects due to the stereophonic features of the aid. The compression aid had output controlled compression with a compression threshold of 75 dB S.P.L. The subjects were 16 severely hearing impaired adults. A number of tests, including speech discrimination tests, were carried out on the subjects whilst wearing the aids. The results showed better speech discrimination with the compression aid on a speech test in quiet but better discrimination with the linear aid for speech tests presented in a variety of background noises. A strong preference was shown for the linear aid (13 out of 16) and it was commonly reported that the background noise appeared 'louder' in the compression aid. Brink et al (1975) also compared patient performance while wearing a stereophonic body level aid with compression with that while wearing a different stereophonic body aid without compression. The subjects were 8-11 year old children with severe to profound hearing losses. The authors resported that they could find no statistically significant differences between the aids for performance in speech tests in quiet, speech tests in white noise, or on a noise test involving identification of everyday sounds. In neither of the above two studies were the t_a's and t_r's of the aid indicated. This makes interpretation of their results difficult.

Ruhrberg and Esser (1973) compared monosyllabic speech intelligibility for compressed and linearly amplified signals. Their specially built compression aid had compression ratio of 2.5 and a t_r of less than 20 ms. The t_a was not specified. The word lists were recorded through the compression and linear aids and presented to the subjects with equivalent peak levels (indicated by an impulse sound level meter). The subjects used had cochlear and retro-cochlear hearing impairments. Both groups recorded significantly higher intelligibility scores with compression amplification, with the improvement for the cochlear loss group being greater than that for the retro-cochlear loss group.

Yanick (1973) compared the performance of a custom fitted compression hearing aid to that of the subject's own aid. Subjects were 12 experienced hearing aid users with mild to moderate sensori-neural hearing losses. The compression aid provided syllabic compression and the gain and compression ratio were individually selected for each patient on the basis of their S.R.T. and discomfort levels. Yanick was able to demonstrate a marked advantage for the compression aid in terms of speech intelligibility. He interpreted the the results as indicating that the advantage of the compression aid was due to its overload protection function. A problem arises in interpreting the results of this study because there was no control over the linear aid's characteristics. How well would the subjects have performed with linear amplification if it too had its parameters carefully chosen to be appropriate for each individual hearing loss?

In a second study Yanick (1976) compared the performance of his aid with syllabic compression to that of a similar aid with compression limiting. There were 6 subjects with mild to moderate sensori-neural hearing losses and essentially flat audiograms. The input to the aids was adjusted so that the compression threshold of the limiter was never exceeded. Under these conditions there was no demonstrable advantage for the syllabic compressor. This finding was seen as supporting Yanick's conclusions about the nature of the advantages of the compression system which were demonstrated in the earlier study.

Bernath et al (1976) took two carefully matched groups of hearing impaired subjects (55 per group) and fitted one with compression aids and the other with equivalent aids without compression. The compression aids provided syllabic compression but little other detail is given. Speech tests in quiet were administered to each group at a number of presentation levels (50, 60, 70, 80 dB S.P.L.). For the 60 dB input the compression group showed a statistically significant improvement in intelligibility over the non-compression group. There were no significant differences between groups for the other three presentation levels. This finding is hard to understand as one would expect that any advantage compression had would show most clearly at the highest input levels where distortion caused by peak clipping is most severe and/or at the lowest input levels where the compressor may provide some additional gain over the linear system.

In a second study Bernath et al (1977) took two randomly selected groups of patients (34 per group) and fitted one group with syllabic compressor aids and the other with linear aids. Speech intelligibility scores were obtained for both groups in quiet listening conditions and in a background of extraneous noise. The authors claim that their results indicate that the compression aids achieved significantly better results than the linear aids with noise present. In fact, the superiority of the compressor is very small, and occurs only at the smallest signal to noise ratio employed (10 dB). In quiet, and at a 20 dB signal to noise ratio the linear aid produced the best performance.

It must be concluded that, despite some manufacturers' claims, recent evaluative studies utilising commercially available compression aids have failed to conclusively demonstrate any consistent significant advantages for these instruments either in terms of speech discrimination or client acceptance. As pointed out earlier, this cannot be taken to mean that compression per se is without merit, but merely that the particular aids tested did not appear to offer any advantages in the particular testing situations employed. These findings are perhaps not surprising in light of the findings of Burnett and Bassin (1976) and Burnett and Schweitzer

(1977). These authors examined the characteristics of 81 commercially available compression hearing aids. They reported that a wide range of time constants were found. The t_a's ranged from 1 to 20 ms, whilst t_r's ranged from less than 10 to more than 500 ms. Very few aids had what might be considered on the basis of recent laboratory studies to be optimum time constants for dynamic range compressors. Some of the aids had undesirably high maximum outputs (e.g. 148 dB S.P.L.) so that the compression may be ineffective in preventing discomfort caused by intense sound. In addition some of the aids exhibited severe ringing during overshoots and undershoots and unnecessarily high distortion level. The importance of appropriate time constants and low distortion levels was demonstrated by Nabalek and Robinnette (1975) who used the modified rhyme test to obtain speech intelligibility scores from normal hearing subjects and subjects with sensori-neural hearing losses while wearing seven different commercial hearing aids with compression. These hearing aids differed from each other mainly in terms of their time constants and in the amount of harmonic distortion generated during overshoots. Significant differences in intelligibility scores across aids were found for both subject groups with differences between aids being less obvious for easy listening conditions than for difficult listening conditions. The hearing aids which yielded the highest discrimination scores for both normal and impaired subjects were generally those with the shorter time constants. Exceptions to this trend occurred where aids with short time constants also exhibited high distortion levels during overshoot.

C. Conclusions: Single Channel Compression Systems

1. Compression Limiting

Several laboratory studies have shown that compression limiting provides better speech discrimination than does peak clipping. This is not surprising since the larger distortion caused by peak clipping can be objectively measured, and the deleterious effect of non linear distortion on speech discrimination has been shown many times (e.g. Gioannini and Franzen, 1978). Although some studies using commercial aids have indicated that compression limiting offered no advantage in at least some circumstances, these studies have not adequately controlled or even described some of the important parameters. Brink et al (1975) and Ludvigsen and Nielsen (1975), for example, do not indicate what compression ratio or time constants employed, while Blegvad (1974) used a system with an attack time too long to prevent uncomfortable overloads and a recovery time too long to handle different syllables independently, but not long enough to be a conventional A.V.C. system. We thus conclude that compression limiting is highly desirable in a hearing aid, but that the compression parameters must be carefully chosen.

2. Syllabic Compression

The need for syllabic compression, where different syllables can receive different amounts of gain and where speech at normal levels activates the compressor, has not been as convincingly established. Some studies using syllabic compressors did not employ a reference linear condition (Johansson, 1973; Schweitzer and Causey, 1977), while others did not ensure that the linear amplification used was as well suited to the subjects as the linear portion of the compression system (Yanick, 1973). Of the studies reviewed here, the deletion of these results leaves only four evaluative and four laboratory studies remaining. Brink et al (1975) found no advantage for commercial compression aids in quiet or in noise, while Ludvigsen and Nielsen found them to be better in quiet but worse in noise. As already mentioned however, the parameters of the aids used were not reported. Bernath et al (1976, 1977) found an advantage for commercial compression aids at only 1 of 4 sensation levels and only 1 of 3 S.N.R.'s. In the laboratory studies, Lynne and Carhart (1963) obtained results showing that compression was better for some subjects (depending on the aetiology of the hearing loss); Ruhrberg and Esser (1973) found positive results for compression; and Vargo (1977) reported results which indicate no advantage for compression but which can be reinterpreted to support the use of compression. In the fourth study, Yanick (1976) found no advantage for compression, but only low level signals were tested.

Even though it might be possible to conclusively establish that syllabic compression is superior to linear amplication, it would still need to be demonstrated that it is better than compression limiting before its use could be justified. Only one investigator (Yanick 1973, 1976) has touched upon this issue. Although his results indirectly provide some support for the contention that syllabic compression has no advantage over limiting, his study was restricted to the use of low level signals.

We thus have a tendency for the results from the laboratory studies to support the use of syllabic compression, but a tendency for the evaluative studies not to. This is not surprising given the study referred to earlier showing the wide range of parameters employed in commercial hearing aids, and the generally closer control employed in the laboratory studies. We view this as providing tentative support for syllabic compression.

3. Automatic Volume Control

A lack of evidence precludes us from reaching a firm conclusion on this type of compression system also. However, as Braida et al (1979) have suggested, its successful use in other speech reproducing systems (including special devices for the hard of hearing) indicates that it is potentially useful in hearing aids. We can say from research evidence that the use of a high compression threshold (Blegvad, 1974), or the use of a long recovery time with a short attack time (Johansson, 1973) in A.V.C. systems will lead to unsatisfactory results, but these are predictable from a consideration of the dynamics of the speech signal anyway. Research into the effects of A.V.C. in hearing aids is required, as a study has shown that at least some hearing aid users find it necessary to adjust their hearing aid's volume control quite frequently (Bradford and Byrne, 1973).

Curiously, in research into compression amplification, there has been very little attention paid to response measures other than speech intelligibility. Apart from several survey type studies which have indicated that patients generally prefer hearing aids with linear amplification, there has been no attention paid to subjective measures such as the acceptability, pleasantness etc. of the amplified signal. There is a clear need for such studies to be undertaken.

D. Recent Studies with Multi-Channel Compressors

The syllabic compressors discussed so far have been single channel devices. That is, a single compressor has been used to cover the entire frequency spectrum. Such a compressor has several limitations in relation to its ability to compensate for reduced auditory dynamic range. One limitation relates to the fact that the hearing impaired subject's dynamic range of hearing is likely to be restricted in different amounts over the frequency spectrum, being typically very much reduced at high frequencies and much less so at low frequencies. It follows that if the C.R. is chosen to reduce the dynamic range of speech to match the subject's needs at low frequencies it will fail to achieve the required reduction of dynamic range at high frequences. What is needed for adequate compensation for recruitment is a system which allows the compression ratio and compression threshold to be varied across frequencies.

Another fundamental limitation of all compressors is that during pauses in the signal the background noise will receive relatively large amounts of gain. This leads to complaints of "noisiness" by the user. The reasons for this are illustrated in figure 18.

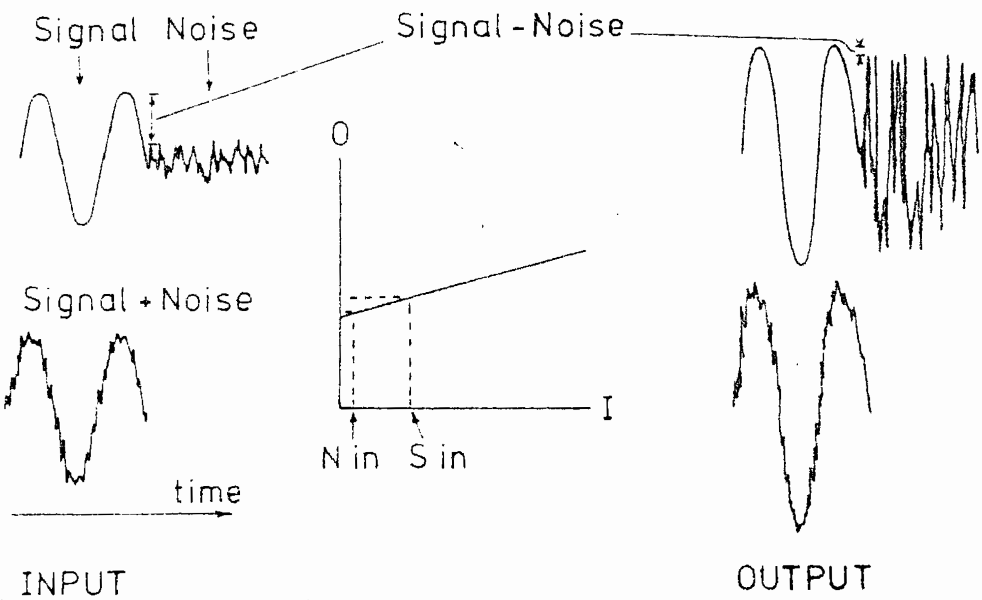


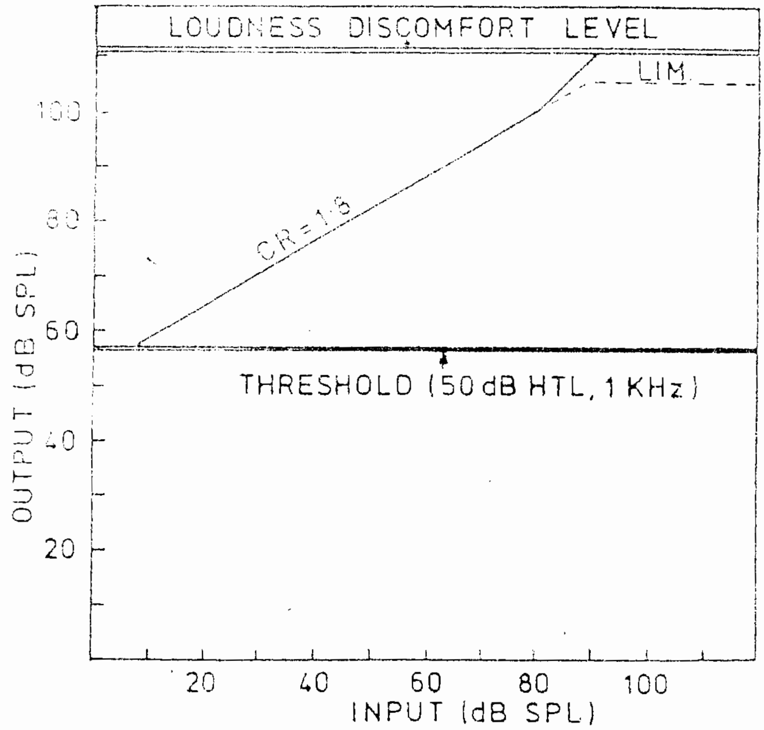
Fig. 18: The upper waveforms show that compression decreases the signal to noise ratio for successive inputs, while the lower waveforms show that the SNR is unaffected when both signal and noise are simultaneously present.

The desirability of compression ratios which vary with frequency has led a number of researchers to investigate multi-channel compression amplifiers. These reasearchers include Villchur(1973), Barfod (1976), Yanick (1976a), Yanick and Drucker (1976), Abramovitz (1979), O'Laughlin (1980), Lippman et al (1981), and Mangold and Leijohn(1979, 1981).

Multi-channel compression act by filtering the incoming signal into a number of frequency bands. Each band of frequencies is amplified independently of the others so that it is possible to set the gain, limiting level, compression threshold, compression ratios and time constants of compression independently in each channel. As well as overcoming the limitations of a single channel compressor outlined above, such a system also permits precise tailoring of the aid's frequency response and maximum output, as well as permitting optimization of time constants on a channel by channel basis. At least one such hearing aid now exists in wearable form (Mangold and Leijohn, 1979).

The basic principle used by researchers with multi-channel compressors has been to compensate for the loudness growth function of the individual. Figure 19 shows the type of compression function required, according to this principle, by a typical case having a sensori-neural hearing loss and recruitment. At threshold, gain is equal to hearing loss. From threshold upwards, there is compression at a rate equal to the degree of recruitment until the level where recruitment is complete. From this point the function becomes linear. However, although this last feature (i.e., linear amplification of high level inputs) would be required for accurate loudness function matching, it is not recommended by any of the advocates of compression amplification. Instead, it is generally agreed that limiting (as shown by the broken line in Figure 19) should be applied at an output level below the client's loudness discomfort level (L.D.L.)

Fig. 19: Type of compression function typically required to compensate for loudness recruitment.



The most precise attempt to use multichannel compression to compensate for recruitment is the study reported by Barfod (1976). He chose compression ratios, for up to 4 channels, on the basis of equal loudness contours, established for each subject at 10 dB intervals. He found no advantage for compression over linear amplification, when both were combined with careful selection of frequency response. (In fact, compression was detrimental when less than 4 channels were used). On the other hand, both Villchur and Yanick have reported advantages for two channel compression systems. However, the studies by both these authors have been criticized on the grounds that the linear amplification which was used was not optimized, and that this may have inflated the apparent advantages of compression (Braida et al., 1979). It may also be significant that Barfod's cases were less severely hearing-impaired than Villchur's.

The concept of recruitment matching would not require the use of expansion. However, Villchur speculated that expansion of low level signals might be useful to improve signal to noise ratio (S.N.R.). This was tested by Yanick and Drucker (1976) who reported that a combination of expansion and compression was superior to compression alone which, in turn, was superior to linear amplification. This apparent demonstration that expansion below compression threshold is advantageous has been queried by Braida et al. According to these authors, the signal to noise ratios used by Yanick and Drucker were such that the system would operate in the compression mode virtually continuously, and therefore, any demonstrated advantage for expansion is inexplicable.

Mangold and Leijohn (1979, 1981) performed monosyllabic speech tests, C.V.C. nonsense syllable tests and tri-syllabic (V.C.V.C.V.) word tests on fourteen subjects whilst wearing their own conventional hearing aid, a three channel experimental aid with individualised frequency response shaping but no compression, and the same three channel aid with frequency response shaping, A.V.C. and syllabic compression. The t_a of the A.V.C. was less than 1 ms while the t_r was said to be 10 seconds in the 1979 report and 200 ms in the 1981 version of the report. The t_a's and t_r's of the output controlled syllabic compressor were independently set in each channel, being shortest in the highest frequency band and longest in the low frequency band. The t_a's were in the range 1-3 ms while the t_r's were in the range 3-9 ms . Each subject wore each aid for a week before testing and each was tested with all three aids. "Equalisation" (frequency shaping) was achieved by plotting the subject's equal loudness contour for third octave filtered bands of pink noise centred on the common audiometric frequencies. This equal loudness contour passed through the most comfortable loudness level established independently for octave band filtered cocktail party noise centred on 1 kH_z. Gain was then chosen in each band to restore the equal loudness contour to "normal" (The "normal" curve was established in the same way as above with 7 normal hearing persons). The compression ratio was selected by plotting another equal loudness contour for the same stimulus but 30 dB lower in intensity. The compression ratio was selected to compensate for the degree of narrowing of the contours which was observed. Monosyllables and tri-syllables were presented in quiet (S.N.R. > 40 dB) and in cocktail party noise with S.N.R.'s of 10 and 20 dB. The C.V.C. nonsense syllables were presented in quiet.

The authors noted that the results differed a great deal between subjects. Three subjects showed significantly improved intelligibility for the multichannel aid over the conventional aid while the others showed the opposite effect or no difference. The results failed to show any consistent superiority for the compression condition over the multichannel frequency shaping only condition, although a significant improvement was shown by some subjects for the noisy listening conditions.

Abramovitz (1979) investigated a four channel compression system and concluded that although significant individual differences and interactions were evident, the trend appeared to be that linear amplifiers without peak clipping yielded the best intelligibility scores for most subjects, both in quiet and in noise, and whether with a flat frequency response or with an individually shaped frequency response. By contrast, linear amplification with peak clipping yielded poorer speech intelligibility scores more often than did either linear amplification without clipping or compression amplification.

Walker and Byrne (1980) also presented some data which supports the contention that multi-band compression may be of significant benefit to some patients but not to others.

O'Laughlin (1980) evaluated a three channel compression system and also concluded that his results failed to support the hypothesis that multi-channel compression leads to markedly improved speech perception.

Lippman et al (1981) conducted experiments with two 16 channel computer controlled compression systems. One system was selected to restore normal loudness and the other had reduced high frequency emphasis and reduced compression ratios. Neither compression system was superior to the best (of four) linear systems.

Villchur (1979) has levelled a number of criticisms at those studies which have failed to demonstrate advantages for multi-channel compression. Firstly, he criticizes the manner in which the speech test material commonly used is recorded. Speech tests are routinely recorded with the talker facing a VU meter and monitoring his own voice level to keep it constant. Villchur argues that this artificially and severely reduces the dynamic range of speech as it is encountered in real life and thus biases the test results against compression. Secondly, Villchur criticises the failure of experimenters to provide "post compression equalisation" (i.e. appropriate frequency response shaping) in their compression amplifiers. This is ironical in light of the criticism levelled at Villchur's own work that he biased the results in favour of compression by failing to provide adequate frequency shaping in his linear condition. Thirdly, he criticizes the choice of subjects for these studies. He points out that when a person is only mildly or moderately hard of hearing, even with recruitment, there is usually sufficient residual dynamic range of hearing remaining for linear amplification to be successful. There is no reason to expect such subjects to show superior results with compression. Villchur maintains that there is ample evidence that subjects with a very much reduced dynamic range of hearing benefit significantly from multi-channel syllabic compression. Even leaving aside the positive findings of Villchur himself (Villchur 1973) and Yanick (1976a) whose experiments have been criticized for introducing a bias in favour of compression, there is some evidence in Abramovitz (1979), Mangold and Leijohn (1979), Lippmann et al (1980) and Walker and Byrne (1980) that some subjects show marked improvement in speech intelligibility with multichannel syllabic compression as compared to multichannel amplification without compression. It is clear that further work is needed in this area, both to confirm the validity of the general approach and to determine optimal parameters of the system.

VI. RECOMMENDATIONS

In this section, we will attempt to make recommendations about the most desirable arrangement of gain controls, frequency shaping networks, and the compression feedback point, and about the optimum values of the parameters of compression hearing aids. As the previous review section has indicated, there is more unknown about the effect of compression hearing aids on speech perception than there is known. Consequently, we will in some cases, be taking the best guess that we can at this time. Other recommendations will be better founded, either on well accepted theoretical grounds, or on evidence from experimental studies, or on both. We shall indicate the reasons for each recommendation and the degree of confidence we have in it. Recommendations for each type of system will be given separately, but with the greatest emphasis on compression limiting systems.

A. Compression Limiting

The recommendations in this section assume that the compression system is functioning solely to prevent amplified intense input signals from exceeding the wearer's loudness discomfort level (L.D.L.). The system block diagram is shown in figure 20.

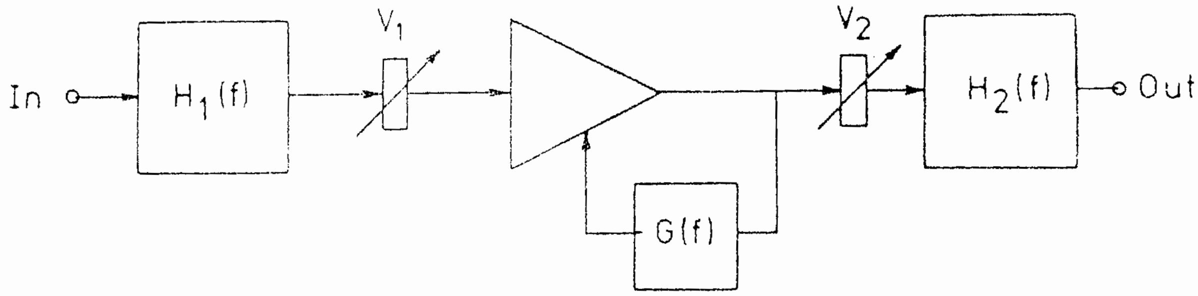


Fig. 20: Block diagram of possible locations of frequency dependent devices and gain controls.

1. Arrangement of Components: Input or Output Control

In the system shown in figure 20, the order of $H_1(f)$ and V_1 can be interchanged without affecting the aid's operation, as can the order of V_2 and $H_2(f)$, since these four subsystems are all linear in operation. However, the two gain adjusters V_1 and V_2 have different effects and one must be chosen to be the manually adjusted volume control. For compression limiting devices, we are unequivocally in favour of output control. That is, V_1 should be the volume control. As the effect of V_2 is to adjust the maximum output of the aid, as well as the gain in the linear regions, V_2 should be labelled as an M.P.O. adjustment. Since there is no reason to believe that a client's M.P.O. requirement varies widely with time, a screwdriver adjustment should suffice here.

2. Arrangement of Components: Frequency Shaping

The conclusions reached here are based heavily on the theoretical discussion presented in section III. The desired response characteristics are the M.P.O. curves, frequency response (F.R.) in the linear region (and in the compression region for simultaneous inputs), and the compression threshold curve, expressed either in terms of the input level at which the compressor just starts to operate (I.C.T.) or at the corresponding output level (O.C.T.). In terms of the system components, these quantities can be expressed as follows:

$$\text{M.P.O.} = H_2/G \quad \ldots\ldots \quad 1$$
$$\text{F.R.} = H_1H_2 \quad \ldots\ldots \quad 2$$
$$\text{I.C.T.} = 1/(H_1G) \quad \ldots\ldots \quad 3$$
$$\text{O.C.T.} = H_2/G = \text{M.P.O.} \quad \ldots\ldots \quad 4$$

Neither of the compression thresholds represent independent choices once the M.P.O. and F.R. curves have been chosen. The O.C.T. curve is identical to the M.P.O. curve. Also, equations 1 to 3 can be rearranged to show that:

$$\text{I.C.T.} = \text{M.P.O./F.R.}$$

(This can also be shown quite simply by drawing the M.P.O. and linear portions of an I/O curve). Thus, we are left with only two independent choices, M.P.O. and F.R., and with three independent filter responses to manipulate in order to achieve them. We thus have considerable freedom in the selection of where to put the frequency shaping. The location of the frequency shaping devices will probably be influenced by technical considerations associated with maintaining an adequate S.N.R. within the aid. Apart from this, there may be a slight preference in having the majority of the shaping in the $H_2(f)$ box, since under these conditions, the same shaping is applied to sequential inputs as it is to simultaneous inputs. We thus recommend that tone controls in particular be placed after the feedback point unless there are strong technical reasons for doing otherwise.

3. Compression Threshold

As explained above, the compression threshold is determined once the linear gain and the M.P.O. are selected. Thus, the average value of the C.T., as well as its variation with frequency is not a design parameter which needs to be selected.

4. Compression Ratio

If the aid is to function as a limiter, the preferred value for the C.R. is clearly a value close to infinity. (This is possible with analog multipliers). Values in excess of 10 would almost certainly suffice, since signals 20 dB in excess of the "normal" input range would be needed to raise the output by 2 dB above the desired maximum output level. Client's loudness discomfort levels are not normally known to better than this accuracy. Compression ratios even lower than 10 would probably be suitable in many cases where the client still retained a medium to large dynamic range and/or was not often in a very noisy environment.

5. Attack Time

It is necessary that t_a be short enough to prevent loud sounds from causing excessive discomfort. For normal hearing people, the loudness of tone bursts or white noise bursts decreases at a rate of approximately 10 dB per decade of duration once the duration becomes less than a certain value, which is usually in the range 100 to 250 ms. (Pollack, 1958). The threshold of tone bursts varies in the same way (Plomp and Bouman, 1959). Unfortunately, the variation of loudness with duration does not appear to have been measured for the hearing impaired. Many studies have shown that when measuring threshold, both the break point and rate of change of loudness with duration are less than for normals. There is thus the possibility that these values are also less for the loudness perception task. An implication of this is that even quite short overshoots from the compressor amplifier may cause loudness discomfort. However, Miskolczy-Fodor (1960) has shown a close correlation between the presence of recruitment and reduced temporal integration. If this also holds true for sounds in the vicinity of L.D.L., then the hearing impaired should have close to normal temporal integration functions, as recruitment is normally complete by the time L.D.L. is exceeded. We shall thus base our recommendation for t_a on normal temporal integration functions, with the proviso that somewhat lower values may be necessary if it is shown that different functions are applicable to the hearing impaired. If we assume a break point of 150 ms and a slope of 10 dB/decade, then a steady state signal which would cause an output 20 dB in excess of L.D.L. will cause a loudness sensation only 2 dB above L.D.L. if its duration can be limited to 2.5 ms. Since the compressor overshoot is not rectangular in shape, it can be calculated that an equivalent duration of 2.5 ms will be obtained from an exponential decay with a time constant of 5 ms, which in turn would be measured as an attack time of 18 ms when a 20 dB input step was used. We thus recommend that the attack time be shorter than 18 ms.

The lower limit for t_a comes from the need not to have it so short that extremely brief noise impulses cause an unnecessary gain reduction. Even though hearing aids can typically only pass frequencies up to $5kH_z$, this is still sufficient to enable impulses with a rise/fall time of 0.1 ms to be passed. A value for t_a of this order would thus allow impulses of duration 0.2 ms to affect the gain even though they would not cause loudness discomfort unless they were extremely intense. In the absence of information about the duration and intensity of impulsive noise in common listening situations, we are thus left with the rather inconclusive result that t_a should be greater than 0.2 ms but less than 18 ms.

Evidence from the literature is not much more informative, except to indicate that the choice of t_a may not be at all critical, at least for some signal to noise ratios. Only three studies have systematically examined the effect of attack and recovery time on speech intelligibility. These have all used systems which we would classify as syllabic compressors, but their findings regarding the effect of t_a and t_r on speech intelligibility can probably be applied to compression limiting systems. If particular values of t_a and t_r do not have an adverse effect on speech intelligibility when the compressor is in continuous operation (as in syllabic compressors), then they should not have adverse effects when the compressor is activated only occasionally (as in limiting compressors). In one of the studies which systematically varied t_a, Johansson (1973) found that as t_a increased from 0.05 ms, no deterioration in speech intelligibility occurred until t_a exceeded 1.5 ms. Unfortunately t_r covaried with t_a in this experiment, so it is not clear whether it was the increase in t_a or in t_r which caused the difficulty. The corresponding value of t_r in this experiment was 300 ms. As the results obtained by Lynn and Carhart (1963) and Schweitzer and Causey (1977) indicate that recovery times in excess of 300 ms are harmful to speech intelligibility, we can guess that it was really the t_r variation in Johansson's experiment which caused the effect. Lynn and Carhart obtained better S.R.T.'s and speech intelligibility scores with 5 ms than with 20 ms attack times. The literature thus tells us only that intelligibility is not affected for t_a values up to 1.5 ms but is definitely adversely affected by the time it reaches 20 ms. When coupled with the above discussion on loudness protection requirements, we can thus estimate that an optimum value for t_a would be in the range 2 to 5 ms. We can only state with confidence, however, that it should be in the range 0.5 to 20 ms.

6. Recovery Time

In one sense, specification of t_r is easier than that of t_a, since it has been shown to markedly affect speech intelligibility. On the other hand, this makes the selection of the optimum value more crucial. The upper limit of acceptable values for t_r comes from the need to prevent gain reductions which have been caused by intense signals from lasting too long. It is hard to calculate theoretically at what point the gain reduction duration becomes excessive. At one extreme, a soft word may follow an intense word after an interval of some hundreds of milliseconds. At the other extreme, a weak consonant may follow immediately after, or even overlap, an intense vowel, as in the word "harp".

A lower limit for the value of t_r comes from the need to avoid creating the distortion that occurs when the gain changes significantly during each voicing period of the speech signal. (We are assuming here that the speech signal of interest will sometimes be of an intensity sufficient to activate the compression limiter.) As the lowest frequency of interest for speech will be about 80 Hz (for the male voice), we therefore require the gain to be stable during each 12 ms period. Figure 21(b) shows how the gain decreases slightly during each waveform peak and then rises gradually until the following peak occurs. Although many authors have claimed that the lowest frequency of interest is determined by the low frequency cut-off point of the hearing aid, this does not appear to us to be justified. Even when speech is high pass filtered with a 500 Hz cut-off, a vowel still maintains a waveform periodicity equal to its fundamental frequency. The hearing aid gain will thus still vary at this lower rate, despite the absence of any significant energy below 500 Hz. The result of this gain variation is a non-linear distortion which will eventually degrade the intelligibility of the speech signal.

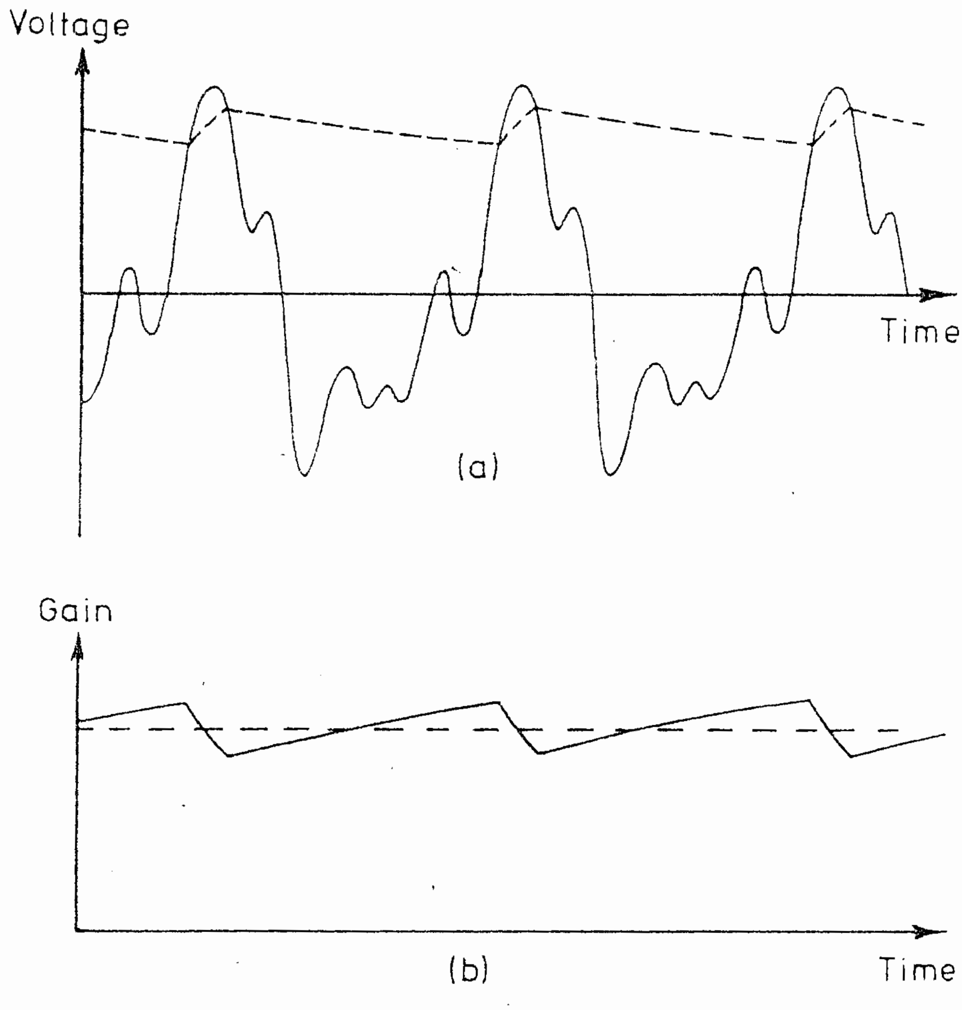


Fig. 21: Gain variations during low frequency signals: (a) Speech waveform (firm line) and signal fed to compressor (dotted line); (b) Resulting gain variation (firm line) and average or d.c. component of gain (dotted line).

An upper limit on the permissable gain variation can be calculated as follows. By referring to figure 21(b), it can be seen that the gain curve can be well represented by a d.c. term plus a sawtooth waveform. Since we must multiply this gain curve by the input waveform to get the resulting output, the output spectrum can be found by convolving the two input spectra. Convolution of the signal spectrum with the d.c. term gives the wanted output signal, while convolution with the components corresponding to the sawtooth results in the distortion components. If we accept a maximum of 10% distortion, then the rms value of the sawtooth waveform must be limited to 10% of the d.c. component. The peak to peak gain variation is thus limited to 34% of the average gain, or, in more familiar units, 3 dB. Getting from this figure of a 3 dB gain variation during each waveform period, to a minimum allowable recovery time is quite straight forward, but can only be done exactly once the relationship between the compressor gain and the feedback voltage is known. This will vary from aid to aid, depending amongst other things on the compression ratio. However, for an infinite compression ratio (worst case situation), the gain must be inversely proportional to the output level and thus the maximum allowable fluctuation in the feedback voltage is also 3 dB. For a fundamental frequency of 80 Hz, the minimum time constant is about 33 ms. The recovery time depends on the size of the level change used to measure it, but a level change of 25 dB results in a recovery time of 4.2 times the time constant. We thus estimate that the minimum recovery time that enables the distortion to be kept within 10% for a signal of fundamental frequency of 80 Hz is 140 ms.

The type of distortion generated by the changing gain should not be particularly objectionable for speech signals. As already discussed, the spectrum of the gain variation consists of a large d.c. component, together with (hopefully) much smaller components at the harmonics, of the sawtooth wave. In the convolution process, each of these smaller components creates a pair of distortion components surrounding each component in the original signal spectrum. However, because the sawtooth and the input signal share the same fundamental frequency, each distortion component falls on a harmonic of the input signal. Also, the largest distortion components are only displaced from their undistorted position by an amount equal to the frequency of the fundamental. The net result of this type of distortion is that the spectrum formants are broadened slightly, and the distortion products tend to be masked by the nearby signal components. This is in marked contrast to the intermodulation distortion that is caused by peak clipping. Here, the distortion products can be generated at frequencies quite remote from those present in the input signal and are consequently more noticeable.

This analysis of the distortion process has been presented here to indicate that a given amount of distortion is probably more tolerable when it has been generated by a compression amplifier than when it has been generated by a peak clipper. The 10% distortion figure referred to above may need reassessing if suitable subjective data becomes available.

Results from experimental studies on the effect of recovery time on speech intelligibility are in reasonable agreement about an upper limit for t_r. Johansson (1973) reported a decrease in intelligibility when t_r was increased from 300 to 1000 ms, but no effect for values down to 10 ms. Lynn and Carhart (1963) reported a small decrease in intelligibility when t_r increased from 150 to 400 ms and a large decrease when it went to 1200 ms. They also found no effect as t_r was reduced down to 30 ms. For normal hearers tested in noise, Schweitzer and Causey (1977) found that recovery times of 70 ms were better than either higher or lower values. For their hearing impaired subjects (tested in quiet), t_r had much less effect but there was a small decrease when t_r increased from 150 to 300 ms.

Based upon these results and the theoretical considerations discussed above, it appears that values for t_r outside the range of 50 to 300 ms are not suitable. We thus recommend a value in the centre of this range, say between 100 and 150 ms. Values lower than this may be quite suitable, or even advantageous, if the compression threshold is exceeded only occasionally.

B. Syllabic Compression

As has already been discussed, there is some evidence that syllabic compression is better than a linear system for some subjects, but no convincing evidence that it is in any way better than compression limiting. Consequently, we do not recommend its use on a routine clinical basis until further research evidence shows it to be justified. The recommendations about the parameters for this sytem will thus be quite brief.

1. Arrangement of Components: Input or Output Control

Again we are in favour of output control, on the basis that the input level will change more from time to time than will the intensity range in which the client would like the output to fall.

2. Arrangement of Components: Frequency Shaping

The arguments and conclusions are identical to those given for compression limiting systems.

3. Compression Threshold and Compression Ratio

There does not as yet appear to be a proven way for selecting these parameters. The most common method has been to shape the I/O curve so that normal loudness contours are established when the aid is being worn. While this has benefited some subjects, it has proved disadvantageous to others. This procedure can only be fully carried out with a multiband aid.

4. Attack and Recovery Time

The same arguments and conclusions apply as for compression limiting.

C. Automatic Volume Control

As the use of A.V.C. in hearing aids has not been sufficiently investigated, we will not make any recommendations about suitable parameter values. However, there do not appear to be any fundamental differences between the requirements of an A.V.C. system for use in hearing aids and the requirements for use in other sound reproducing areas.

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