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CONSOLIDATING COMPRESSION AND REVISITING EXPANSION: AN ALTERNATIVE AMPLIFICATION RULE FOR WIDE DYNAMIC RANGE COMPRESSION

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ABSTRACT

The fitting process of modern compression hearing aids is becoming increasingly more complex as the processing capabilities of modern hearing aid devices continue to grow. However, a simplification of hearing aid parameters would enable more accessible user customization of a hearing aid, which is especially applicable in the growing field of over-the-counter devices. This work presents an alternative to the conventional compression input/output curve which consolidates numerous parameters while retaining equivalent performance, and revisits the concept of expansion in compression devices. The consolidated amplification rule is a continuous function without knee points, controlled by two parameters – gain and compression ratio. Experimental results have shown that the expansion properties of the function reduce amplification of low-level ambient noise by up to 4 dB, and improves signal-to-noise ratio by up to 1 dB (25%) in moderate noise conditions.

Index Terms— Wide dynamic range compression, over-the-counter hearing aids, self fitting, expansion, open source

1. INTRODUCTION

Wide Dynamic Range Compression (WDRC), the process of reducing the dynamic range of an audio signal, is used in nearly all modern hearing aids [1]. However, there are numerous issues with compressive hearing aids leading to user dissatisfaction [2], which include complaints about speech intelligibility, sound pleasantness, distortion, and excessive amplification of background noise.

With the growing number of adjustable hearing aid parameters, it is also becoming increasingly more difficult to fit a hearing aid [3]. Algorithms have been developed to generate WDRC prescription parameters from an input audiogram, the most prominent of which is the NAL-NL2 algorithm [4]. However, the matter of personal preference makes it impossible to develop a universal fitting procedure to satisfy all users, which places a responsibility on the audiologist to analyze user feedback and deduce the necessary hearing aid adjustments.

The purpose of this work is to consolidate the complexity of conventional WDRC while preserving all of its functionality, and also to propose a method of improving the issue of background noise amplification. In this paper, we present a new WDRC input-output curve with the following properties:

1. Gain and compression are decoupled parameters.
2. The number of parameters defining the curve is reduced to only gain and compression.
3. Reduced amplification of objectionable background noise signals.
4. The replacement of angular knee points with a continuous function.

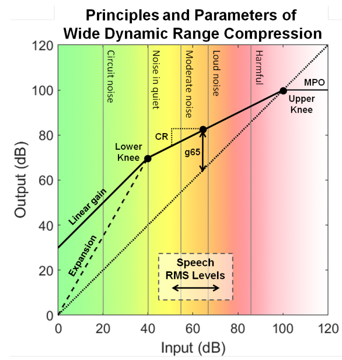

Figure 1: An example of a conventional WDRC input-output curve, as well as typical speech and noise levels in dB SPL.

Experimental results have shown that the consolidated curve reduces amplification of low-level ambient noise by up to 4 dB, and improves signal-to-noise ratio by up to 1 dB (25%) in moderate noise conditions. The proposed strategy also has high applicability in the growing field of over-the-counter hearing aids by allowing a user more intuitive and versatile control over the hearing aid.

2. BACKGROUND

Wide dynamic range compression (WDRC) is conventionally implemented as a piece-wise segmented curve, such as the one pictured in Fig. 1, with a region of hard-limiting above the upper knee point, a region of compression, and a region of linear (constant) gain below the lower knee point. Such a compression curve is defined by a set of four parameters – $g65$, Compression Ratio (CR), Maximum Power Output (MPO), and $knee_{low}$, where $g65$ is the gain at an input level of 65 dB SPL. Alternatively, parameters $g65$ and CR can be replaced with $g50$ and $g80$.

Although WDRC has been shown to improve audibility [1], a number of adverse effects have been observed for speech in noise. Empirical and analytical studies have shown that compression reduces long-term average output SNR [5, 6, 7, 8], which is a highly important metric for hearing aid outcomes [9]. Consistent with expectations, the reduction in output SNR increases with an increase in input SNR [6, 8, 10], since quieter signals receive greater amplification. Moreover, as the number of WDRC channels increases, output SNR further decreases [10], likely due to the distribution of

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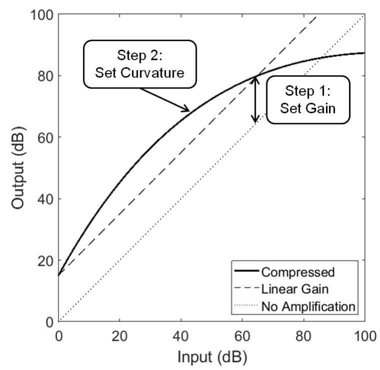

Figure 2: The proposed amplification curve is governed by two parameters – gain and compression (curvature).

signal energy across a greater number of bands, resulting in effectively lower signal levels. The amplification of low-level signals may also cause user dissatisfaction due to the excessive audibility of microphone, circuit, and low-level ambient noise [2, 11].

One solution for improving the output SNR and reducing the noisiness of a hearing aid is the use of expansion [12, 13, 14]. Expansion, as shown in Fig. 1, is the opposite of compression, wherein gain is proportional to input levels, which serves the purpose of applying less gain to signals of lesser value, such as noise. Studies have shown that the use of expansion can significantly improve the subjective perception of hearing aid sound quality, but at the same time, it can also degrade speech recognition [14]. Although expansion has shown positive results, research in this area has been hindered by the historical lack of expansion features on commercial hearing aids. In modern devices, expansion has become a useful feature on commercial hearing aids, but there is a lack of research on how to best to utilize this approach (NAL-NL2 does not offer expansion gains), and whether the speech intelligibility drawbacks can be mitigated.

We believe that some of the drawbacks of expansion identified in previous studies may have resulted from using expansion too aggressively. In this paper, we revisit expansion techniques in the context of realistic listening scenarios, compression, and linear amplification.

3. THEORY

The objective of the proposed amplification rule is to define a function with separable parameters for controlling volume (vertical displacement) and compression (curvature). The function must yield linear amplification when compression is disabled, and a smooth curve with logarithm-like behavior when compression is enabled. The proposed amplification curve is shown in Fig. 2, and has the following mathematical form:

$$y = Cxe^{-\alpha x} + b \tag{1}$$

Where x is the input level in dB and y is the output level in dB, and the controlling parameters are:

- α : Controls the curvature of the function. When α is set to zero, the curve is a straight line.
- b : Sets the vertical displacement of the curve, which corresponds to overall volume.
- C : Sets the slope of the curve at the origin, which affects the tilt of the curve.

3.1. Comparison with Conventional Amplification

The proposed amplification curve provides equivalent performance to a conventional piece-wise WDRC curve when three parameters are matched – $g65$, CR, and linear gain, which is equivalent to the y-intercept, and which we will call $g0$. To this end, we substitute values into (1) to construct a system of three equations using two data points, $g65$ and $g0$, as well as the derivative $^1/_{CR}$.

$$65 + g65 = C \times 65 \times e^{-65\alpha} + b \tag{2a}$$

$$g0 = b \tag{2b}$$

$$\frac{1}{CR} = Ce^{-65\alpha}(1 - 65\alpha) \tag{2c}$$

The solution to this system of equations requires numerical computational methods. However, this offline calculation only needs to be performed once, resulting in a curve which matches the performance of conventional WDRC, as seen in Fig. 3 (left).

3.2. Amplification with Expansion

Although the proposed amplification curve can be substituted for a conventional curve, the purpose of this work is to rethink the way compression is addressed in hearing aids. As shown in Fig. 2, we propose a function which supplies constant linear gain to a signal and then balloons outward to provide compression in the region where the signal of interest lies. Contrary to conventional WDRC, the gain does not remain constant for all low-level signals, and instead, falls back to the initial linear value. This reduction of gain for quiet signals is known as expansion.

To accomplish this goal, let us simplify (2a-c) by substituting the expression $^1/_{e^{-65\alpha}}$ for parameter C in (2c), such that parameter C becomes a function of α:

$$\frac{1}{CR} = \frac{1}{e^{-65\alpha}}e^{-65\alpha}(1 - 65\alpha) = 1 - 65\alpha \tag{3}$$

With this substitution, only one unknown variable remains, yielding a closed form solution for α. Subsequently, all other unknown parameters can be solved for, yielding the following solutions for parameters α, b, and C:

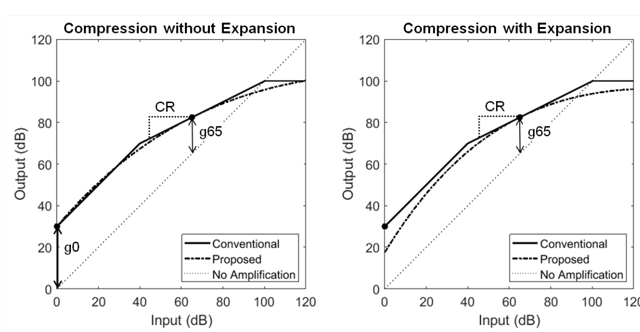

Figure 3: (Left) The proposed curve matched to a conventional WDRC curve; (Right) The proposed curve with expansion.

Table 1: Noise levels, speech levels, and signal-to-noise ratios of generalized listening environments for hearing aid users

Environment	**Noise Level (dB)**	**Speech Level (dB)**	**SNR (dB)**
Very Quiet	45	65	20
Quiet	50	65	15
Moderate	60	68	8
Loud	70	70	0

$$\alpha = \frac{1}{65}\left(1 - \frac{1}{CR}\right) \tag{4a}$$

$$b = g65 \tag{4b}$$

$$C = e^{65\alpha} \tag{4c}$$

The significance of the results in (4a-c) is that the curve in Fig. 2 is defined entirely with familiar audiological parameters – $g65$ for overall gain and CR for compression. A comparison between the proposed amplification curve and a conventional WDRC curve without expansion is shown in Fig. 3 (right).

4. EXPERIMENTAL RESULTS

4.1. Experimental Setup

To validate the effectiveness of the proposed method for audio compression, we processed recorded audio signals using a digital software hearing aid [15] programmed with the proposed input-output amplification curve and with conventional NAL-NL2 standard prescription gains [4]. In the first experiment, noise signals were passed through the hearing aid processing chain, and RMS output levels in dB SPL were recorded. In the second experiment, speech signals in the presence of noise were passed through the same hearing aid processing chain, and output SNR levels in dB were recorded. Results were compared against the NAL-NL2 prescription method – a standard gain prescription algorithm in hearing aid industry and research.

A number of studies have been conducted to explore and quantify the real-world listening environments encountered by hearing aid users and individuals with normal hearing. One of the earliest studies by Pearsons et al. [16] collected speech, noise, and SNR data in various environments with normal hearing participants. Later, similar studies for hearing impaired users were conducted by Smeds et al. [17] using data from [18], and later by Wu et al. [19]. Although findings vary slightly between the different studies, key takeaway findings are consistent through the three studies. Summarizing the data found in [16, 17, 19], we determined a set of representative listening situations, summarized in Table 1, which we used in our experiments. (In some of the studies listed above, sound levels are reported both in units of dB, and in weighted units of dBA. Since the hearing aid operates on unweighted signals, we will be using dB units throughout this work)

Speech data was obtained from the Clarity Speech Corpus [20] – a public dataset created for the Clarity Project, which includes short high quality recordings of speech in the English language by 40 speakers of both genders. Noise data was obtained through the FSD50k Freesound Dataset Project [21] – a public dataset of crowd sourced human annotated audio recordings spanning a wide variety of sound categories. Of the 200 classes of recordings available in the FSD50k dataset, we selected five classes that represent three

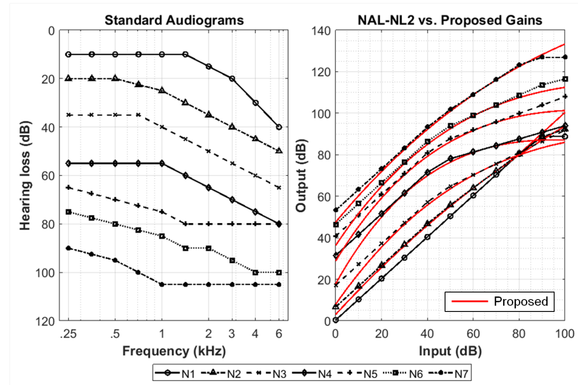

Figure 4: Standard hearing loss audiograms N1-N7 (left); Examples of respective NAL-NL2 prescription gains, with overlayed proposed counterparts (right).

Table 2: Output levels in dB RMS of low level ambient noise after hearing aid processing using NAL-NL2 prescription gains and the proposed amplification rule.

	Very Quiet Noise (45 dB RMS)									
	Dishwasher			**Fan**			**Microwave**			
	NAL.	Prop.	Δ	NAL.	Prop.	Δ	NAL	Prop.	Δ	**Avg Δ**
N1	55.6	53.4	2.2	51.7	50	1.7	51.6	49.7	1.9	**1.9**
N2	65.7	62.1	3.6	60.6	56.7	3.9	60.9	56.5	4.4	**4.0**
N3	74	70.1	3.9	68.8	64.5	4.3	69.1	64.3	4.8	**4.3**
N4	84.2	80.5	3.7	79.2	75	4.2	79.6	75	4.6	**4.2**
N5	90.5	87.3	3.2	85.7	82.3	3.4	86.4	82.7	3.7	**3.4**
N6	97.3	94.7	2.6	92.3	89.3	3	93.1	89.9	3.2	**2.9**
N7	101	99.4	1.6	97.4	95.8	1.6	97.7	96.1	1.6	**1.6**
S1	60.5	57.3	3.2	55.6	52.5	3.1	55.3	51.9	3.4	**3.2**
S2	71.9	69	2.9	66.2	62.6	3.6	67.2	63.3	3.9	**3.5**
S3	85.4	82.1	3.3	79.8	75.3	4.5	80.5	75.9	4.6	**4.1**

common types of listening situations: indoor noise, traffic noise, and babble noise [17, 19]. The selected sound classes are "dishwasher", "fan", "microwave", "car", and "chatter", which we will call "babble" for consistency with audiology literature.

For all experiments, proposed gains are compared to NAL-NL2 prescription gains, with $knee_{low}$ set to 45 dB, attack time set to 5 ms, and release time set to 100 ms, in accordance with common practice [22, 1]. The hearing loss patterns used in our experiments are the ten standard hearing loss audiograms specified by the International Standard for Measuring Advanced Digital Hearing Aids (ISMADHA) [23]. The standard audiograms, their corresponding NAL-NL2 prescription gains, and the respective proposed amplification curves with matching $g65$ and CR are pictured in Fig. 4.

The hearing aid software used for these experiments is an open source digital hearing aid, and is part of the Open Speech Platform for hearing aid research [15]. Subband decomposition is performed in eleven frequency bands, corresponding to the standard audiometric frequencies [24]. The open source nature of the digital hearing aid makes it possible to access internal gain values and obtain an exact measurement of output signal and noise power for SNR calculations.

4.2. Noise in quiet conditions

In this experiment, we pass low level noise signals through the hearing aid and record output RMS levels for the NAL-NL2 prescription

Table 3: Output SNR levels in dB of speech in noise compressed with NAL-NL2 prescription gains and with proposed compression gains, as well as the average difference in output SNRs.

	Speech in Quiet (15 dB SNR)						
	Dishwasher		**Car**		**Babble**		
	NAL.	Prop.	NAL.	Prop.	NAL	Prop.	**Avg Δ (%)**
N1	13.1	13.7	12.8	13.3	12	12.5	**0.5 (23%)**
N2	11.4	12.4	11.2	12	9.7	10.5	**0.9 (20%)**
N3	10.9	11.8	10.7	11.5	9	9.9	**0.9 (18%)**
N4	10.7	11.7	10.3	11.2	8.6	9.4	**0.9 (18%)**
N5	11.2	12.2	10.4	11.2	8.5	9.4	**0.9 (18%)**
N6	12	13.1	11.2	12.3	9.4	10.3	**1.0 (25%)**
N7	13.1	14	12.2	12.9	10.6	11.3	**0.8 (25%)**
S1	12.1	13	12.6	13.3	11.5	12.3	**0.8 (27%)**
S2	12.6	13.4	12.1	12.8	10.2	10.9	**0.7 (22%)**
S3	11.8	12.7	11.5	12.4	9.7	10.5	**0.9 (22%)**

	Speech in Moderate Noise (8 dB SNR)						
	Dishwasher		**Car**		**Babble**		
	NAL.	Prop.	NAL.	Prop.	NAL	Prop.	**Avg Δ (%)**
N1	6.3	6.7	6	6.3	5.2	5.5	**0.3 (15%)**
N2	4.6	5.2	4.7	5.2	3.3	3.8	**0.5 (14%)**
N3	4	4.7	4.2	4.8	2.8	3.3	**0.6 (14%)**
N4	3.9	4.6	3.9	4.6	2.5	3.1	**0.7 (15%)**
N5	4.4	5.1	3.9	4.6	2.4	3	**0.7 (15%)**
N6	5	6.1	4.4	5.5	2.8	3.7	**1.0 (26%)**
N7	6	7	5.2	6.1	3.6	4.4	**0.9 (29%)**
S1	5.2	5.8	5.8	6.2	4.8	5.3	**0.5 (18%)**
S2	5.8	6.3	5.5	6	4	4.4	**0.5 (16%)**
S3	4.9	5.6	4.9	5.5	3.4	3.9	**0.6 (17%)**

	Speech in Loud Noise (0 dB SNR)						
	Dishwasher		**Car**		**Babble**		
	NAL.	Prop.	NAL.	Prop.	NAL	Prop.	**Avg Δ (%)**
N1	-0.8	-0.7	-1.2	-1.1	-1.8	-1.7	**0.1 (8%)**
N2	-1.9	-1.6	-1.7	-1.6	-2.8	-2.6	**0.2 (9%)**
N3	-2.3	-2	-2	-1.8	-3.1	-2.9	**0.2 (9%)**
N4	-2.3	-2	-2.1	-1.9	-3.2	-3	**0.2 (9%)**
N5	-2.1	-1.7	-2.4	-2	-3.6	-3.3	**0.4 (14%)**
N6	-2	-1.1	-2.2	-1.5	-3.5	-3	**0.7 (27%)**
N7	-1.3	-0.5	-2.1	-1.3	-3.4	-2.9	**0.7 (31%)**
S1	-1.4	-1.2	-1.1	-1	-1.8	-1.7	**0.1 (9%)**
S2	-1	-0.8	-1.1	-1	-2.4	-2.3	**0.1 (9%)**
S3	-1.7	-1.4	-1.7	-1.4	-2.9	-2.7	**0.3 (13%)**

gains and the proposed method. We randomly selected 20 audio recordings from three noise classes, "dishwasher", "fan", and "microwave", to represent common low level indoor ambient noise. We also added white Gaussian noise to the audio recordings to simulate internal circuit noise of the hearing aid. Although internal noise levels vary among hearing aids, we chose 20 dB SPL as a value representative of typical hearing aids [25, 26]. The combined signals were scaled to 45 dB SPL, and average output levels were recorded for the ten standard hearing loss profiles, as well as the average difference between the NAL-NL2 output and the proposed output. Results, depicted in Table 2, show that the proposed method provides up to 4 dB less gain (roughly two turns of a volume knob on a typical sound system) to low level noise signals due to the expansion properties of the proposed amplification rule.

4.3. Speech and noise in various environments

In the second experiment, noise was mixed with speech at various input SNR levels, as designated in Table 1. The mixed signal was then passed through the hearing aid. Output SNR levels were observed for each combination of listening scenario, hearing loss pattern, and noise type.

The speech data consisted of twenty speech passages from twenty different speakers, which were mixed with twenty randomly selected noise recordings from each category, resulting in twenty noisy speech passages. Output SNR levels in dB were averaged over the twenty noisy speech passages. The average difference between the NAL-NL2 output SNR and the proposed method SNR (Δ) is obtained for each scenario and hearing loss pattern. The percentage difference ($\%$) is calculated as the ratio of the difference (Δ) to the average SNR drop yielded by the NAL-NL2 prescription gains. The results of the experiment are summarized in Table 3.

The reduction of output SNR seen in Table 3 for both NAL-NL2 and the proposed system is consistent with findings in [6, 7, 8]. Results show that the proposed amplification rule offers up to 1 dB SNR improvement for quiet and moderately noisy situations, and up to 0.7 dB in loud noise situations. For comparison, the SNR drop at moderate noise levels is in the range of 2-5 dB, and 1-4 dB for environments with loud noise. Thus, the improvement offered by the proposed method has an impact of up to 30%, as seen in Table 3. It is also noteworthy that the greatest improvements in SNR are seen at more severe hearing loss patterns where more intervention and higher signal quality are needed, likely due to the higher compression ratios prescribed at these hearing loss levels. Overall, this experiment shows that the consolidated WDRC curve provides equivalent or better output SNR than NAL-NL2 with simplified parameterization.

5. DISCUSSION

While it has been proven that compression offers benefits for speech intelligibility, many hearing aid users are dissatisfied with the sound quality of their hearing aids [2]. The increasing flexibility of modern hearing aids helps to improve user satisfaction, but introduces the challenge of properly setting the growing number of adjustable parameters [3]. Moreover, user satisfaction is largely influenced by preference. For example, users may prefer settings that maximize music appreciation over settings that maximize speech intelligibility [27].

Recent advances in technology have enabled the introduction of self-fitting hearing aids [28], which offer a solution for the problem of preference and satisfaction in hearing aid fitting. Self-fitted and more recent over-the-counter hearing aids are already seeing growing acceptance within the audiology community [29, 30]. However, simple and intuitive strategies for self-fitting are still being developed, and prominent over-the-counter hearing aids, such as [30, 31], do not offer full customization of subband gains and compression ratios. We believe that the decoupled two-step compression rule presented in this paper may be applicable to highly-customizable hearing aids for users seeking greater control over their hearing aids.

6. CONCLUSION

In this paper, we presented an alternative amplification input-output rule for Wide Dynamic Range Compression (WDRC) in hearing aids. The consolidated input-output curve is an adjustable function with logarithm-like behavior, governed by two parameters – gain and compression, where gain applies constant amplification to a signal, while compression adjusts the curvature of the function for reducing the dynamic range of sounds in the region of interest. The proposed method redefines the way a conventional compression curve is specified, from a piece-wise function with four or more parameters, to a smooth function with only two parameters. The proposed method also revisits dynamic range expansion, wherein gain decreases with signal level, to reduce the amplification of low-level ambient and circuit noise. The alternative amplification rule was evaluated using noise and speech-in-noise signals and compared to conventional WDRC. Experimental results have shown that the expansion properties of the consolidated curve reduce amplification of low-level ambient noise by up to 4 dB, and improve signal-to-noise ratio by up to 1 dB (25%) in moderate noise conditions. The consolidated WDRC amplification curve significantly simplifies the programming of a hearing aid, which is highly applicable in self-fitted and over-the-counter hearing aids.

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