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A Real Time Hearing Loss Simulator

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Summary

Several hearing loss simulators (*HLS*) have been developed to demonstrate the effects of hearing loss on auditory perception to normal hearing (NH) listeners, and to facilitate prediction of the perception of sound products by hearing impaired customers. This paper describes a real-time *HLS* based on an inverse, compressive GammaChirp (GC) filterbank, and how it was used to temporarily handicap NH listeners participating in a traditional notched-noise (NN) masking experiment (e.g. [1]) with a 2-kHz signal frequency. Sets of NN thresholds were obtain with a wide range of symmetric and asymmetric notches at two noise spectrum levels while participants listened to the sounds presented both with and without the *HLS*. The NN data were used to derive auditory filter shapes and input/output (IO) functions, which demonstrate that the *HLS* can simulate the elevation of pure tone threshold and the flattening of the input/output function commonly observed in sensory-neural hearing loss.

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1. Introduction

Simulation of sensory processing disorders provides a powerful tool for investigating hearing itself and hearing loss. In the past, there have been two main approaches: The first was equivalent-threshold masking, designed to simulate the reduced performance of hearing impaired listeners in one task or another, without regard for the quality of the perceived sound (see Lum and Braida for a review [2]). For example, to simulate specific audiometric losses, they often simply mixed a loud broadband band noise with the shape of the audiogram to the signal producing a totally different experience from what the impaired listener heard. In the second approach, they made an attempt to simulate the actual perception of the hearing impaired listener. In the simplest case, they reduced the level of the sound with a linear FFT-filter having a frequency response close to that of the target audiogram. This was used to demonstrate the consequences of sensory hearing impairment to the general public. But this linear attenuation of the signal is a poor imitation of what happens when someone loses the active process of the peripheral auditory system and its associated gain. More recently, some integrated models of hearing impairment have been proposed [3, 4], which were intended to simulate all aspects of moderate, sensory-neural hearing impairment, including what the impaired person hears.

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2. Model of hearing impairment

The principle of the current hearing loss simulator is essentially the same as that of the simulator developed by Irino and colleagues, referred to as an inverse, dynamic compressive Gammachirp (dcGC) auditory filter [4, 5, 6]. It was used by Matsui et al [7] to simulate the effect of hearing loss in a syllable perception task. However, they did not derive either the auditory filter or the input/output function of the cochlea using the simulator. As the name suggests, the inverse dcGC simulator was designed to cancel the natural compression of the normal hearing listener. In the current dcGC model, cochlear compression is simulated in three stages: 1) The signal is filtered into 32 bands using a bank of passive GammaChirp (pGC) filters. 2) The level at the output of each pGC filter is estimated. 3) The level is used to control the center frequency of a high-pass asymmetry function (HP-AF) that represents the active mechanism in that filter band. The center frequency of the HP-AF decreases as the output level of the pGC increases, reducing filter gain and increasing filter bandwidth in the process. Thus, as in the cochlea, the gain is maximal at low levels and minimal at high levels, and the system provides fast acting compression over a large dynamic range, separately in each dcGC band.

To cancel the natural compression of the dcGC filterbank, the hearing loss simulator *HLS* applies a second ver-

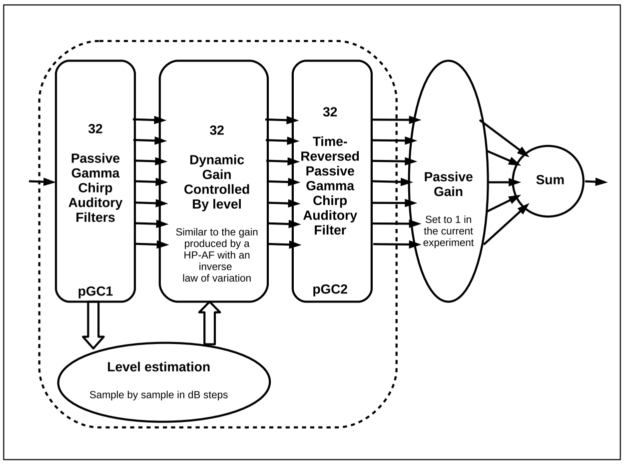

Figure 1. *HLS* signal processing: 1) pGG filters; 2) channel-by-channel level estimation; 3) calculation of HP-AF level-dependent filter coefficients and application of gain in each band; 4) time reversed pGC to cancel the delay group of each band; 5) passive gain to add a passive hearing loss (not used here); 6) sum all bands to re-synthesize.

sion of the active mechanism in reverse (see figure 1), that is, the center frequency of the second HP-AF increases as the level out of the pGC increases. In this way, the simulator acts as an inverse compressor in each frequency band, in a way that should cancel the natural compression of a normal listener.

The processing is done with a mix of python and Open Computing Language (opencl). All filter coefficients are designed with a cascade of biquad filters. The HP-AF coefficients are computed in advance for all levels and the resulting gains stored in a lookup table. All steps are computed sample by sample in each band at 44.1 kHz; the use of a Graphics Processing Unit (GPU) and opencl make it possible to process the 64 bands associated with binaural audio streams in real time. This means the hardware version of the simulator can be inserted in any audio system to simulate a sensory-neural hearing impairment. The *HLS* software can be downloaded as an open source project[1].

3. Notched-Noise Experiment

A form of GC *HLS* has previously been used to simulate the performance of a group of hearing impaired (HI) listeners on a speech-in-noise task [8]. The average audiogram of the HI group was used to fit the HLS for the normal hearing listeners. It showed the presence of a moderate hearing loss that, in turn, explained their speech intelligibility deficit. However, it was not clear whether the deficit was entirely attributable to their hearing losses or whether it was at least partially due to a more general deterioration of the signal. To resolve the ambiguity and validate the current GC *HLS*, we designed a NN experiment, centered at 2 kHz, to measure the effect of the *HLS* on absolute threshold, auditory filter shape, and the IO function of a group of normal hearing (NH) individuals, making a direct comparison with and without the *HLS*. A detailed description of the NN experiment and the derivation of auditory filter shape with a GC filter model is presented in Patterson et al (2003) [9].

3.1. Methods

Six young, normal hearing listeners were tested in their best ear, having given informed consent prior to the start of the experiment.

Two sets of NN thresholds were collected, one without the *HLS* (referred to as the *ByPass* condition) and one that included the *HLS* (referred to as the *HLS* condition). In the latter condition, the system was set to simulate a complete loss of compression in all bands. In this case, the *HLS* prediction for absolute threshold at 2 kHz increases by about 37 dB SPL.

Absolute threshold at 2 kHz was measured using a two-interval, two-alternative, forced-choice procedure with a 2-down, 1-up tracking paradigm. The intervals were 200 ms in duration, separated by 500 ms. The timing of the intervals was indicated visually on a computer display. One interval, randomly selected, contained a 200 ms sinusoid. The task of the listener was to indicate the interval that had this signal by a button press. The initial level of the tone was 40 dB SPL in the *ByPass* conditions and 77 dB SPL in the *HLS* conditions. The initial step size was 8 dB. It was reduced to 4 dB after two reversals, and to its final level of 2 dB after 2 more reversals. Threshold measurement was terminated after 16 reversals. Threshold was taken to be the average of the last 12 reversals. The conditions were presented in random order.

The same experimental procedure was used to estimate signal threshold in the two NN conditions – with, and without, the *HLS*. The only difference was that the 200-ms notch noise was present in both intervals of each trial. The spectrum level of the NN was 25 or 45 dB SPL in the *ByPass* condition and 45 or 60 dB SPL in the *HLS* condition. The initial level of the tone was set to 30 dB above the spectrum level of the NN in all conditions (*i.e. 55 dB, 75 dB or 90 dB*). The widths of the lower and upper noise bands were fixed at 400 Hz. The notch noise was generated by filtering a white noise with a 16th order butterworth bandpass filter to establish the extremities of the NN. The notch was then added using a 16th-order Butterworth, band-reject filter. Depending on the condition, the notch was positioned either symmetrically or asymmetrically about the signal frequency, 2 kHz. Nine or ten symmetrical notches were used for each noise level. In addition, there were six NN conditions where the upper band was shifted up by an extra 0.2, and six conditions where the lower band was shifted down by an extra 0.2. Based on a preliminary experiment, the widths of the notches have been independently chosen for each condition in the range 0-0.4 to try to optimize the fitting. A low-pass noise with cutoff 0.2 kHz and a spectrum level 20 dB below that of the NN was included to mask any distortion components. The conditions were presented once in random order. They

[1] https://github.com/samuelgarcia/HearingLossSimulator

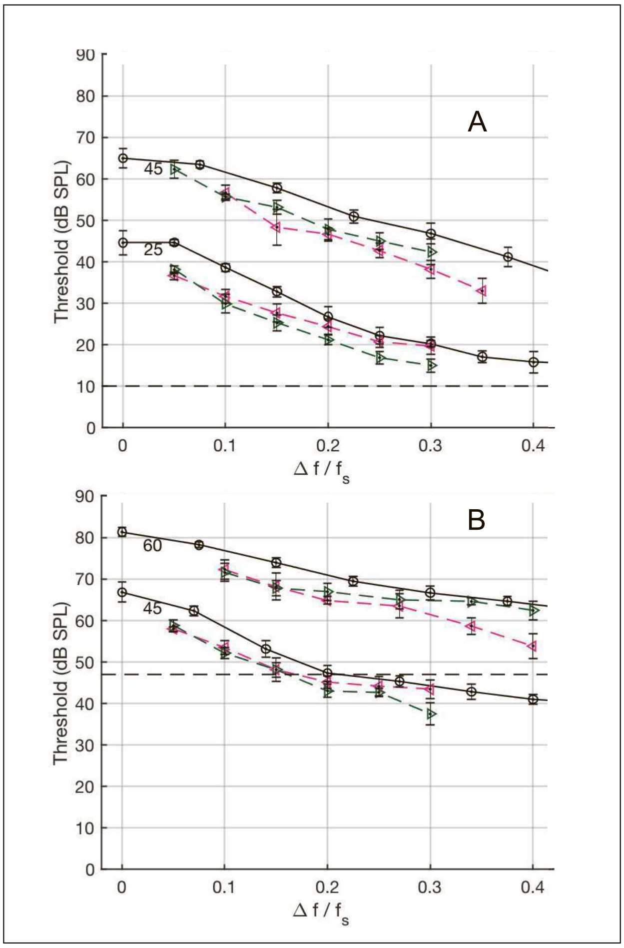

Figure 2. (Colour online) Notch noise data for the *ByPass* condition (A) and the *HLS* condition (B). The black, green and magenta symbols and lines show the data with a symmetric, right-shifted and left-shifted notch noise respectively. The black horizontal dashed lines show absolute threshold in the two conditions. The bars indicate the standard error across all 6 listeners.

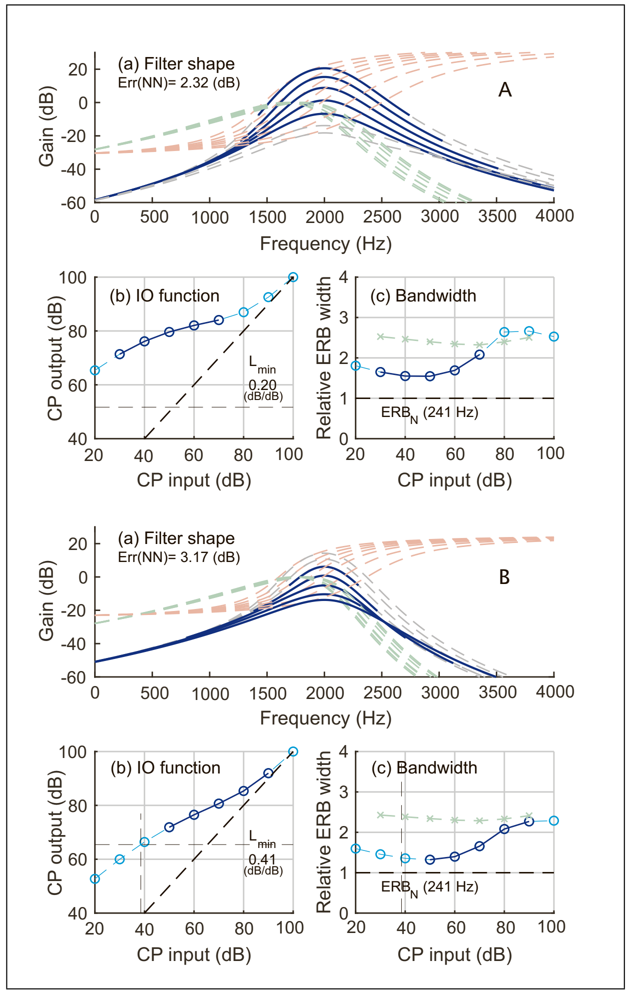

Figure 3. (Colour online) Auditory filters, input/output functions, and bandwidth functions derived from the averaged data for the *ByPass* (A) and *HLS* (B) conditions of the experiment. The upper parts of panels A and B show the estimated dcGC auditory filters (blue lines), the pCG (green lines) and the HP-AF (pink lines) for 5 input levels in the same range than the data. The lower parts show the estimated input/output functions (variation of gain at peak frequency) and bandwidth of the dcGC auditory filters function of the input level in dB (CP input). In the bandwidth pannels, A(c) and B(c), the blue lines show the estimated bandwidths of the dcGC filters and the green lines the estimated bandwidths of the pCG filters.

were grouped into three blocks (A, B and C) to provide for breaks in the testing; each block contained about the same number of notch widths and levels. Overall, 21 or 22 notch conditions were tested at each of 2 noise levels with, and without, the *HLS*.

The stimuli were passed through the numerical optical output of an RME sound card. This output was then connected to the numerical optical input of the same RME sound card, and this was the input to the *HLS*. The output of the simulator was presented monaurally to the best ear of the listener through a Sennheiser HD250, linear II headphone. The stimuli were calibrated using a class A sound level meter (Larson Davis 824) connected to an artificial ear (Larson Davis AEC101). The listeners sat in a double-walled sound booth. The experimental paradigm was formally approved by a national ethics committee (CPP Léon Bérard).

3.2. Results

The average threshold data for the *ByPass* and *HLS* parts of the experiment are plotted, as a function of notch width, in the upper and lower panels of Figure 2, respectively. The upper and lower threshold curves in the *HLS* condition have very similar shapes to the upper and lower threshold curves in the *ByPass* condition, indicating that the effect of the *HLS* is basically what it should be – a sophisticated, fast acting sound attenuator. Relative to overall level, widening the notch produces a very similar effect on threshold after the intervention of the *HLS*, and this is true for the asymmetric notches (green and magenta symbols) as well as the symmetric notches (black symbols). The main difference between the two patterns of threshold curves is that the range of thresholds obtained with the *HLS* is somewhat compressed relative to the pattern in the *ByPass* condition.

The average value of absolute threshold is shown by the black, horizontal dashed line in each panel; the value was 10.0 dB SPL (std=3.50) in the *ByPass* condition and 47.0 dB SPL (std=1.24) in the *HLS* condition. The difference, 37 dB, is exactly the change in absolute threshold predicted by setting the degree of compression to zero in the *HLS*. Note, however, that whereas absolute threshold is about 5 dB below the lowest NN threshold in the *ByPass* condition, it is a little above the lowest NN threshold in the *HLS* condition. We return to the differences between the *ByPass* and *HLS* threshold values in the Discussion.

In order to derive the auditory filters, the notch-noise data have been fitted using the same P0 power spectrum model of masking as described previously in this issue [10] and earlier [11, 12]. In each condition, the minimization of equation 4 [10]:

$$c_{gc}^{(P_0)} = \underset{c_{gc}}{\operatorname{argmin}} \left\{ \frac{1}{N} \sum_{i=1}^{N} \left(P_{s_i} - \hat{P}_{S_i}^{(P_0)} \right)^2 \right\}$$

provides the full set of parameters of the dcGC model which best predicts the data. Using these parameters, figure 3, presents 5 auditory filters, at 5 input levels (every 10 dB in the range of the data), derived with the dcGC model in the *ByPass* and *HLS* parts of the experiment in the upper parts of panels A and B, respectively. The blue lines show that the auditory filter provides gain in the passband region in both the *ByPass* and *HLS* conditions. The errors between the threshold values predicted by this dcGC filter model show that the model provides an accurate description of the *ByPass* data with an error equal to 2.32 dB as indicated in Figure 3, and also an reasonable description of the *HLS* data with an rms error equal to 3.17 dB. In this condition, the prediction of the minimum threshold is about 5 dB below absolute threshold, and the model predictions are a little above the corresponding data at the wider notch widths. This discrepancy is reflected in the higher rms error in this condition than in the *ByPass* condition.

Note that the design maximizes the number of different NN conditions in the experiment, in preference to replicating a smaller number of conditions, and so the error in the GC fits includes the intra-individual variability (i.e. the error in individual threshold estimation).

The input/output (IO) functions and the bandwidth (BW) functions provided by the dcGC model are plotted, as a function of stimulus level, below the corresponding filter shape plots in Figure 3. The blue portions of the IO and BW curves show estimates from roughly the same range of levels as the measured thresholds; the cyan sections show extrapolations to lower and higher levels.

4. Discussion

The IO function for the *ByPass* condition is strongly compressive, as expected, with a slope of 0.2 dB/dB for input levels around 60 dB SPL. The IO function for the *HLS* condition is much less compressive; the minimum is 0.41 dB/dB. The form is consistent with the loss of compression that would be expected from a HI listener with a 37-dB hearing loss.

The BW values for filters in the level range of the threshold data (the blue portion of the BW function) are 1.6-2.0 times the normal ERB (ERB_N) value [13] in the ByPass condition, and 1.3–2.2 times the ERB_N value in the HLS condition. Part of the difference arises from the fact that the ERB_N BW values were derived with a roex filter-shape which has been shown to underestimate the actual width of the tip of the auditory filter (see [13], subsection IV.B).

It remains the case, however, that the average BW value in the *HLS* condition is somewhat smaller than that in the *ByPass* condition. This is because the HLS simulates the loss of gain in the HI by reducing the level of the stimuli (signal + maskers) presented to the NH listeners in HLS condition. That is, the sounds are actually being presented to these NH listeners at a much lower level than the CP input axis would suggest. In the ByPass condition, the stimuli are being presented at the stated CP input level. The BW of the NH listeners is greater at higher levels, so the BW values are greater in the ByPass condition than in the HLS condition. This does, however, mean that the HLS is limited to simulating the loss of gain in the HI; it does not simulate the increase in BW associated with the need to present stimuli at higher levels for the HI.

5. Conclusions

The *HLS* was observed to raise absolute threshold substantially and reduce compression, making the auditory system appear more linear. These changes are qualitatively consistent with the presence of a flat hearing loss of around 40 dB. A similar simulator [8] has been shown to produce a reduction of intelligibility for speech presented in noise, similar to that observed with HI listeners. The results of the current experiment allow us to conclude, more generally, that the *HLS* illustrates the joint effects of reduced audibility and reduced compression commonly encountered in HI listeners.

Acknowledgements

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