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Modeling alarm detection in noise for normal and hearing-impaired listeners: the effect of elevated thresholds and enlarged auditory filters

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Abstract—A model was developed to assess how elevated absolute thresholds and enlarged auditory filters can impede the ability to detect alarms in a noisy background, such alarms being of paramount importance to ensure the safety of workers. Based on previously measured masked thresholds of 80 listeners in five groups (normal hearing to strongly impaired), the model was derived from signal detection theory (SDT) applied to Glasberg and Moore’s excitation pattern model. The model can describe the influence of absolute thresholds and enlarged auditory filters together or separately on the detection ability for normal hearing and hearing-impaired listeners with various hearing profiles. Furthermore, it suggests that enlarged auditory filters alone can explain all of the impairment in this specific alarm detection task. Finally, the possibility of further development of the model into an alarm detection model is discussed.

Index Terms—Hearing impaired, Detection in noise, Warning signal, Model

I. INTRODUCTION

IN many workplaces, acoustic warning signals are used to rapidly alert workers of a hazardous situation. In practice, a number of factors may affect the efficiency of these warning signals by reducing their audibility [1], [2]. These factors can be as various as worker hearing status, the presence of background noise or use of hearing protection devices (HPDs). An impaired ability to detect a warning signal could lead to failure to react to these signals, thereby causing a possibly dangerous situation.

The ability to predict a masked threshold (i.e., the threshold of a target signal embedded in a background noise) would be of prime interest in relation to warning detection and accident prevention. Such models have already been put forward, e.g., the Detectsound model implemented by Zheng et al. [3] or Glasberg and Moore’s model [4]. The former proposes optimal warning signal levels, in 1/3 octave bands, based on the knowledge of background noise, hearing status and HPD attenuation. Glasberg and Moore’s model [5] involves computing excitation patterns to estimate loudness in several frequency bands.

The main objective of the work described in this article is to propose a model to investigate the role of hearing status

in alarm detection ability. This model is mainly derived from signal detection theory (SDT) [6], [7] applied to Glasberg and Moore’s [5] excitation pattern model and can be applied to hearing-impaired listeners.

For its application to various hearing statuses, the model must be capable of considering absolute hearing thresholds and auditory filter widths since these two parameters may strongly influence signal detectability in noise [3]. Absolute thresholds set the limit below which signals cannot be perceived by listeners. In practice, while the detectability of a warning signal in noise is rarely impeded by absolute thresholds in normal-hearing listeners, hearing-impaired listeners may exhibit absolute thresholds so high that they do impede their detection ability. Furthermore, it may be difficult to assess to what extent elevated absolute thresholds alone affect detection because various suprathreshold effects can still affect detection when sound levels are higher than absolute hearing thresholds. The effect is more apparent when studying detectability with HPDs. Lazarus [8] and Arz et al. [2] have shown that detection may be impeded when hearing-impaired listeners are wearing HPDs. In fact, because HPD attenuations add up to their absolute thresholds, this may result in a significant increase of their detection thresholds. Auditory filters also play an important role since they are at the basis of frequency selectivity. Fletcher [9] has shown that, when the ear listens to a tone, a filter centered on the target tone excludes the noise outside the filter bandwidth. The broader the bandwidth, the more noise is processed by the ear and the larger is the masking of the target tone. Glasberg and Moore [10] have shown that the filter bandwidth increases with increasing sound level, thereby degrading frequency selectivity and increasing masking effects at higher levels. The auditory filters of hearing-impaired listeners tend to be broader than those of normal-hearing listeners when the absolute thresholds exceed approximately 30 dB [11], [12], [13]. Beyond 30 dB, the filter width tends to increase with increasing absolute thresholds, resulting in stronger and more troublesome masking effects.

This article first describes the general framework of the model. Second, a previously conducted experiment, involving the measurement of masked thresholds of warning sounds, is presented. Third, the model is further detailed by focusing on simulating hearing impairment. Finally, the results of the

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previous experiment are modeled and analyzed to assess how the internal noise described by SDT varies with the hearing status and how elevated absolute thresholds and/or enlarged auditory filters influence detection in noise.

II. GENERAL PRINCIPLE OF THE MODEL

SDT states that the perception of a signal can be represented as a distribution of the internal response upon a perceptive axis [6], [7]. When detecting a signal embedded in a masking noise, the two distributions corresponding to the noise alone (N) and the noise + signal (NS) are centered at different positions. Each distribution is defined by its mean value μ (μ_N for noise, μ_{NS} for noise + signal) and by its variance σ (σ_N for noise, σ_{NS} for noise + signal). The assumptions usually made in SDT are that the distributions are normal and that their variances are equal, thus $\sigma_N = \sigma_{NS} = \sigma$ [7]. The variance σ represents the “internal noise” which is the cumulative effect of any noise that impacts the detection ability of a listener. It can relate to a neuronal noise, a hearing impairment and/or a cognitive noise. A high internal noise (i.e., a large value of σ) leads to the superposition of the two distributions and makes the detection task harder. In fact, when both stimuli (N and NS) have similar distributions, it will be difficult to discriminate the stimulus which contains the target signal from the one with the noise alone. In the SDT framework, from the internal noise (σ) and the difference between the two distributions’ mean values ($\Delta\mu = \mu_{NS} - \mu_N$), a detectability index d' can be computed as follows:

$$d' = \frac{\Delta\mu}{\sigma}. \quad (1)$$

This detectability index is related to the degree of distribution superposition [6], [7] and thus correlated with the difficulty of the detection task (the smaller d' , the harder the detection task); it is therefore related to the detection percentage targeted in the detection task.

The value of $\Delta\mu$ is directly linked to the physical characteristics of the sounds used in the detection task and, at threshold, $\Delta\mu$ allows us to obtain the target signal level in decibels. We can therefore estimate the internal noise σ , using $\sigma = \Delta\mu/d'$, based on the targeted detection percentage in a detection task (i.e., linked to the value d') and the value of $\Delta\mu$ (related to the measured threshold in decibels). Conversely, if we know the value of σ , we should be able to calculate the value of $\Delta\mu$ and the corresponding masking threshold of the target signal.

III. MATERIALS FOR THE ESTIMATION OF σ

The model computes the values of internal noise σ from the masked thresholds measured in a previous detection experiment presented by Arz et al. [14], which is only reminded of here.

A. Participants

Eighty subjects aged from 18 to 81 years (M 51.6 years; SD 15.3 years) with hearing status ranging from normal to highly impaired took part in a detection experiment. Under French law, this study does not fall within the scope of research

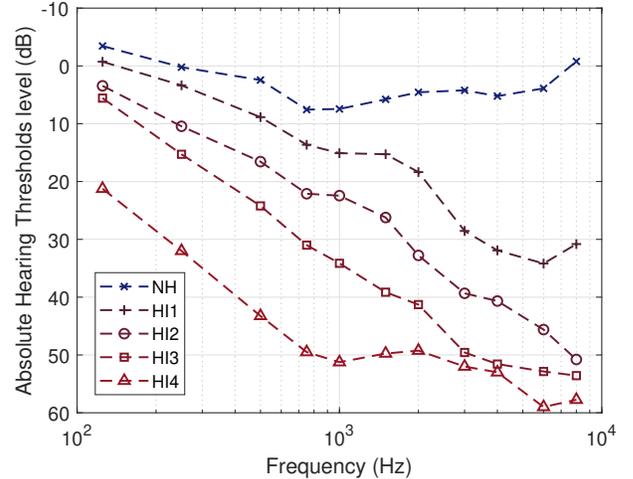


Fig. 1. Average audiograms for the five hearing groups that participated in Arz et al.’s study [14]. Note: NH = normal hearing group; HI = hearing-impaired group.

involving human beings for the purpose of improving biomedical knowledge. There was no ethical committee in place at INRS at the time the study was conducted. Nevertheless, it has been approved by an independent scientific committee and every precaution has been taken to ensure the ethical conduct of the research. The subjects gave their free, written and informed consent to participate in this study and the data have been collected and stored in accordance with the requirements of the European General Data Protection Regulation. The subjects’ absolute thresholds were measured using pure-tone audiometry at the 11 standard audiometric frequencies from 125 to 8000 Hz (see Figure 1). Five hearing groups were considered. Subjects with absolute hearing thresholds below 20 dB at every frequency formed the normal hearing group, denoted NH (18 subjects). The remaining subjects were distributed into four groups according to their average absolute hearing threshold at 500, 1000 and 2000 Hz in their best ear (pure tone average [PTA]). The following hearing-impaired groups were considered:

- group HI1: $10 < \text{PTA} \leq 20$ dB, 22 subjects
- group HI2: $20 < \text{PTA} \leq 30$ dB, 18 subjects
- group HI3: $30 < \text{PTA} \leq 40$ dB, 14 subjects
- group HI4: $\text{PTA} > 40$ dB, 8 subjects

Figure 1 displays the best ear mean absolute thresholds for each hearing group.

Additionally, the shapes of the auditory filters centered at 500, 1000, 2000 and 3000 Hz were evaluated from a notched-noise experiment [15], [16], [17] in which the noise level was fixed and the level of the pure tone varied. Six different notches (four symmetric and two asymmetric) were used. Their lower and upper limits, expressed as a fraction of the central frequency of the filter, were: [0–0] (i.e., no notch), [0.2–0.2], [0.3–0.3] and [0.5,0.5] for the four symmetric notches; and [0.3–0.5] and [0.5–0.3] for the two asymmetric notches. The noise was a white noise with a default level of 40 dB/Hz.

Hearing group	Filter center frequency			
	500 Hz	1000 Hz	2000 Hz	3000 Hz
NH	1.13	1.10	1.20	1.05
HI1	1.14	1.25	1.32	2.09
HI2	1.25	1.23	2.59	2.05
HI3	1.53	1.70	2.76	2.84
HI4	2.06	2.84	3.73	2.77

TABLE I
MEASURED AUDITORY FILTER ENLARGEMENT FACTORS FOR EACH HEARING GROUP. NOTE: NH = NORMAL HEARING GROUP; HI = HEARING-IMPAIRED GROUP.

However, for some hearing-impaired listeners, this level had to be raised to 50 dB/Hz (or even to 60 dB/Hz in some rare cases), the criterion for this adjustment being that the threshold measured for the [0.5–0.5] notch should be at least 4 dB higher than the absolute threshold at this frequency. The threshold measurements were performed monaurally using TDH 39 headphones (Telephonics, USA). The auditory filters were assumed to have the form of an asymmetric rounded exponential ($\text{roexp}(p_u, p_l, r)$) as described by Patterson et al. [18] and Glasberg and Moore [12]. The parameters (p_u , p_l and r) of the filter were obtained using an iterative procedure that minimizes the mean-squared error between predicted and measured thresholds.

The equivalent rectangular bandwidths (ERBs) of the filters were then obtained from these parameters. Table I presents the mean enlargement factors for the five hearing groups at the four frequencies, defined for each frequency as the ratio between the measured ERB and the ERB of normal-hearing listeners (ERB_N) provided by Glasberg and Moore [5]. For each center frequency and each listener, the ear with the best frequency selectivity was considered.

B. Detection task

The masked thresholds of seven warning signals commonly used by the French national railroad company (SNCF) were measured (four of them were measured in one masking noise, and the remaining three signals were measured in another masking noise; see Arz et al. [14]). The two masking noises dominated at low frequencies ($f < 500$ Hz) and most of their energy was below 3 kHz. The seven alarms were all harmonic sounds except for one signal which was composed of two inharmonic pure tones. The alarms exhibited different frequency contents: some alarms had mainly low frequency contents ($f < 1500$ Hz) with only a few harmonics, and other alarms had mainly a high frequency content with a lot of harmonics over the frequency spectrum. Out of the seven alarms, five were stationary sounds and the two remaining consisted of two alternating tones. This set of alarms is therefore diverse, mostly stationary and illustrates a wide variety of situations.

The alarm stimuli were presented through one speaker (KH 120A; Neumann, Germany) to the listeners and, for each warning signal, the masked threshold was estimated in the presence of its respective masking noise using an adaptive, two-interval forced-choice (2IFC) procedure with

a two-down–one-up adaptive rule. The masking noises were generated through two LSR6332 speakers (JBL, USA). This procedure led to 70.7% detection [19] and a detectability index d' of 0.78 [20]. The noise levels were fixed at 86 dB(A) while the warning signals started at 86 dB(A) and varied according to the listeners' responses. The initial 5-dB step size was reduced to 3 dB after the first three reversals and finally to 1 dB after two further reversals. The thresholds were calculated as the average level of the last four reversals.

The masked thresholds measured when no HPD is worn (see Arz et al. [14] for more details) were used in the present article to estimate the values of internal noise σ (see Section IV-A).

IV. MODEL

A. Estimating σ

The first step of the model is to compute the excitation pattern of the noise alone ($EP(N)$) and the excitation pattern of the noise mixed with the signal ($EP(NS)$) at the threshold measured in the 2IFC procedure by Arz et al. [14]. Based on Glasberg and Moore [5], this computation involves 321 overlapping auditory filters (roexp filters [18]) for normal-hearing listeners. For hearing-impaired listeners, the excitation patterns can be computed with broader filters depending on the measured ERBs (see Section IV-B).

The second step of the model is to estimate, from the previously computed patterns, the value of $\Delta\mu$ that is linked to the sound levels of the noise and the signal. However, having a single value for $\Delta\mu$ as expected from Equation 1 cannot account for the auditory system's capacity to integrate information across a large frequency spectrum. Instead, it is necessary to use the multi-band energy detector model expression [21]. The detectability index is then expressed in terms of the detection index in each independent frequency band (d'_i) by the following equation:

$$d' = \sqrt{\sum_{i=1}^{N_b} d_i'^2} = \sqrt{\sum_{i=1}^{N_b} \frac{\Delta\mu_i^2}{\sigma_i^2}} \quad (2)$$

where N_b = number of independent bands (33 bands for normal-hearing listeners or fewer bands for hearing-impaired listeners). The bands are defined on the ERB_N number scale [5], [22] (referred to simply as the ‘‘E-scale’’ in the rest of this article). In each independent band, $\Delta\mu_i$ is the difference between the excitation pattern of NS and the excitation pattern of N summed over this band: $\Delta\mu_i = EP_i(NS) - EP_i(N)$.

As the variation of σ across frequency bands is unclear, an additional experiment involving detection of pure tones in noise with normal hearing listeners has been conducted. From these results, a weight function w_i was derived to account for variations in σ . All the details of the experiment and the weight function are described in Supplemental data. The expression for σ is then as follows:

$$\sigma = \frac{\sqrt{\sum_{i=1}^{N_b} \Delta\mu_i^2 w_i}}{d'} \quad (3)$$

Among the seven alarm signals used, two were non-stationary and consisted of two alternating tones. Therefore, the ability to detect these two signals in noise may be time-dependent (i.e., the detection of these signals might be due to the detection of only one of the two alternating tones). To take these aspects into account, σ is computed on several portions of the signal, all lasting 150 ms and being spaced by a 10-ms time step, and the maximum value is kept. From Equation 3, the final expression of σ then becomes:

$$\sigma = \frac{\max_{j \in [1; N_T]} \sqrt{\sum_{i=1}^{N_b} \Delta\mu_{ij}^2 w_i}}{d'} \quad (4)$$

where N_T = total number of time frames.

B. Hearing impairment simulation

For hearing-impaired listeners, the model can take into account independently both absolute threshold elevation and auditory filter enlargements. To consider the higher absolute thresholds of hearing-impaired listeners, the absolute thresholds values are first converted from hearing levels to sound pressure levels using the Standard No. NF EN ISO 389-1 [23] conversion table. The absolute threshold values are then compared with the excitation patterns levels in each independent frequency band. Three situations may then arise:

- If the absolute threshold is lower than the noise pattern level (i.e., N is audible in this band), then $\Delta\mu_i$ is equal to the difference between the two excitation patterns: $\Delta\mu_i = EP_i(NS) - EP_i(N)$.
- If the absolute threshold is higher than the NS excitation pattern (i.e., neither N nor NS are audible in this band), then $\Delta\mu_i = 0$.
- If the absolute threshold is between the two patterns (i.e., N is not audible but NS is audible in this band), then $\Delta\mu_i$ is equal to the difference between the NS excitation pattern and the absolute threshold: $\Delta\mu_i = EP_i(NS) - \text{absolute threshold}$.

The original excitation pattern model [5] was also modified to take into account the auditory filter enlargement that occurs in hearing-impaired listeners [11], [12], [13]. The purpose is to account for the wider auditory filters of hearing-impaired listeners which will modify the shape of the excitation pattern compared to the patterns obtained for normal-hearing listeners. The consequence of these auditory filter enlargements is also to reduce the number of independent bands so that σ is

computed with fewer independent bands than the 33 usual bands used for normal-hearing listeners. To deduce the number of independent bands from the four measured enlargement factors presented in Table I, the 33 bands used for normal-hearing listeners are first separated into four frequency regions on the E-scale. Each region is associated with one of the four measured enlargement factors at the corresponding center frequency (to cover the 33 normal bands, the regions associated with 500 and 3000 Hz are noticeably broader than the region associated with 1000 and 2000 Hz). The number of bands contained in each region is then divided by the associated enlargement factor. The resulting number is rounded to the nearest integer to finally determine how many independent bands can fit into a given frequency region when the enlargement factor is considered.

Once the number of independent bands fitting into a region has been determined, their characteristics (central, upper and lower frequencies) must be computed. This computation is fairly easy because each band has a unit width on the E-scale. It is therefore easy to place the new bands on the E-scale and the subsequent frequency conversion is straightforward using the reciprocal of Equation 4 in Glasberg and Moore [5]. Lastly, the width of each overlapping filter used by the excitation pattern model is determined by interpolating the bandwidths of the independent bands. Hence, wider filters are used to compute the excitation patterns and fewer but wider independent frequency bands are used to compute σ for each hearing-impaired listener.

In essence, the model can compute σ while simulating the impairment by:

- using wider, overlapping auditory filters to compute the excitation patterns (see Figure 2)
- modifying the excitation patterns values based on the absolute thresholds before computing the $\Delta\mu_i$ values
- using fewer than 33, wider than normal, independent frequency bands to compute the $\Delta\mu_i$ values (see Figure 2) – the number of bands used depends on the hearing status simulated and is smaller with stronger impairment

These choices reflect the assumption that the internal noise in SDT is due to three different components: a component related to elevated absolute thresholds, a component related to enlarged auditory filters and a third component related to all other unknown aspects (e.g., age, cognitive factors, etc.). When the hearing status is not simulated in the excitation pattern, all components of the internal noise are included in σ . When filter enlargement and/or threshold elevation are simulated in the model, the related noise components are theoretically excluded from σ . Thus, the nature of the internal noise that is computed is different depending whether or not we add any simulation of hearing impairment. When excluding those components from σ , it is expected that the internal noise σ will be reduced. It is therefore presumed that taking impairment into account for hearing-impaired groups will bring their internal noise value closer to that of the normal-hearing group.

C. Statistical data analysis

Internal noise σ was computed for four different simulation conditions (see Section V) and Bayesian repeated-measures

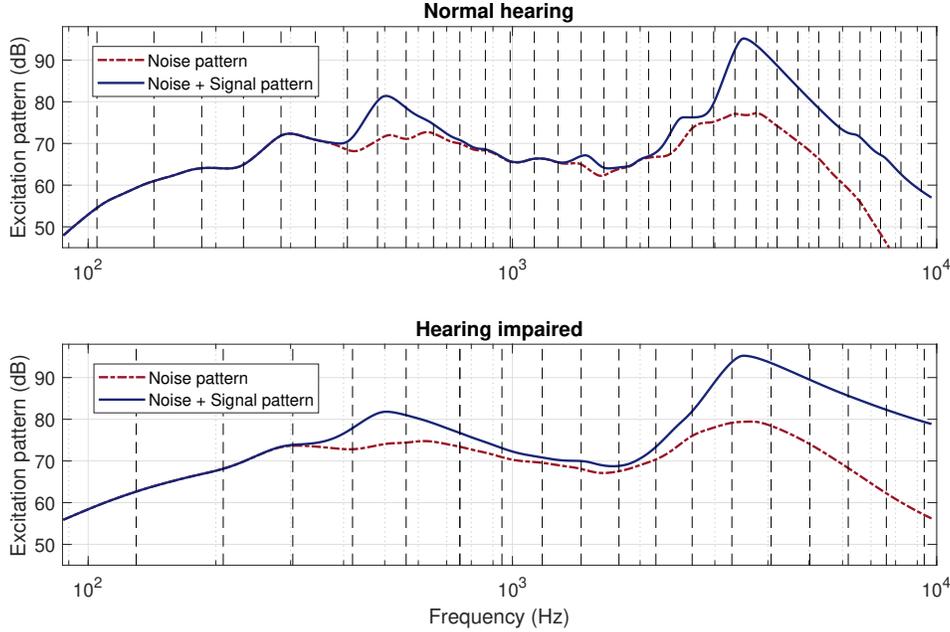


Fig. 2. Example of modified excitation patterns. Top: normal hearing patterns with independent bands corresponding to the ERB_N . Bottom: hearing-impaired patterns are wider because they were computed with broader filters, and the independent bands used to compute $\Delta\mu_i$ are fewer and broader. Note: vertical dashed lines are independent band limits. $\Delta\mu_i$ = difference between the excitation pattern of noise+signal (NS) and the excitation pattern of noise only (N) summed over the band; ERB = equivalent rectangular bandwidths; ERB_N = number scale.

analysis of variance (ANOVA) was carried out to compare σ across hearing groups with the impairment simulation condition as the repeated factor, and the hearing group as the between-subjects factor. Post-hoc comparisons (groups HI_i compared to group NH) were carried out to test factors evidenced by the Bayesian ANOVA.

Bayesian analysis does not simply reject null or alternative hypothesis, but instead compares different model candidates to describe the data and expresses how much evidence there is toward one of the models [24], [25]. The Bayes factor BF_{01} can be interpreted as how many times the null model is more or less likely to represent the data compared to the specific model tested. This test is inconclusive between 3 and 1/3.

On the one hand, a value between 1/10 and 1/3 indicates moderate evidence of the specific tested model; strong evidence between 1/100 and 1/10, and decisive evidence for a value smaller than 1/100. On the other hand, a value between 3 and 10 indicates moderate evidence of the null model; strong evidence between 10 and 100, and decisive evidence for a value higher than 100. To observe the strength of a single effect, we also report the BF_{excl} factor, which tests the evidence of the null model against a single model. All of the analyses were carried out with JASP version 0.16.0.0 [26].

V. RESULTS

To evaluate σ , and to estimate independently or in combination the effects of both elevated absolute thresholds and enlarged auditory filters, the values of σ were computed in the following four simulation conditions:

- without hearing impairment simulation (condition D)

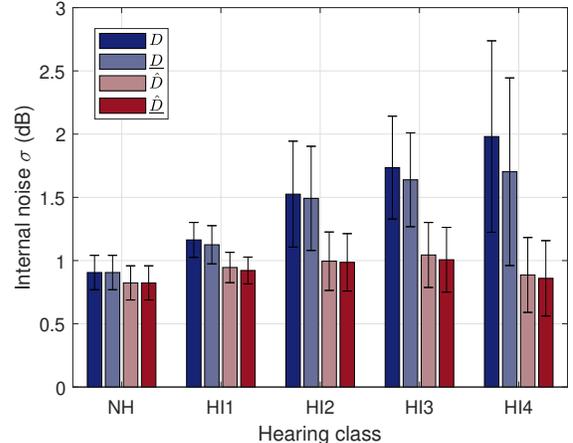


Fig. 3. Values of σ averaged across target signal for participants grouped by hearing status: normal hearing group (NH) and hearing-impaired groups (HI1, HI2, HI3 and HI4). Note: D , \hat{D} , \tilde{D} and $\hat{\tilde{D}}$ bars correspond to different hearing impairment simulations. D : no simulation; \hat{D} : simulation of elevated thresholds only; \tilde{D} : simulation of enlarged equivalent rectangular bandwidths (ERBs) only; $\hat{\tilde{D}}$: simulation of elevated thresholds and enlarged ERBs. (see text for details). Black lines are 95% confidence intervals.

- with elevated absolute threshold simulation only (condition \hat{D});
- with enlarged auditory filter simulation only (condition \tilde{D});
- with both elevated absolute threshold and enlarged auditory filter simulation (condition $\hat{\tilde{D}}$)

The internal noise σ was computed for each listener and for each of the seven warning signals by computing the excitation patterns at the threshold. When hearing impairment (elevated absolute thresholds and/or enlarged auditory filters) was simulated in the excitation patterns, the individual absolute thresholds and the individual auditory filter widths estimated in the previous experiment were used. For each listener, the values of σ were averaged across the seven alarms.

Figure 3 shows the mean values of σ for the four simulation conditions and the four hearing groups as well as the associated 95% confidence intervals (CIs).

The Bayesian repeated-measures ANOVA indicates two complex models that represent best the data, including the two factors “group” (i.e., hearing group) and “simulation” (i.e., simulation condition), and two simple models:

- simulation+group+simulation \times group: $BF_{01} = 5.421E-44$
- simulation+group: $BF_{01} = 2.996E-28$
- simulation: $BF_{01} = 1.1214E-27$
- group: $BF_{01} = 0.246$

These results show a strong significant impact of the factor simulation while the effect of the factor group, although significant, is considerably less marked. The subsequent Bayesian analysis of effect confirms these findings with $BF_{\text{excl}} = 1.219E-27$ for the factor simulation, $BF_{\text{excl}} = 0.247$ for the factor group and $BF_{\text{excl}} = 1.809E-16$ for the interactions between these two factors.

Further Bayesian ANOVA was conducted to observe the effect of hearing impairment simulations on the different hearing groups. When there is no hearing impairment simulation (condition D), the Bayesian ANOVA shows decisive evidence for the factor group ($BF_{01} = 0.006$). Post-hoc tests do not enable to conclude about the potential difference between group HI1 and group NH ($BF_{01} = 0.351$). This was expected because group HI1 has a hearing status very similar to that of group NH in terms of absolute thresholds and auditory filter enlargements (see Figure 1 and Table I). Group HI2 shows a moderate difference from group NH ($BF_{01} = 0.125$) while groups HI3 and HI4 exhibit very strong differences from group NH ($BF_{01} = 0.004$ and $BF_{01} = 0.005$, respectively).

When introducing the simulation of elevated thresholds (condition \underline{D}), the Bayesian ANOVA also shows strong evidence for the factor group, although weaker than the case without hearing impairment simulation ($BF_{01} = 0.034$). In this simulation condition, post-hoc tests are inconclusive between groups HI1 and NH and show moderate difference between groups HI2 and NH ($BF_{01} = 0.497$ and $BF_{01} = 0.157$, respectively). Group HI3 differs strongly from group NH ($BF_{01} = 0.006$) while group HI4 still differs strongly from group NH, but much less than without taking absolute thresholds into account ($BF_{01} = 0.005$ without simulation, $BF_{01} = 0.052$ when taking into account elevated thresholds), highlighting the strong impact of elevated absolute thresholds for this group.

When simulating enlarged auditory filters (condition \hat{D}), the Bayesian ANOVA reveals that there is significantly no effect of the factor group ($BF_{01} = 6.073$).

Finally, accounting for both elevated thresholds and enlarged auditory filters (condition $\hat{\underline{D}}$), the Bayesian analysis evidences that the results are similar across groups (i.e., the null hypothesis is validated; $BF_{01} = 7.312$).

Additional Bayesian ANOVA was conducted to assess the effect of impairment simulation over each group. Group NH is significantly not affected by the impairment simulation as $BF_{01} = 7.602$ shows evidence for the null model. Groups HI1 and HI2 do not enable to conclude about the simulation model, with $BF_{01} = 0.472$ and $BF_{01} = 0.358$, respectively. Post-hoc tests reveal inconclusive difference between conditions D and \underline{D} (with elevated thresholds) for group HI1.

Group HI3 is strongly affected by the impairment simulations, and the Bayes factor $BF_{01} = 0.027$ shows strong evidence for this. Post-hoc tests are inconclusive with condition D when only the elevated thresholds are accounted for ($BF_{01} = 2.684$), while conditions \hat{D} and $\hat{\underline{D}}$ differ strongly from condition D ($BF_{01} = 0.109$ and $BF_{01} = 0.078$, respectively). Simulation of enlarged auditory filters with or without elevated thresholds therefore reduces the mean value.

Group HI4 is also strongly affected by the impairment simulations ($BF_{01} = 0.038$). Post-hoc tests show that simulation conditions \hat{D} and $\hat{\underline{D}}$ differ from the case without impairment simulation ($BF_{01} = 0.143$ for \hat{D} and $BF_{01} = 0.128$ for $\hat{\underline{D}}$).

VI. DISCUSSION

The simulations of hearing impairment enable a worthwhile analysis as the estimated σ values yield meaningful information regarding the mechanisms underlying detectability. When hearing status is not considered (i.e., no hearing loss simulation, condition D), the results show that the σ values obtained are very different across the hearing groups and that the higher the absolute thresholds, the higher the internal noise. This is consistent with the idea that when internal noise includes all aspects of hearing impairment, internal noise increases with hearing impairment.

When hearing impairment is simulated in the model, there are less discrepancies in the σ values across the groups of listeners (Figure 3) and the hearing-impaired groups get closer to the normal-hearing group to the point that there are no more statistical differences between groups. CIs also get smaller when hearing impairment simulation is introduced. Testing the different hearing impairment conditions (conditions D to $\hat{\underline{D}}$) then enables to assess independently the influence of elevated absolute thresholds and the influence of enlarged auditory filters for each group of listeners.

As expected, group NH is not affected by hearing impairment simulations. Quantitatively, their σ values decrease when enlarged filters are simulated; this could have hinted that the known formulae for ERB_N [5] may have given narrower auditory filters than those measured for this group of normal-hearing listeners if the decrease in σ was not anecdotal.

Group HI1 is fairly similar to group NH and never differs from group NH whatever the impairment simulation condition, meaning that they remain comparable despite their slight hearing impairment. They benefit mainly from the addition of the enlarged auditory filters in order to obtain σ values close to

those of group NH listeners (Figure 3), so it appears that their degradation arises mainly from poorer frequency selectivity rather than from elevated thresholds.

All other groups (HI2, HI3 and HI4) differ from group NH when there is no impairment simulation. Simulation of elevated thresholds only has a small impact on the σ values obtained for these three groups. Indeed, Figure 3 shows a decrease in the σ values that becomes more pronounced with increasing hearing impairment but the analysis did not reveal that this decrease is significant. However, when simulating enlarged auditory filters (conditions \widehat{D} and \widehat{D}), none of the three groups differ anymore from group NH and they all have mean values of σ similar to the mean σ value of group NH.

Overall, without any simulation of hearing impairment, internal noise values increase continuously with increasing hearing impairment while hearing impairment simulation brings internal noise values close to those of normal hearing listeners. Moreover, the larger the hearing impairment, the stronger the benefit of the simulation of elevated thresholds and/or enlarged auditory filters. This comes from the fact that the simulations remove the components of the internal noise related to threshold elevation and/or enlarged auditory filters, thus finally leading to equal internal noise values across groups when both are simulated. In more details, the model reveals that for all groups, impeded detection ability seems to be exclusively caused by enlarged auditory filters.

It would have been expected for the groups with strongest hearing impairment (groups HI3 and HI4) that their highly elevated thresholds would play a major role in the impairment of their detection in noise ability. However, σ values were derived from masked thresholds obtained by Arz et al. [14] when no HPD is worn. The authors show that masked thresholds are similar for all of the hearing-impaired groups and the normal hearing group: the differences among the groups are small, and out of seven alarms signals, only three (respectively four) yield significant difference between group HI4 (respectively group HI3) and group NH. When wearing HPDs, however, all alarms but one give masked thresholds significantly different from group NH for both groups HI3 and HI4. Given that the elevated absolute thresholds can impair detection mainly when combined with the HPD's attenuation [2], [27], the absence of significant effect of elevated thresholds on σ (derived from open ear masked thresholds) is ultimately not surprising.

At present, although the model cannot describe all of the phenomena associated with hearing impairment, simulating hearing threshold elevation along with degraded frequency selectivity makes it possible to make internal noise comparable across all groups of listeners (the null model is supported by $BF_{01} = 7.312$). The other effects reported in the literature, such as aging and temporal aspects in signal detection, are about negligible in this study to estimate accurate σ values in the context of alarm detection in noise.

VII. PROSPECTS

The model described in this article enables an easy assessment of internal noise in relation to hearing impairment. It also offers the ability to evaluate independently the impact

of elevated absolute thresholds and of degraded frequency selectivity on the specific alarm detection task presented here. The same analysis could be done on a specific signal to assess which part of the hearing process is relevant for detection of this given signal, thereby assisting warning signal design or selection.

Additionally, using the values of σ obtained from the 80 subjects, the model could be used to predict the signal level (expressed by $\Delta\mu$) required to reach a desired detection percentage (expressed as d'); thereby providing a predictive detection tool that is able to easily account for impairment. Such a predictive model could also consider the attenuation provided by a HPD. It is known that the use of HPDs combined with hearing impairment may increase the risk of failed detection depending on the acoustic characteristics of the warning signal and the background noise [8], [27], [28]. The attenuation of such a protection device can easily be added to the model, thus providing assistance to select or design alarms and HPDs. Such a predictive model could simulate elevated thresholds or/and enlarged auditory filters, so it would enable to predict the masked thresholds of a listener knowing his hearing status with more or less accuracy (absolute thresholds and/or auditory filters or only hearing group). Such a model would be also expected to prompt better predictions if the hearing status is precisely known. The accuracy to predict detection thresholds with such a predictive model should be further tested and falls outside the scope of this article.

VIII. CONCLUSION

A model to compute the values of internal noise with respect to various degrees of hearing impairment has been developed. The model uses an approach based on SDT, allowing it to assess the separate or combined influences of enlarged auditory filters and elevated hearing thresholds on auditory detection. By applying the model to a specific signal, it can identify which of these two phenomena most impacts the signal detection. Finally, the model could be used to predict detection thresholds of warning signals for normal and hearing-impaired listeners based on the internal noise values provided in this article.

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APPENDIX SUPPLEMENTARY INFORMATION

A. Variation of σ across frequency bands

The assumption that the internal noise is constant over all frequency bands (i.e. 33 bands for normal-hearing listeners) is rather unlikely. On the contrary, we assume that the following weight function could account for the variation of σ across frequencies:

$$\sigma = \sum_{i=1}^{N_b} \frac{\sigma_0}{\sqrt{w_i}} \quad (\text{A.1})$$

where σ_0 is a constant and w_i is the weight associated to the i -th critical band. In this experiment, as a first approximation, we assume that this weight function would be similar for normal and hearing impaired listeners as the strong frequency dependence of the hearing for hearing impaired listeners is already taken into account by adding both the simulation of elevated thresholds and the simulation of enlarged auditory filters depending of frequencies. As a result, Equation 2 can be rewritten as:

$$d' = \sqrt{\sum_{i=1}^{N_b} \frac{\Delta\mu_i^2}{\sigma_i^2}} = \frac{\sqrt{\sum_{i=1}^{N_b} \Delta\mu_i^2 w_i}}{\sigma_0} \quad (\text{A.2})$$

The weights w_i are derived from the σ_i values obtained experimentally in a group of normal-hearing individuals. To that end, we measured the detection thresholds of pure tones masked by a broadband noise in an experiment involving normal-hearing subjects and deduced the associated values of the internal noise. The aim of this supplementary material is then to describe the experiment and results that led to the weighting values used in the article.

B. 2I-2AFC Experiment

1) *Participants*: Ten volunteers aged from 18 to 43 took part in the experiment. They all had absolute thresholds below 20 dB from 125 to 8000 Hz on both ears excepted two subjects that had larger losses at 6000 Hz (one had a 34 dB loss in left ear, and the other had a 28 and 24 dB loss in left and right ears, respectively). These losses occurring at a frequency much greater than the investigated frequencies, they should in no way affect the results.

2) *Material*: The experiment took place in a soundproof booth. The stimuli were generated with MATLAB version 9.8, processed through a sound card RME Babyface Pro (RME, Germany), and finally presented over circumaural headphones (Beyerdynamic DT 770 Pro, Beyerdynamic, Germany) calibrated with Larson Davis AEC101 artificial ear and Model 824 sonometer (Larson Davis, USA). Besides, subjects' responses were monitored and recorded using a MATLAB interface.

3) *Stimuli*: The stimuli used in the experiment are detailed in Table A-II. The masker was a TEN Noise as defined in [A1]. Besides, the frequencies of the target tones were chosen to be the center frequencies of four independent critical bands: respectively number 9, 14, 20 and 23, corresponding to ERB_N numbers 11, 16, 22 and 25, respectively. The tones were temporally centered in the masker and all stimuli had raised-cosine onset and offset ramps of 20 ms in order to avoid any clicking effect.

Signal	Type	Frequency	Duration
Target 1	Pure tone	520 Hz	300 ms
Target 2	Pure tone	1055 Hz	300 ms
Target 3	Pure tone	2222 Hz	300 ms
Target 4	Pure tone	3158 Hz	300 ms
Masker	TEN noise	Broad band	500 ms

TABLE A-II

SUMMARY OF THE STIMULI USED IN THE MASKED THRESHOLD MEASUREMENT EXPERIMENT. NOTE: TEN = THRESHOLD EQUALIZING NOISE.

4) *Experimental procedure*: The experiment consisted in a two-interval, two-alternative forced choice (2I-2AFC) detection task with a 2-down 1-up adaptive rule. The constant level of the background noise for the test was 75 dB, as well as the starting level of the target signal. At each step, one of the two intervals contained the background noise alone, the other one, a combination of the noise and the pure tone to be detected. The two intervals were successively presented in a random order and the listeners had to indicate the interval they believed to be containing the tone. The level of the pure tone varied according to the listeners' responses. The initial 5 dB step size was reduced to 3 dB after the first 3 reversals and finally to 2 dB after 2 further reversals. The retained masked threshold value was the mean of the last four reversals. For each stimulus, the threshold was measured three times. If the standard deviation of the three measured thresholds exceeded 3 dB, a fourth measurement was made and the definitive threshold was computed as the mean of the three closest values.

C. Experimental results

The model was then run with a constant- σ hypothesis using the experimentally measured masked thresholds of every subject for each tone to determine the associated values of the internal noise. Considering that the frequencies of the tones were the center frequencies of the critical bands number 9, 14, 20 and 23, the associated values of σ are respectively equal to σ_9 , σ_{14} , σ_{20} and σ_{23} since there is no contribution of the other independent critical bands. As the boxplot of the values of σ_i presented on Figure A.1 shows, there is a clear influence of frequency on the internal noise values. This observation is confirmed by a significant effect of frequency evidenced by a repeated measures Bayesian ANOVA applied on the σ_i values ($BF_{01} = 1.041E-6$).

Once it has been established that σ_i has a different value depending on the independent critical band of interest i ,

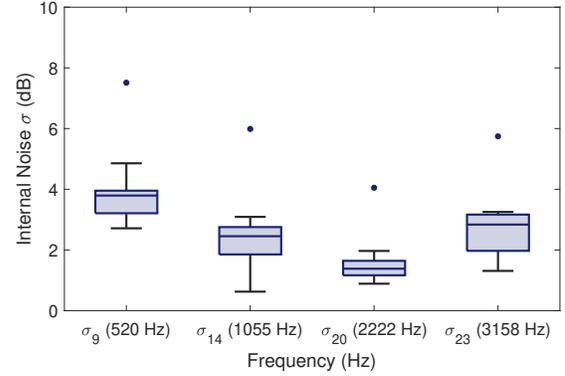


Fig. A.1. Distribution of the values of σ_i among the subjects for $i = 9, 14, 20$ and 23 . On each box, the central mark indicates the median, and the bottom and top edges of the box (of numerical values $q1$ and $q3$, respectively) indicate the 25th and 75th percentiles, respectively. The whiskers extend to the most extreme data points not considered outliers, and the outliers are plotted individually using the “+” symbol. The data points are drawn as outliers if they are greater than $q3 + 1.5 \times (q3 - q1)$ or less than $q1 - 1.5 \times (q3 - q1)$.

we know that the variations of σ_i with frequency have to be considered in the model. For this reason, we need to describe the evolution of σ_i as a function of frequency over all critical bands. However, since measurements have only been made at four frequencies, we cannot directly provide the σ_i values corresponding to the other critical bands. These last are therefore obtained with a second order polynomial regression performed using the experimental data.

Noting $x_i = \log_{10}(f_i)$ where f_i denotes the center frequency of the i -th critical band in Hz, the internal noise in this band can be estimated by the following function:

$$\hat{\sigma}_{i,NH}(x_i) = 8.786x_i^2 - 56.569x_i + 92.959 \quad (\text{A.3})$$

The subscript NH refers to the fact that the estimate is based on data collected from normal-hearing subjects. This function is represented in Figure A.2.

We are well aware that such a way to determine σ_i induces an error, especially at very high and very low frequencies, since they are far away from the range of frequencies we investigated in the experiment. However, this should not affect the accuracy of the model, since it is dedicated to the study of auditory alarms, and these warning signals almost never show energy in these regions of the audible spectrum.

D. Derivation of the weight function

Based on Equation A.1, the square root of the weight of the i -th independent critical band is proportional to the inverse of σ_i calculated in this same band. Therefore, we initially set the values of w_i to $1/\hat{\sigma}_{i,NH}$ and then scale them over the 33 independent critical bands of normal-hearing people so that their sum is 1:

$$\sqrt{w_i} = \frac{1/\hat{\sigma}_{i,NH}}{\sqrt{\sum_{i=1}^{33} (1/\hat{\sigma}_{i,NH})^2}} \quad (\text{A.4})$$

The weights are shown as a function of frequency in Figure A.3.

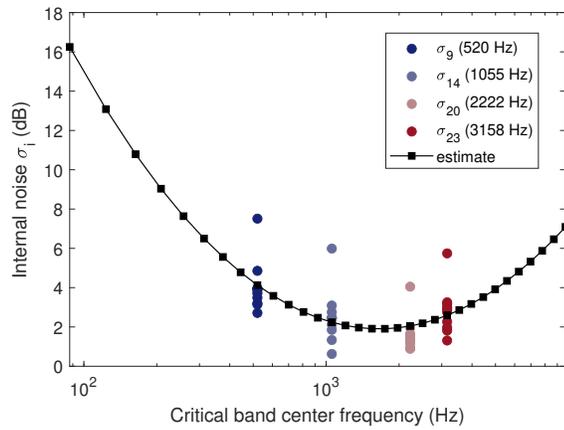


Fig. A.2. Representation of σ_i as a function of the critical band center frequency obtained by polynomial regression.

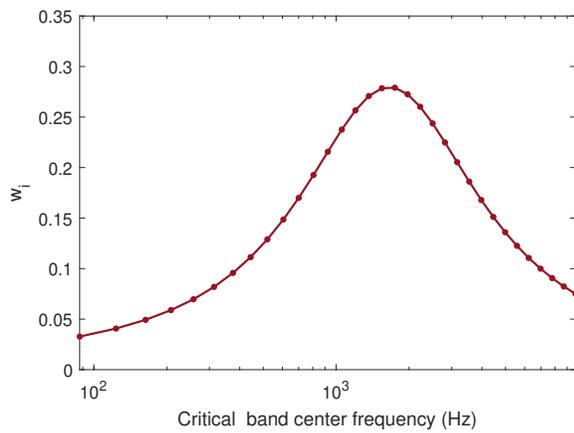


Fig. A.3. Representation of the weights used in the model as a function of the critical band center frequency.

SUPPLEMENTARY REFERENCES

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