Using Neural Metrics to Determine Optimal Hearing Aid Gains for Individual Patients

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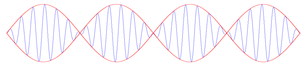
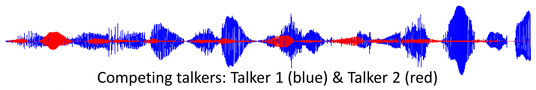
Introduction

Hearing aids can restore some degree of hearing to many patients with auditory impairments. Unfortunately, even with some of the most sophisticated digital hearing aids, many patients have great difficulty understanding speech in the presence of competing background chatter. We hypothesize that this is due, in part, to fitting procedures that do not take into consideration the physiological differences among patients. A hearing aid strategy that accounts for individual differences can be designed to improve temporal coding of the neural signals, thus improving speech perception in noisy conditions. We test this hypothesis by quantifying envelope and fine structure coding in simulated neural responses for normal, impaired, and aided-impaired auditory systems.

The slowly varying envelope of the speech signal is known to be sufficient for speech recognition in quiet. [1] However, it is believed that listeners with normal hearing take advantage of the acoustic waveform's temporal fine structure (TFS) to listen in the short "dips" of the background noise, and perhaps use this information to improve auditory stream segregation.[2] Psychophysical data in the literature suggests that there is a wide variability among individual patients, with some hearing impaired subjects showing a near-normal response to fine structure information.[2]



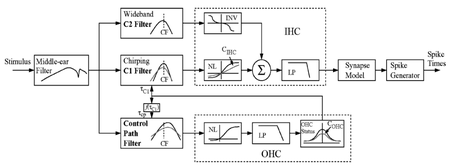
Envelope (solid line) and Temporal Fine Structure (dashed line)



Competing talkers: Talker 1 (blue) & Talker 2 (red)
Note that *fine-structure* cues may be especially important for segregation

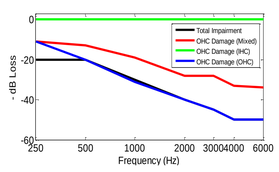
Computational Model Of Hearing Impairment

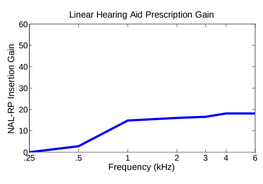
We used a recent computational model of the auditory nerve[3]. This model expands upon several previous models and was chosen because it allows selective control over the health of both outer and inner hair cells. (Where outer hair cells provide gain and sharp tuning, and inner hair cells transduce the acoustical energy to electrical signals.)



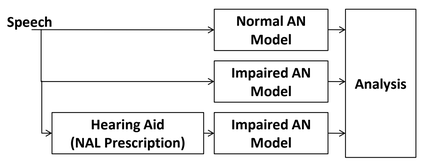
(From Zilany & Bruce 2007)

Several (30) fibers were modeled to represent center frequencies ranging from 250Hz to 8kHz. For impaired hearing simulations, the coefficients C_{OHC} and C_{IHC} were chosen to result in a mild hearing loss, as shown below (only IHC damage, 2/3 OHC damage, only nearly all OHC damage).





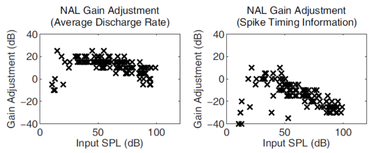
A hearing aid gain profile was fit to this audiogram, based on the NAL-R prescription[4]. A speech stimulus was then run through 3 separate scenarios: a normal-hearing case, an impaired case, and an impaired case with a hearing aid. The resulting neural spike patterns were than analyzed for comparisons across these three cases to quantify the ability of the hearing aid to restore normal temporal coding. The gain was set to the NAL-R prescription, but also adjusted (-40 to +40dB) to determine if the optimal gain differed from the prescribed gain.



Previous Work

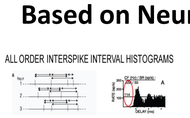
Bruce et al[5] used neurograms to compare normal and impaired responses to individual phonemes in a sentence. The authors modeled auditory nerve fibers of various spontaneous rates across characteristic frequencies from 250Hz to 8kHz. They used a mixed impairment and used two metrics to calculate optimal gains.

Their results suggested that gains above the standard NAL-R prescription are needed to restore average discharge rate to normal, but lower gains are required to restore spike timing information.

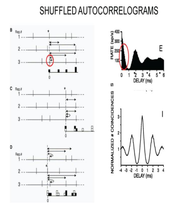


However, the only difference between these two measures was the length of the averaging window; the authors used a very short window size to determine timing. They averaged spike counts using a Hamming window length of 256µs, which has the effect of attenuating fluctuations faster than approximately 2.5kHz. This metric might therefore measure timing (e.g. phase locking) in response to low frequencies, but it may not be sufficient because synchronous timing can be measured up to at least 5kHz in the auditory nerve[6]. Because precise timing may be important for hearing in complex situations, future physiologically-based designs should consider metrics that include both long-term rate and precise temporal coding.

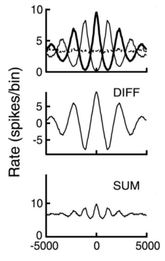
Quantifying Envelope and Temporal Fine Structure Coding Based on Neural Spike Patterns



- An All-Order Interspike Interval Histogram[7] can be used to analyze spike timing, but the metric suffers from poor temporal resolution due to the refractory period of an individual neuron.

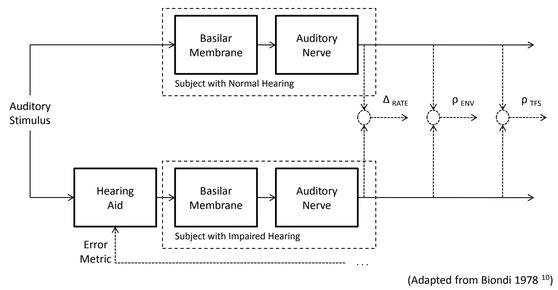


- A Shuffled Autocorrelation[8] (SAC) is similar but, instead of measuring time between spikes in a single spike train, it measures the time from a single reference point (in a different spike train). This allows us to see the temporal resolution that would be represented by a large population of nerve fibers.



- A Cross-Polarity AutoCorrelation (XpAC) can be calculated by comparing the spikes from the original stimulus to spikes from an inverted form of the stimulus. This inversion will flip the polarity of the *Temporal Fine Structure*, but the *Envelope* remains the same.
- The difference between the original SAC and the inverted polarity XpAC represents encoding of anything that is different between these two cases (i.e. the *Temporal Fine Structure*).
- The sum of the original SAC and inverted XpAC represents encoding of that which is the same across both stimuli (i.e. the *Envelope*).

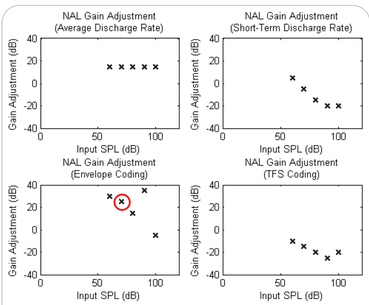
The correlation between normal envelope and aided+impaired envelope (ρ_{env}), as well as the correlation between normal TFS and aided+impaired TFS (ρ_{tfs}) can be calculated.[9] By comparing *Envelope* and *Temporal Fine Structure* coding of aided+impaired systems to the coding of normal systems, we can identify the gain settings which result in neural coding closest to normal.



(Adapted from Biondi 1978 [10])

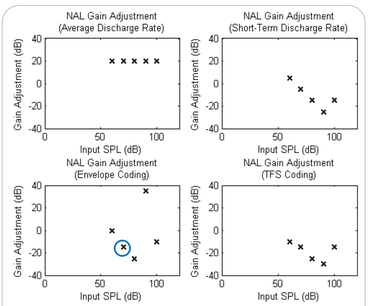
Results

The mean firing rate, ρ_{tfs}, and ρ_{env} were compared across the normal and aided+impaired cases (where the hearing aid applied a simple linear gain to a word in quiet). Each of these metrics was averaged across fibers. The optimal gain was the one which most closely restored the metric to normal.



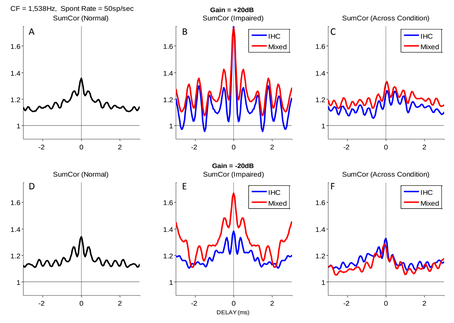
Optimal gains for *mixed hair cell damage*.

Note that gains above NAL are generally better for long term metrics, but gains below NAL are better for the short term metrics. (Results were similar for OHC damage.) Details regarding the circled data point are shown below.



Optimal gains for *inner hair cell damage*.

Note that gains above NAL are still best for average discharge rate, but gains below NAL are better for all other metrics (including envelope coding). Details regarding the circled data point are shown below.



The SumCors for the impaired systems with gain 20dB below NAL are shown in E. Notice that the peak of the IHC curve has lowered substantially, as compared to B, to be much closer to normal envelope coding (see A,D). Also notice than the SumCor of the cross-correlation (C,F) is reduced as the gain is reduced for the mixed damage case (indicating envelope coding further from normal), but the SumCor has increased as the gain is reduced for the case with only IHC damage (indicating envelope coding closer to normal).

Discussion

It is interesting to note that compression, a reduction in gain with increasing level, is needed to preserve timing information, in terms of both short-term rate (based on the neurogram) and temporal fine structure (as calculated using the correlation metrics). Little or no compression seems to be necessary for preserving rate information <100Hz (given an 8ms Hamming window), but it appears that some compression may in fact be necessary for envelope coding (which was only limited in frequency by the characteristic frequency of each nerve fiber).

Of particular interest is the result suggesting that the optimal gain for preserving envelope in a patient with primarily OHC damage is drastically higher than the gain for a patient with primarily IHC damage. If this is in fact true, it would be beneficial to clinically assess a patient's OHC/IHC damage before fitting a hearing aid. Given some information about the underlying physiology, a hearing aid could be better fit for the individual patient. For example, a computational model could be used to match the patient's behavioral performance, then the hearing aid parameters could be adjusted to improve performance of the model. These optimized parameters could then be tested on the patient, thus minimizing the patient's time in the clinic but potentially maximizing performance.

Acknowledgements

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