Improving hearing aid dynamic gain control algorithms based on auditory nerve coding

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Qualifying Literature Assessment

August 19, 2008

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Abstract

Hearing aids have been tremendously successful at restoring some hearing abilities for people with hearing impairments. However, even with some of today's most advanced technologies, patients still have an abnormal degree of difficulty understanding speech in the presence of competing talkers. Hearing scientists believe that listening in complex acoustic environments such as this may require the use of details in the acoustic waveform that might otherwise be unimportant. Some research suggests that the adaptation rate at which hearing aids actively amplify sounds may determine the amount of acoustic information that gets passed to the auditory nerve. Research also suggests that the best adaptation rate for a given patient may depend on the underlying physiology of that particular patient. This paper evaluates the importance of this *fine structure* information in terms of the underlying physiology, hearing aid amplification strategies, and the resulting neural responses. By combining ideas from the behavioral and neurophysiological studies presented here, I propose a study to identify how the amplification speed may be chosen to improve the encoding of *fine structure* information for patients based on their individual physiological impairments.

1 Introduction

The World Health Organization estimates that 278 million people have a disabling hearing loss[62]. These impairments can severely limit communication, and have been shown to reduce quality of life [60]. However, experts suggest that regular use of a prescribed hearing aid can significantly reduce depression in addition to improving communication, cognitive function, and social and emotional well-being [61].

In a normally functioning cochlea (as shown in Figure 1), several rows of outer hair cells actively amplify the vibration of the basilar membrane and a single row of inner hair cells transduces this vibration to electrical signals for the brain. One common type of hearing loss is the result of impaired hair cells (often due to noise exposure or aging) and is referred to as sensorineural hearing loss (SNHL). The typical model of SNHL assumes that most of the impairment is due to damaged outer hair cells which normally amplify low intensity sounds but apply less gain to sounds that are already high intensity.[58]

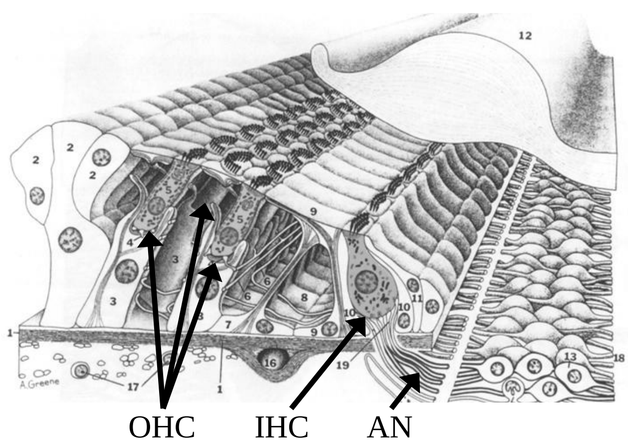

FIGURE 1. ANATOMY OF THE COCHLEA, INDICATING INNER HAIR CELLS (IHC), OUTER HAIR CELLS(OHC), AND AUDITORY NERVE (AN) FIBERS [ADAPTED FROM KIANG 1984[47]]

A hearing aid often attempts to restore this gain (amplification), compressing the dynamic range to make soft sounds audible while keeping loud sounds relatively comfortable. Although hearing aids have been tremendously successful in many situations, patients still have an abnormal degree of difficulty in acoustically complex environments.[19]

Normal hearing people have the remarkable ability, commonly known as the 'cocktail party effect'[12], of understanding a single person in a room full of other people speaking simultaneously. Hearing impaired listeners often complain of an inability to perform such tasks, even when all the sounds are individually audible. Duquesnoy[17] pointed out that people may in fact "listen in the dips" of the of the background noise to extract information about important sounds from a complex mixture. Normal hearing listeners seem to be able to use the small amount of auditory information in short, relatively quiet intervals, but hearing impaired listeners have trouble hearing in this situations. Lorenzi and colleagues[50] suggested that the temporal fine structure (TFS) of the acoustic waveform is important for understanding speech in complex acoustic environments and showed that the ability to listen in the presence of modulated noise is strongly correlated with the ability to utilize TFS. Figure 2 shows an example of a signal decomposed into a slowly varying envelope (E) and rapidly fluctuating temporal fine structure (TFS).

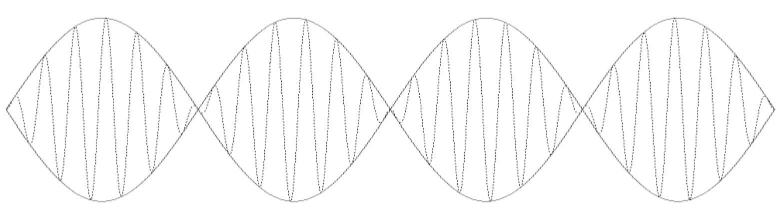

FIGURE 2. ENVELOPE (SOLID LINE) AND TEMPORAL FINE STRUCTURE (DASHED LINE)

Moore's research team[56] has recently suggested that hearing aids with fast-acting gain control may help restore some of this ability. They propose that fast time constants (as will be defined in chapter three) may be necessary to preserve TFS and applying such a strategy may in fact improve speech perception in cocktail party scenarios.

By comparing the neural coding of normal ears with impaired (and amplified) ears[44, 72], one should be able to identify the compression time constants that best restore this coding to normal. More specifically, if precise temporal coding is indeed important for speech perception in a complex acoustic environment, design modifications that enhance temporal coding may also significantly improve speech perception in noisy environments.

Hearing aid gain strategies are first discussed in terms of both behavioral and physiological responses. Auditory coding is then discussed, again with respect to both the behavioral relevance and neurophysiological evidence. Finally, some novel experiments are proposed, which aim to identify improvements to current amplification algorithms for some patient populations.

2 Methods of Literature Review

A vast amount of literature was evaluated for inclusion in this paper. Topics were generally classified as neurophysiological studies, behavioral (psychoacoustic) studies, or review papers (which often try to compare physiology with behavior). As the goal of the proposed research is to utilize physiological metrics to improve behavioral performance, a good balance of these topics was a primary goal when analyzing the literature. However, because the focus is on peripheral neurophysiology (i.e. auditory nerve), studies focused on more central auditory processes were generally excluded. Behavioral studies were limited to those that specifically address the importance of stimulus envelope and temporal fine structure. Other studies were only included for support and to provide context.

Searches for literature generally began with journal articles that were discussed in classes and lab meetings. To find related papers, the reference list was often used as a source. In some cases, correspondence with the authors resulted in more sources. Citeseer and Google Scholar were used to find other (and possibly more recent) articles that cited ones previously found. Occasionally, searches began without any previous material. These literature searches usually began with a Google Web or Google Scholar search for a specific topic. If useful information was found on a website, a more extensive search was performed to find the source of the information. Only information from technical books and peer-reviewed journals was included in this paper. A thorough patent search was also conducted, using the USPTO website and Google Patents. Only patents that addressed both temporal processing and physiological modeling were considered for inclusion.

The Purdue Library search engine was used extensively for finding the text of these papers, usually using the databases included in the '*Medicine + Health*' and '*Engineering + Tech'* category searches (such as PubMed, MedLine, Compendex, Inspec, Science Citation Index, etc). Articles that were not freely available through Purdue were often found on the authors' websites or obtained through personal correspondence.

3 Amplification Strategies

3.1 Automatic Gain Control

The basic function of a hearing aid is to amplify sound. Whether the goal is simply to restore loudness or to improve speech intelligibility, the gain of most modern hearing aids is nonlinear[6]. Just as a normal (nonlinear) cochlea would, a hearing aid can apply substantial gain to low intensity sounds and less gain to sounds that are already of high intensity. As shown in Figure 3, the slope of the input/output function of a nonlinear hearing aid is, by definition, less than one. The gain in this example is decreased as the input level increases, thus compressing the dynamic range of the sound presented to the ear. Because hearing impaired listeners often have linear loudness growth and perceive loud sounds normally (known as loudness recruitment), the

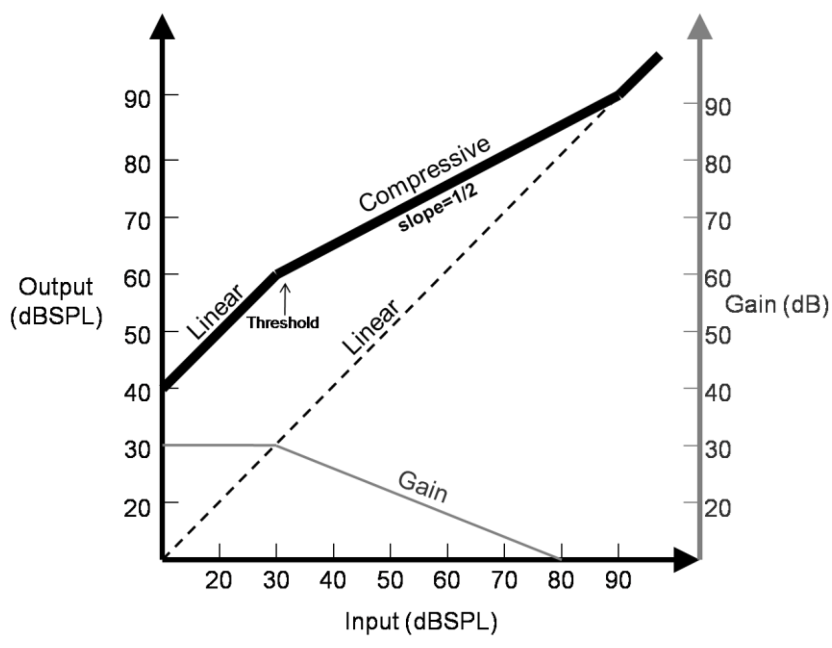

FIGURE 3. STATIC GAIN CURVE - LOW LEVEL INPUTS ARE AMPLIFIED BY A CONSTANT GAIN, BUT ABOVE A GIVEN THRESHOLD THE GAIN IS REDUCED, RESULTING IN A COMPRESSED RANGE OF OUTPUT LEVELS (GAIN IS SHOWN IS GRAY; INPUT-OUTPUT FUNCTION IS SHOWN IN BLACK)

compressive gain of a hearing aid is designed to restore nonlinearity by amplifying soft sounds but minimally affecting more intense sounds.

To automatically control the gain electronically, the incoming sound level is detected and the gain changed accordingly[45]. As shown in Figure 4, a hearing aid typically splits the signal into at least two frequency bands, detects the incoming level, and applies the appropriate amount of gain.

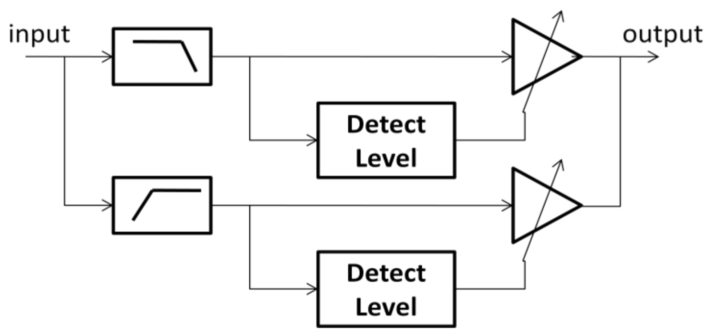

FIGURE 4. SIMPLIFIED HEARING AID BLOCK DIAGRAM – THE INPUT IS FILTERED INTO 2 OR MORE FREQUENCY BANDS, THE INPUT LEVEL IS DETERMINED, AND THE GAIN FOR EACH FREQUENCY BAND IS ADJUSTED

However, this gain adjustment does not occur instantaneously. Because a single sample does not accurately represent the intensity of the signal, the level must be detected over some time interval. Additionally, the gain is often controlled to change somewhat slowly over time and, thus, minimize distortion[82]. (Fast amplitude modulation can result in spectral components that may not otherwise exist.) Typically, the change in amplitude is described by the exponential function, $y = y_0 e^{-\frac{t}{\tau}}$, where y is the amplitude with initial condition y_0, t is time, and τ is a time constant that is chosen by the designer.[54] Figure 5 shows some examples of fast and slow compression.

When the signal level rises above the threshold, the gain is reduced as a function of time. For fast time constants, the gain is reduced over a short period of time. As implemented in many hearing aids, this time constant is often a few milliseconds, which corresponds to the

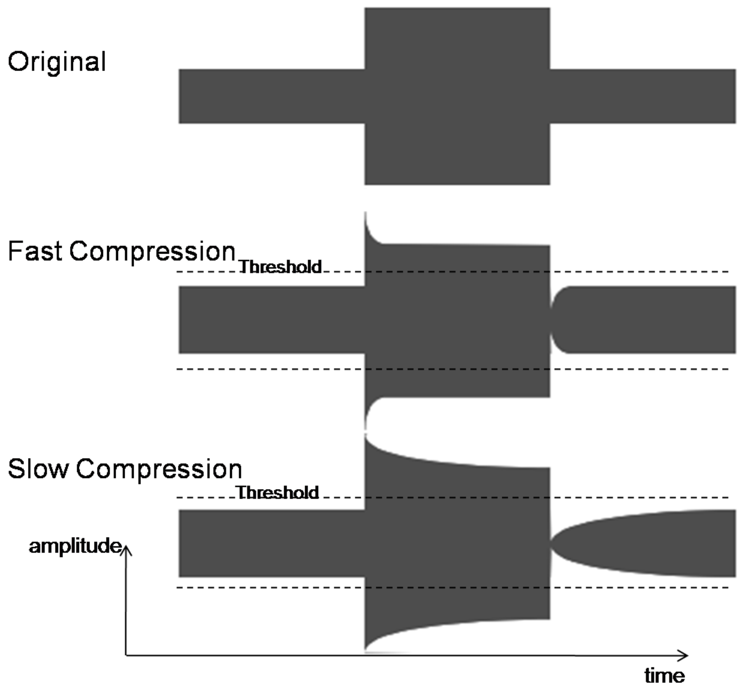

FIGURE 5. DYNAMIC GAIN CURVES – SOUNDS ABOVE THE THRESHOLD ARE COMPRESSED OVER TIME. FAST COMPRESSION RESULTS IN FASTER GAIN ADJUSTMENTS THAN DOES SLOW COMPRESSION

approximate length of a syllable. For slow time constants, the gain is changed slowly, taking many seconds in some designs.[54] One advantage of a fast compression system is that short, quiet syllables will be boosted to levels near those of nearby syllables. However, in addition to the desired signal, noise is also boosted, often resulting in an objectionable *pumping* or *breathing* sound[55]. Slow compression systems do not suffer from noise pumping, but any quiet sounds that follow a period of intense sounds may not be returned to audible levels.[54] It has also been pointed out that relatively slow time constants may be desired to preserve the slowly varying envelope of the signal.[66] However, an 'optimal' time constant will balance the needs for both proper envelope and TFS encoding.

3.2 Noise Control

In a 'cocktail party' situation, in which many people are speaking at once, hearing impaired listeners often have trouble segregating the voice of the person directly in front of them from all the other voices. Unfortunately, this is a very difficult problem to solve. Hearing aid manufacturers have tried to minimize background sounds with directional microphones, adaptive digital noise reduction, and use of binaural hearing aids to improve localization.[6] Many of these techniques are used in modern digital hearing aids, but, unfortunately, only about 50% of patients are satisfied with the performance of their hearing aids in noisy situations.[18] Perhaps one reason for this dissatisfaction is that computational algorithms are not currently able to decide what information is important to preserve and what is background noise. Interestingly, one of the most successful technologies is the use of two hearing aids which are connected (often wirelessly) to provide both stereo compression and microphone beam-forming for improved directionality.[18] This technology simply helps the patient localize the various sounds and leaves the job of filtering to the ear and brain.

One reason for the difficulty with noise reduction is the fact that the physiological mechanisms of hearing impairment and subsequent amplification are not well understood. Although a vast amount of research has gone into the behavioral results of hearing aid design and fitting strategies[56, 65, 82, 83], little is known about the underlying physiology of hearing aid use. *Knowledge of such physiology may be beneficial and may offer new insights into the design of auditory prostheses.*

3.3 Gain Control in Normal Physiology

The ear-brain system normally provides its own gain control. In this physiological "algorithm", there are three primary sources of active cochlear gain adjustment, as shown in Figure 6. The outer hair cells (OHC) provide the first stage of gain control, reacting to the acoustic stimulus by changing length, thus amplifying the vibration of the organ of Corti.[5, 49] (In fact, the mechanical motion of the outer hair cells creates additional acoustic energy, known as otoacoustic emissions, that can be detected with a microphone in the ear canal.[46, 78]) As Rhode[68] showed in his seminal work, the gain of the cochlear amplifier is compressive. The gain provided by the outer hair cells is also frequency dependent – it may be as high as 60dB at the base of the cochlea (in response to low-level high frequency stimuli), with little to no gain at the apex.[70] Recio and colleagues[67] showed that compression could be seen in as little as 100µs. Compression this fast would normally introduce severe distortion but, because the gain is applied to only a very localized region, any distortion is band-limited and would most likely be imperceptible.

The second source of cochlear gain control is the Medial Olivo-Cochlear (MOC) system (see Figure 6). Cells near the medial superior olive in the brainstem project axons back into the cochlea and innervate the outer hair cells. The strength of this reflex appears to be equally controlled by both ears[31] and is known to have two distinct time courses. The fast effect has a time constant of 30-60ms, while the slow effect has a time constant closer to 10-50sec.[13] This moderately fast gain control mechanism appears to be, in part, a protective mechanism[52], and some scientists have suggested that the MOC system may serve to increase the signal-to-noise ratio in noisy conditions.[37] Unfortunately, the strength of the MOC reflex (as measured with otoacoustic emissions) appears to vary substantially from person to person, even among those with normal hearing.[1]

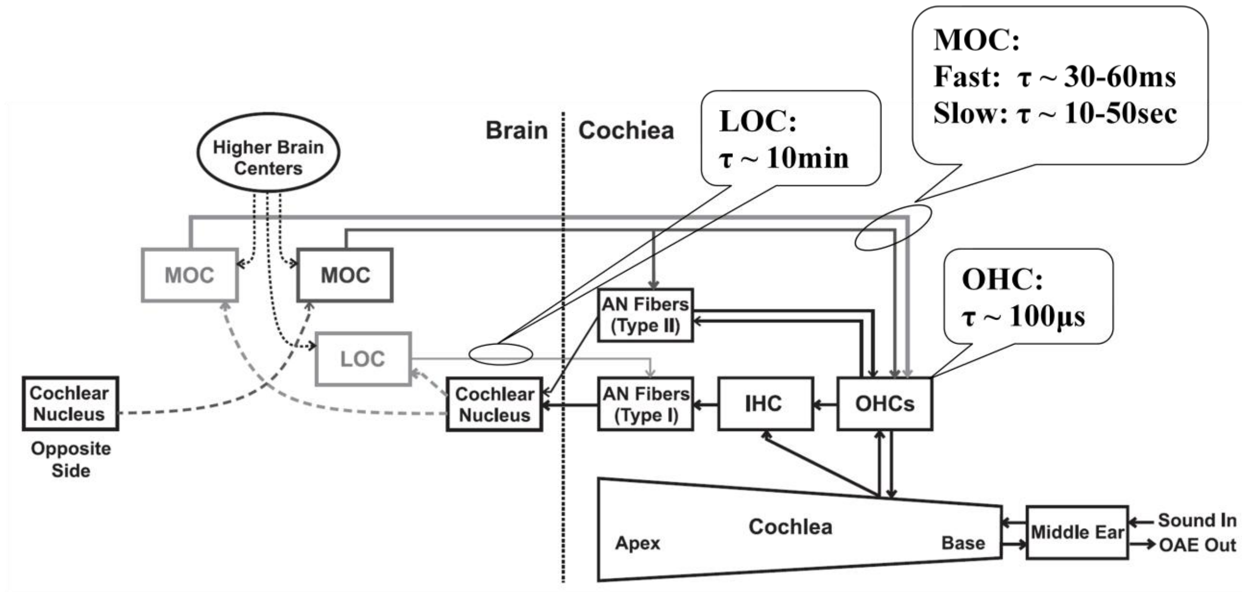

FIGURE 6. PHYSIOLOGICAL GAIN STRUCTURE AND TIME CONSTANTS – THREE PRIMARY GAIN CONTROL MECHANISMS (LOC – LATERAL OLIVOCOCHLEAR EFFERENTS, MOC – MEDIAL OLIVOCOCHLEAR EFFERENTS, OHC – OUTER HAIR CELLS) ARE SHOWN ALONG WITH THEIR RESPECTIVE TIME CONSTANTS

[ADAPTED FROM GUINAN 2006 [31]]

A third and slower gain control system, the Lateral Olivo-Cochlear (LOC) system, uses a set of efferent fibers that come from the lateral superior olive. These cells receive signals from both ears and innervate the auditory nerve fibers in the same hemisphere. LOC efferents appear to be useful for slowly (τ ~10min) balancing the output of the two ears, based on interaural level differences. [15, 30] There may also be some efferent control from higher-level brain structures[59], but little is known about such pathways.

A hearing aid should be designed to replace any impaired gain control mechanism. If OHCs are damaged, it would make sense to partially replace both the fast-acting gain of the OHCs themselves, but also the slower acting gain of the MOC efferents. If IHCs are damaged, it would make sense to use a primarily slow gain control system in a hearing aid. A mixture of both IHC and OHC damage will likely require a moderate time constant that is tuned for the precise ratio of IHC/OHC damage.

3.4 Physiology-Based Hearing Aid Design

Biondi[2] suggested that, by comparing normal and *impaired+aided* neural codes, a hearing aid might be designed that minimizes this difference. Several scientists have used computational models of auditory physiology to implement such a system.[3, 11, 44, 80] For example, Bondy[3, 34] attempted to minimize the difference between normal and impaired coding. However, the authors considered only the rate-place encoding of the auditory signals and did not calculate any measure of phase locking. (Phase locking refers to the fact that auditory neurons tend to fire in sync with a particular phase of the stimulus waveform.) More recently, Bruce[8] claims to have calculated neural information based on both average discharge rate and spike timing. 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. Bruce 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[42] 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.

Goldstein[26-28] has proposed a physiology-based hearing aid design that provides both instantaneous gain (similar to outer hair cells) and a slow adjustment of that gain (to mimic MOC efferent control). This is a promising system, but it is unclear how well it performs or how individual differences would affect the results. A comparison of neural coding in the impaired+aided system with a normal system would shed some light on these questions.

Unfortunately, no previous study has thoroughly characterized neural coding in response to hearing aid gain. Of particular interest is the effect of time constants on neural coding, because this may have a significant impact on speech intelligibility in complex listening conditions. Stone[85] evaluated various compression speeds in a competing talker scenario and argued that fast compression speeds may degrade speech intelligibility in these conditions. However, Gatehouse[20, 21] and Moore[54] point out that some patients may benefit from different compression speeds than others. They argue that individual differences in physiology (e.g. different combinations of inner and outer hair cell damage) may necessitate different time constants. If this is indeed the case, *a thorough physiological study of auditory coding in response to various hearing aid time constants would be beneficial*. Such a study has not yet been published.

4 Auditory Coding

There are two primary ways in which sounds are encoded in the auditory nerve. The first method, often referred to as rate-place encoding assumes that the cochlea acts as a spectral analyzer, resulting in neural firing rates that vary with the level at various frequencies. While this is accurate, the timing of the action potentials also appears to be important. For various stimulus frequencies (especially frequencies below about 4-5kHz), the neural spikes are phase-locked to the stimulus.[32] This second method of encoding is often simply referred to as temporal coding. In fact, the auditory system appears to use both timing and rate as a function of place along the basilar membrane.[10]

Neural coding of speech, in terms of both rate and temporal coding, is an important area of research. Sachs and Young first characterized both rate coding and temporal coding of vowels in the late 1970's [73, 88]. Since that time, there has been much research on the topic of speech coding, but one particularly interesting debate in the field now is the relative roles of the envelope and temporal fine structure in both neural coding and speech perception.[24, 50, 89]

4.1 Envelope and Temporal Fine Structure

Rosen[69] categorized the temporal features of speech into three primary groups – envelope, periodicity, and fine-structure. He defined the envelope as the temporal variations between 2 and 50 Hz, which primarily correspond to the relative amplitude and duration cues that translate to manner of articulation, voicing, vowel identity, and prosody of speech. Rosen defined periodicity as temporal fluctuations between 50 and 500 Hz which contain information about whether the signal is primarily periodic or aperiodic. Rosen's third category, *temporal fine-structure*, consists of the small variations that occur between periods of a periodic signal or within short time intervals of an aperiodic sound. These fast fluctuations, between approximately 600Hz and 10kHz, contain information useful for sound identification. Vowel formants would fall into the category of *temporal fine-structure*. One method for calculating the envelope of a signal is to simply rectify and low-pass filter the signal according to the frequency selection criteria described above. However, Rosen's classifications are somewhat arbitrary and dependent on the specific stimuli that he used.

Another common method for differentiating between envelope and fine-structure is to use the Hilbert Transform. The Hilbert Transform of a signal u(t) is defined as

$$H_{\mathrm{u}}(\mathrm{t})=-\frac{1}{\pi}\lim_{\epsilon\downarrow 0}\int_{\epsilon}^{\infty}\frac{u(\mathrm{t}+\tau)-u(\mathrm{t}\text{-}\tau)}{\tau}\,d\tau\,.$$

The envelope can then be calculated as the magnitude of the analytic signal $u_a(t)$, such that

$$E = |u_a(t)| = |u(t) + i \cdot H_{\mathrm{u}}(t)| = \sqrt{u^2(t) + {H_{\mathrm{u}}}^2(t)}.$$

Note that this calculation of envelope (E) corresponds to both *envelope* and *periodicity* as defined by Rosen. The temporal fine structure (TFS) is then defined as the cosine of the phase of the analytic signal, such that

$$TFS = \cos\varphi(t) = \cos\left(\tan^{-1}\left(\frac{H_{\mathrm{u}}(t)}{u(t)}\right)\right)$$

Note that the signal may then be reconstructed from E and TFS, such that $u(t) = E \times TFS$.[81]

Unfortunately, this distinction is not so clear in the literature since E and TFS are not independent and it may not be possible to completely separate the two in experimental designs.[23, 77] In fact, any attempt to modify one of these will inherently modify the other. (Although much of the research discussed in the next section did not consider this fact.)

4.2 Behavioral Relevance of TFS

Shannon[76] demonstrated that envelope cues alone were sufficient for speech intelligibility in quiet. Smith and colleagues[81] suggested some years later that, while envelope is important for speech recognition, fine-structure is primarily important for pitch perception and localization. (As Bregman[4] points out, pitch and localization are particularly important cues for segregating a mixture of sounds into individual streams, which often correspond to separate sources.) Shannon[75] later found that the number of analysis frequency bands needed for speech recognition was dependent on the difficulty of the listening situation. This implies that, although frequency resolution is not very critical for speech intelligibility in quiet conditions, it becomes necessary as the listening condition becomes more difficult.

Lorenzi's group[50] argued that the inability of hearing-impaired listeners to utilize TFS may be one reason that they have such difficulty listening in noisy environments. The authors showed that, with training, normal hearing listeners could correctly identify approximately 90% of their TFS speech stimuli. However, both young and elderly hearing impaired listeners performed very poorly at the same task, even with training. They also showed a significant correlation between the speech identification scores and the amount of masking release in a separate experiment with the same subjects. (Masking release measures the ability of a listener to hear soft sounds immediately after more intense sounds.) This suggests a relationship between TFS and *listening in the dips* of the acoustic mixture. For example, a listener might identify some sounds during a brief dip in the overall level as belonging to the background. This background sound could potentially then form a separate perceptual *stream*, making it easier to segregate foreground sounds from background. Without the ability to *listen in the dips*, a listener would hear a co-modulated foreground and background, possibly making segregation very difficult.

Hopkins[38] measured speech reception thresholds of keywords in a sentence when there was a competing talker present. (In other words, the authors measured the lowest signal-to-noise ratio at which the target speech was intelligible.) The authors split the spectrum into 32 bands and systematically noise-vocoded (i.e. removed TFS from) a number of the upper frequency bands in an effort to determine the cutoff frequency at which listeners with impairments could no longer utilize TFS. They showed that, although normal hearing listeners were able to improve performance as TFS was introduced at higher frequencies, hearing impaired listeners showed no such improvement. This suggests that the inability to use TFS may not be as simple as a loss of phase locking. However, the authors pointed out that there was a rather large variability among their subjects, with one subject performing very near normal although that subject's hearing thresholds were very similar to the other hearing impaired subjects. Unlike the Lorenzi [50] study, which separately evaluated 7 young listeners (ages 21-35) and 7 elderly listeners (ages 63-72), Hopkins[38] did not separately test young and elderly patients.

There are, in fact, two important factors to consider when working with elderly subjects– age and cognitive ability. Gatehouse [19] showed that listeners with lower cognitive abilities are generally less able to benefit from temporal information, suggesting that more central neural processes may become the limiting factor in such cases. Although cognitive ability is generally assumed to deteriorate with age, Gatehouse and colleagues[19, 20] did not include age in their analysis. (Their subjects ranged in age from 54 to 82.) Separate consideration of cognitive ability and age is particularly important because age-related hearing loss (presbyacusis) may cause unique physiological changes that could potentially affect peripheral encoding[22, 74]. Any study with elderly hearing-impaired subjects should, therefore, be interpreted carefully.

In summary, it appears that envelope cues are often sufficient in quiet, but temporal fine structure is important for listening in complex, noisy conditions. If hearing impaired patients could somehow regain the ability to use TFS, perception could potentially be drastically improved. A thorough analysis of E and TFS encoding at the level of the auditory nerve may reveal new insights into possible therapies.

4.3 Quantitative Analysis of Neural Coding

Characterization of the temporal properties of neural responses has historically been based on simple periodic stimuli such as pure tones or other periodic stimuli. Although metrics such as vector strength[25] or synchronization index[40] provide useful information, they do not apply to complex stimuli such as speech. A metric based on autocorrelation of actual nerve spikes, such as the interspike interval histogram[9], is likely more physiologically realistic and may be used across a variety of auditory stimuli.

Joris[41, 43] recently extended the neural analysis work of Perkel[64] by applying a "shuffled" autocorrelation (SAC) function to differentiate between E and TFS in the auditory system. The SAC is calculated by summing up the number of spikes (in each of many small temporal windows) across several repetitions of a stimulus, but with a single spike within one of the repetitions used as a reference. Unlike the often used all-order interspike interval histogram which simply measures timing between spikes (as shown in Figure 7, A-B), the SAC is not limited by the refractory period of the individual neural responses (Figure 7, C-D).

Joris[43] applied the SAC to a strategic set of signals in order to determine what part of the temporal code was responding to the envelope and which part was due to the temporal fine structure. He presented a stimulus, A+, recorded the resulting neural pattern, and calculated the SAC (Figure 7,E1). He then presented an inverted version of the same stimulus, A-, again recorded the spikes, and calculated the SAC (Figure 7,E2) which matches the first SAC (except for some scaling due to adaptation). By analyzing the spikes from A- in reference to A+, a cross-stimulus autocorrelation (XAC) is then calculated (Figure 7,E3). The envelope of the signal is the same for both A+ and A-, and anything that is common between the SAC and XAC is a measure of the envelope. The sum of the SAC and XAC (labeled SUMCOR in Figure 7.E.4) therefore represents the amount of temporal coding in response to the envelope. The difference (DIFCOR) represents the amount of temporal coding in response to the temporal fine structure

(Figure 7.E.5). The peak of the SUMCOR and DIFCOR functions can then be used to represent the amount of envelope and fine-structure temporal encoding.

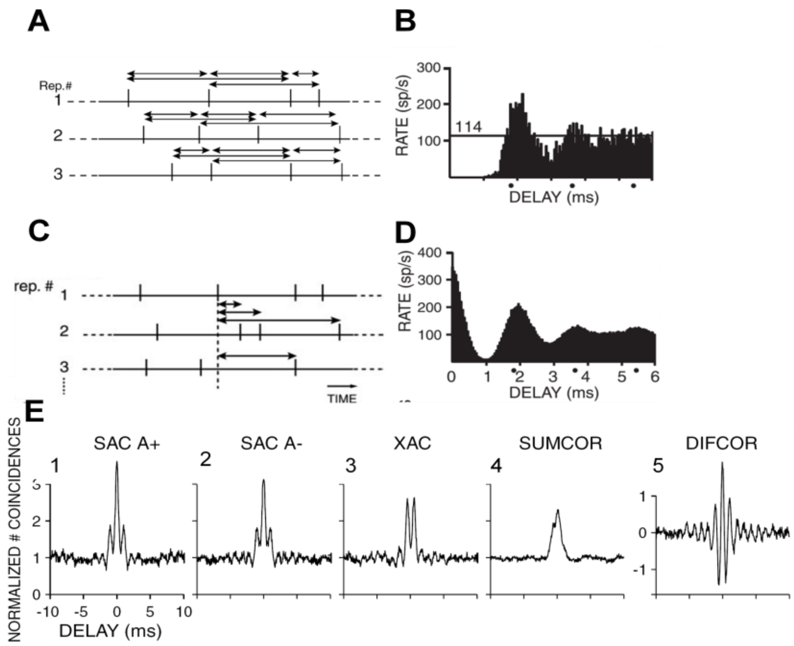

FIGURE 7. A) METHOD FOR CALCULATING AN ALL-ORDER INTERVAL HISTOGRAM(B); C-D) SHUFFLED AUTOCORRELOGRAM (SAC) ACROSS SEVERAL STIMULUS REPETITIONS; E) SAC OF 2 DIFFERENT STIMULUS CONDITIONS (1-2), FOLLOWED BY THE CROSS-STIMULUS AUTOCORRELATION (XAC, 3), THEN THE SUM (4) AND DIFFERENCE (5) OF THE A+ AND A- CONDITIONS   [GRAPHS A,C,E FROM JORIS[43]; GRAPHS B&D FROM LOUAGE[51]]

Unfortunately, because these metrics measure responses to the stimulus as presented to the outer ear (not as sensed in the inner ear), they may not accurately quantify coding in response to the acoustic E and TFS. As Ghitza [23] pointed out, the narrow-band filtering of the cochlea can change the relative amount of E and TFS available to the auditory nerve. The narrow filters of the cochlea are in fact able to generate a "recovered envelope" from the broadband fine-structure (as depicted in Figure 8).

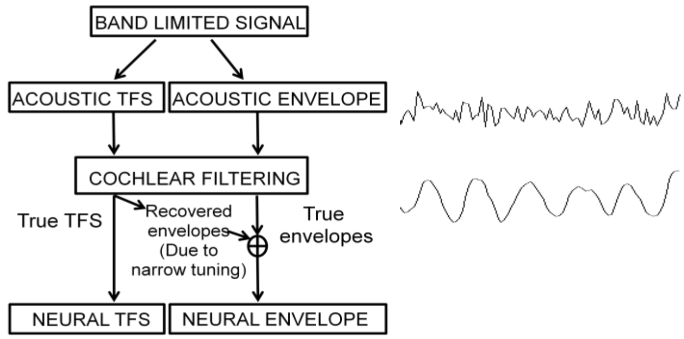

FIGURE 8. THEORETICAL FRAMEWORK FOR IMPROVED QUANTIFICATION OF NEURAL ENVELOPE AND TEMPORAL FINE STRUCTURE (WAVEFORMS ILLUSTRATE A SIGNAL BEFORE AND AFTER COCHLEAR FILTERING)

Our lab has therefore extended these metrics (not yet published) by calculating the correlation between two separate measurements of SUMCOR or DIFCOR (ρ_{env} and ρ_{tfs} respectively). These metrics, as calculated below, can be used to evaluate the extent of envelope recovery and thus give a better approximation of actual TFS and E encoding. ρ_{tfs} and ρ_{env} are defined as:

$$\rho_{tfs} = \frac{difcor(AB)}{\sqrt{difcor(A) \times difcor(B)}}$$

and

$$\rho_{env} = \frac{\sum_{f=10Hz}^{f_c}[CSD(AB) - CSD_{NF}(AB)]}{\sqrt{\sum_{f=10}^{f_c}[PSD(A) - PSD_{NF}(A)] \times \sum_{f=10}^{f_c}[PSD(B) - PSD_{NF}(B)]}}$$

where CSD is the cross-spectral density of the SUMCOR, PSD is the power spectral density of the SUMCOR, and CSD_{NF} & PSD_{NF} are the CSD and PSD values at the noise floor respectively. (See any signal processing textbook[35, 63] for definitions of CSD and PSD.) Note that ρ_{env}, as defined here, only considers spectral content between 10Hz and some frequency f_c. This frequency selection is used to avoid any residual TFS information and may be adjusted to match the E/TFS boundaries specified by Rosen[69].

Unfortunately, these analysis techniques have never been applied to hearing aid design. If used to compare normal, impaired, and impaired+aided responses, we will be better able to identify the strengths and weaknesses of different design approaches.

5 Discussion and Proposed Research

Because temporal fine structure appears to provide important cues for listening to speech in temporally varying noise, a hearing aid algorithm that improves TFS encoding may also improve speech intelligibility in temporally varying noise. Following the suggestions of Gatehouse[19, 21] and Moore[54], scientists may be able to find an ideal set of time constants for a particular patient based on his/her unique environment and physiology. While the clinician can easily learn about the patient's natural listening environment, the physiology is a much more difficult issue. Other work in our lab is attempting to determine the details of impaired physiology from simple behavioral experiments, but even if we knew the underlying physiology, we do not currently know how to tailor treatment based on this information. In fact, even for a generic hearing impairment (i.e. based on a large population of hearing impaired subjects), we do not understand the effects of hearing aid algorithms on temporal coding.

An approach that considers temporal properties of neural coding in the choice of hearing aid time constants is hereby proposed. The proposed study has two stages: computational modeling and single-unit physiological recordings. ***The goal of this proposed work is to identify the time constant that results in the best TFS coding for a particular physiological pathology.***

5.1 Modeling Study

In the first stage of this study, a recent model of impaired cochlear physiology[91, 92] will be used to compare and contrast neural coding of sentences in noise in response to both fast and slow gain control time constants. This computational model, which is an extension of several previous models[7, 10, 36, 90], was chosen because it provides control of both inner and outer hair cell functionality. The model has been compared with data obtained with cats[53, 87] and was found to match the physiological data quite well.[92] Although this model was fit to cat data, the primary difference between a cat cochlea and a human cochlea is the size. The position along the length of the cochlea can be compared across species using the function $x = \frac{1}{a}\log_{10}\left(\frac{F}{A} + k\right)$, where x is the distance from the base of the cochlea, F is the frequency, and the constants a, A, and k depend on the species.[29] Although the frequency tuning in humans may be sharper than in many laboratory mammals[79] (although this remains a topic of some controversy[71]), the trend in all species tested (including humans) is that tuning gets sharper at high frequencies. Any within-species comparisons of normal versus impaired hearing would then be expected to show similar trends in other species.

The computational model will enable analysis of a large amount of data in a relatively short period of time. A simple multichannel hearing aid algorithm[85] will be used to apply dynamic gain. Standard hearing aid prescriptive gains will be used, as defined by the two most common clinical approaches – DSL[i/o], which aims to restore loudness, and NAL-NL1, which aims to maximize speech intelligibility[16]. As shown in Table 1, the coefficients of E and TFS encoding (ρ_{env} and ρ_{tfs}) will be measured in response to various impairments and treatment strategies. (Where ρ_{env} and ρ_{tfs} measure the correlation between the normal system and the system being examined.)

Impairment will be set at either 30dB of hearing loss (dBHL) or 60dBHL. For each of these audiometric thresholds, three variations will be simulated – one in which 100% of the impairment is due to outer hair cell (OHC) damage, one in which 100% of the impairment is due to inner hair cell (IHC) damage, and one with mixed OHC and IHC damage (to approximate noise-induced hearing loss[48]).

TABLE 1. TEST CONDITIONS FOR MEASUREMENT OF E AND TFS ENCODING

		Normal	Impaired	Aided-Impaired			
				DSL[i/o]		NAL-NL1	
				fast	slow	fast	slow
Hearing Loss	Physiology						
30dBHL	100% OHC Damage						
	100% IHC Damage						
	67% OHC (33% IHC) Damage						
60dBHL	100% OHC Damage						
	100% IHC Damage						
	67% OHC (33% IHC) Damage						

If the ratio of inner to outer hair cell damage appears to affect TFS encoding, an additional modeling experiment may be performed to parametrically study this relationship. The ideal time constant (to maximize ρ_{tfs}) could be plotted against this ratio, thus indicating an optimal setting for a given physiological pathology.

In addition to analysis of temporal coding, neurograms will be plotted for comparison with previous data. (A neurogram is simply a three-dimensional plot of neural firing rate as a function of time, across fibers that respond best to different frequencies.) This will also allow us to determine which hearing aid strategy most closely restores rate coding. However, the best strategy for rate coding will likely not be the same as the strategy that maximizes TFS encoding.

Previous studies in our lab (unpublished) have led us to believe that damage to the outer hair cells has a greater detrimental effect on fine structure encoding than does inner hair cell damage.

Therefore, we expect a greater improvement in TFS coding for cases with OHC damage than with only IHC damage.

Another possible experiment is to extend the recent behavioral experiments of Stone and Moore[84, 85] to model neural responses. These authors identified a problem with hearing aid gain control algorithms- the amplitude modulation in response to one sound will also result in modulation of other sounds. This could potentially distort envelope (and TFS) cues, making segregation and intelligibility more difficult. This experiment would involve calculating the gain control signal based on the mixture of a foreground talker and a background talker, but applying the gain to each individually. Stone and Moore calculated the "*across-stimulus modulation correlation*" (ASMC) based on the envelopes of the two signals *a* and *b* such that

$$ASMC = \frac{1}{N}\sum_{i=1}^{N} r(a_i, b_i)$$

where a_i and b_i are the envelopes of *a* and *b*, respectively, in a frequency band *i* and *r* is the Pearson correlation. This could be extended to the neural domain by calculating ρ_{env} and ρ_{tfs} for the Stone and Moore stimuli. In this case, time constants should be chosen which minimize both ρ_{env} and ρ_{tfs}.

5.2 Animal Study

The second stage of the proposed study (which may in fact be implemented iteratively in combination with the first stage if needed) is designed to verify the results of the computational modeling study with similar measurements on animal models. Animals (chinchillas) will be exposed to either narrowband noise or to ototoxic drugs such as Carboplatin (which causes inner hair cell damage[86]) or Kanamycin (which causes outer hair cell damage[14]).

For each animal, an auditory brainstem response[39] (ABR) will be measured to assess the approximate degree of hearing damage and a multiband hearing aid (implemented in Matlab) will be fit to the animal, based on the experimental condition being tested. The resulting action potentials from the auditory nerve will be recorded[53] and analyzed according to the methods discussed earlier.

In particular, the experimental parameters shown in Table 1 will be varied in the animal study to determine the reliability of the modeling study results. By comparing the model results with the neurophysiology results, we can assess the extent to which we can rely on the results of some of the more precise parametric studies proposed. The combination of modeling and animal experiments provides an efficient, yet physiologically realistic, experimentation protocol. The results of these experiments will be used to identify time constants that have the potential of improving speech perception for individual patients in complex listening environments.

6 Conclusion

Hearing impaired listeners have difficulty listening in noisy conditions (such as cocktail parties, construction sites, crowded restaurants, etc.), and current hearing aid algorithms are limited in their usefulness for such conditions. Evidence shows that cues related to temporal fine structure (e.g. pitch perception, localization, timbre, etc.) are important for sound source segregation and listening in noisy conditions. However, the effect of hearing aid gain on TFS is not well understood, and TFS coding may depend on both the algorithms and the underlying physiology. If neural encoding of TFS can be improved for certain physiological impairments through the use of fast compression, hearing aids might be made more effective for listening in noise. *The* *proposed study is designed to find optimal time constants for particular pathologies which could be applied to improving hearing aid design in the clinic.*

If our hypothesis is correct and fast compression improves TFS encoding, we will move to human studies to verify clinical applicability. This will involve assessment of each patient's physiology, using a variety of metrics such as otoacoustic emissions[33], behavioral estimates of cochlear frequency tuning[57], and analysis of confusions made in response to competing talkers. If fast compression has no significant effect on TFS encoding, we will evaluate the reasons and decide whether or not to continue this line of research. One possible extension of this research project is analysis of spectrotemporal encoding[10]- the relative timing of neural responses as a function of location along the basilar membrane. Also, because the proposed research does not take efferent signals into consideration, another possible extension is to include efferent pathways (in both modeling and animal studies).

Current work in our lab is focused on identifying the relationship between individual physiological differences and speech perception in the presence of a competing talker. Our research uses three approaches – single-unit recordings at the auditory nerve, computational modeling of both neural and perceptual processes, and psychophysical experiments with both normal and hearing-impaired subjects. By extending this work to more translational research by including the effects of hearing aid design on both neural coding and perception, we hope to develop technologies that can be used in the clinic.

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