PERCEPTION OF SPECTRAL RIPPLES AND AMPLITUDE COMPRESSED SPEECH BY INDIVIDUALS WITH COCHLEAR HEARING LOSS

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MANASAGANGOTHRI, MYSORE – 570 006

JUNE-2011

Dedicated to

My Amma, Appaji,

Sister, Hazel

&

To my beloved teachers

<u>**CERTIFICATE**</u>

This is to certify that this dissertation entitled ***"Perception of Spectral Ripples and Amplitude Compressed Speech by Individuals with Cochlear Hearing Loss"*** is a bonafide work submitted in part fulfilment for the degree of Master of Science (Audiology) of the student Registration No: 09AUD021. This has been carried out the under guidance of a faculty of this institute and has not been submitted earlier to any other university for the award of any diploma or degree.

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CERTIFICATE

This is to certify that this dissertation entitled ***"Perception of Spectral Ripples and Amplitude Compressed Speech by Individuals with Cochlear Hearing Loss"*** has been prepared under my supervision and guidance. It is also certified that this dissertation has not been submitted earlier to any other university for the award of any diploma or degree

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DECLARATION

This is to certify that this master's dissertation entitled ***"Perception of Spectral Ripples and Amplitude Compressed Speech by Individuals with Cochlear Hearing Loss"*** is the result of my own study under the guidance of Dr. K. Rajalakshmi, Reader in Audiology, Department of Audiology, All India Institute of Speech and Hearing, Mysore, and has not been submitted earlier to any other university for the award of any diploma or degree.

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TABLE OF CONTENTS

Table

List of Figures

Chapter 1

INTRODUCTION

Human speech is highly redundant with spectral and temporal cues. Speech signals contain two forms of information, envelope & temporal fine structure (TFS). Envelope cues (also called as amplitude modulations) correspond to the slow amplitude variations that rate below 50 Hz and fine structure cues correspond to rapid frequency fluctuations that rate above 250 Hz (Rosen, 1992). Importance of these cues for speech recognition has been the research interest in the recent decades. The temporal envelope cues from 3-4 bands are sufficient for the speech recognition in quiet (Shannon et al., 1995). However, recent studies have indicated that the envelope cues alone are not sufficient for the robust speech recognition in noise (Fu & Shannon, 1999; Zeng & Galvin, 1999; Stickney, Zeng, Litovsky & Assmann, 2004; Nie, Stickney & Zeng, 2005). Adding fine structure cues along with envelope significantly improves the speech recognition under background noise (Nie, Stickney & Zeng, 2005; Hopkins & Moore, 2008; Lorenzi & Moore, 2008).

Physiologically information about both the envelope and the TFS is carried by the timing of the auditory nerve discharges. It is commonly believed that envelope cues are represented in the auditory system as fluctuations in the short-term rate of firing in auditory neurons, while TFS is represented by the synchronization of nerve spikes to a specific phase of the carrier (phase locking). In most mammals, phase locking is weak for frequencies above about 5000 Hz (Palmer & Russell, 1986), so TFS information is presumably not conveyed to the brain, or is conveyed with reduced accuracy, for frequencies above 5000 Hz. A reduced ability to use TFS information could explain some of the perceptual problems of hearing-impaired subjects (Lorenzi et al., 2006).

Recent research evidences suggest that cochlear hearing loss adversely affects the ability to use TFS information for speech perception (Qin & Oxenham, 2003; Stickney et al, 2005; Zeng et al, 2005; Lorenzi et al, 2006; Hopkins et al, 2008; Gnansia et al, 2008; Lorenzi & Moore, 2008). This seems likely to be one factor that contributes to the difficulty experienced by cochlear hearing loss individuals when trying to understand speech in the presence of background especially when the noise is also modulated (Festen & Plomp, 1990; Peters et al, 1998; Hopkins et al, 2008). The ability to use TFS information can vary markedly across hearing-impaired individuals (Hopkins & Moore, 2008).

Furthermore, measurements of frequency resolution may be helpful in selecting listener appropriate hearing-aid characteristics (Thornton & Abbas, 1980; Hannley & Dorman, 1983; Tyler et al., 1984). Ability to process TFS may have implications for the choice of compression speed in hearing aids (Moore, 2008). Compression is one of the essential components in hearing aids to fit the wide range of signal levels occurring in everyday life (Levitt, 1982) into the typically small dynamic range of the hearing-impaired person (Miskolczy-Fodor, 1960). An individual who has little or no ability to process TFS information will rely largely on temporal envelope cues in different frequency channels to understand speech. Even though compression offers comfortable hearing to hearing impaired individuals, it also has adverse effect on speech intelligibility by altering temporal envelope cues. Stone and Moore (2003, 2004, and 2008) have shown that fast-acting compression can disrupt the ability to use envelope cues more when compared to slow-acting compression.

Need for the study

Most important information that TFS carries is harmonics of the signal (Moore, Glasberg & Hopkins, 2006). Perception of harmonics are important for perception of pitch and thus for source segregation (Oxenham, 2008). Frequency resolving ability is one factor which determines the perception of harmonics and thus for stream segregation (Bernstein & Oxenham, 2006). Two stimuli (target and interferer) having different harmonic structure but unresolved at cochlear level may form single auditory stream and resulting in poor discrimination.

As discussed earlier when individual cannot perceive fine structure due to reduced frequency resolution, he/she might rely on envelope. So, clinicians must be cautious while prescribing compression parameters which are deleterious to envelope. Spectral ripple discrimination test assess the frequency resolution of an individual’s auditory system. The present study was conducted to investigate whether spectral ripple discrimination test can be used for prescription of compression time constants, and also to correlate between perceptions of spectral ripples with amplitude compressed speech by individuals with cochlear hearing loss.

Aim of the study

To correlate the perception of spectral ripples with the perception of amplitude compressed speech by individuals with cochlear hearing loss.

Objectives of the study

1) To compare the spectral ripple discrimination sensitivity between individuals with normal hearing and cochlear hearing loss.

2) To measure the SNR loss in three conditions which are;
    (i) Original speech
    (ii) Speech stimuli compressed using slow-acting compressor
    (iii) Speech stimuli compressed using fast-acting compressor

3) To investigate the possible correlation between SNR loss & spectral ripple discrimination sensitivity in cochlear hearing loss individuals.

Chapter 2

REVIEW OF LITERATURE

The inner ear is also known as the cochlea. It is shaped like the spiral shell of a snail. However, the spiral shape does not appear to have any functional significance, and the cochlea is often described as if the spiral had been ‘unwound’. The cochlea is filled with almost incompressible fluids, and it has bony rigid walls. It is divided along its length by two membranes: Reissner’s membrane and the basilar membrane (BM). Inward movement of the oval window, produced by movement of the stapes, results in a corresponding outward movement in a membrane covering a second opening in the cochlea, the round window. Such movements result in pressure differences between one side of the BM and the other (i.e., the pressure differences are applied in a direction perpendicular to the BM), and this results in movement of the BM.

A third membrane, the tectorial membrane, lies close to and above the BM and also runs along the length of the cochlea. Between the BM and the tectorial membrane are hair cells, which form part of the organ of Corti (Fig. 1). The hair cells are divided into two groups by an arch known as the tunnel of Corti. Those on the side of the arch closest to the outside of the spiral shape are known as outer hair cells (OHCs), and they are arranged in three to five rows. The hair cells on the other side of the arch form a single row and are known as inner hair cells (IHCs). The stereocilia of the OHCs touch the tectorial membrane, but this may not be true for the IHCs. The tectorial membrane is effectively hinged at one side (off to the left in Fig. 1). When the BM moves up and down, a shearing motion is created; the tectorial membrane moves sideways (in the left-right direction in Fig. 1) relative to the tops of the hair cells. As a result, the stereocilia at the tops of the hair cells are moved sideways. The deflection of the stereocilia of the IHCs leads to a flow of electric current through the IHCs, which in turn leads to the generation of action potentials (nerve spikes) in the neurons of the auditory nerve. Thus, the IHCs act to convert mechanical movements into neural activity.

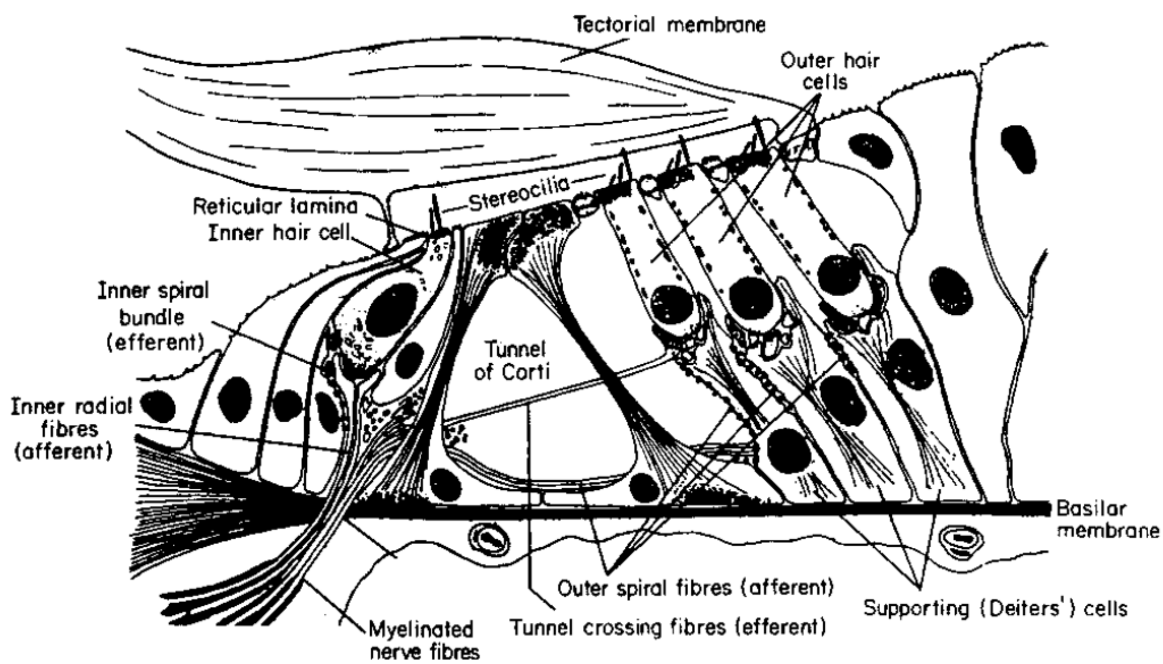

Figure 1. Cross-section of the organ of Corti (Moore, 2003).

The main role of the OHCs is probably to actively influence the mechanics of the cochlea. The OHCs have a motor function, changing their length, shape, and stiffness in response to electric stimulation (Ashmore, 1987; Robles & Ruggero, 2001) and they can therefore influence the response of the BM to sound. The OHCs are often described as being a key element in an active mechanism within the cochlea. The exact way in which the active mechanism works is complex and is still not fully understood. The response of the BM to stimulation with a sinewave (also called a sinusoid or pure tone) takes the form of a wave that moves along the BM from the base toward the apex (Békésy, 1960). The amplitude of the wave increases at first with increasing distance from the base and then decreases rather abruptly.

The response of the BM to sounds of different frequencies is strongly affected by its mechanical properties, which vary progressively from base to apex. At the base, the BM is relatively narrow and stiff. This causes the base to respond best to high frequencies. At the apex, the BM is wider and much less stiff, which causes the apex to respond best to low frequencies. Each point on the BM is tuned; it responds best (with greatest displacement) to a certain frequency called the characteristic frequency (CF), or best frequency and responds progressively less as the frequency is moved away from the CF. The CF decreases monotonically with distance from the base.

It is now believed that the tuning of the BM arises from two mechanisms. One is referred to as the passive system or passive mechanism. This depends on the mechanical properties of the BM and surrounding structures, and it operates in a roughly linear way. The other mechanism is the active mechanism. This depends on the operation of the OHCs, and it operates in a nonlinear way. The active mechanism depends on the cochlea being in good physiological condition, and it is easily damaged. When the OHCs operate normally, the BM shows sharp tuning, especially for low input sound levels. In a living healthy cochlea, the envelope of the traveling wave would have a much sharper peak.

A second function of the active mechanism is to provide level-dependent amplification (also called gain) on the BM. The gain is greatest for low-level inputs (levels below approximately 30 dB SPL), and it decreases progressively with increasing level for levels up to 90 to 100 dB SPL (Sellick, Patuzzi, & Johnstone, 1982; Ruggero, Rich, Recio, Narayan, & Robles, 1997). This level-dependent gain means that the response on the BM is compressive. For example, if the input level of a sinewave is increased from 50 to 60 dB SPL, the response on the BM at the place tuned to the frequency of that sine wave may increase only by approximately 2.5 dB. The compression occurs only for tones with frequencies that are reasonably close to the CF for the place whose response is being measure.

The two theories that are the most accepted nowadays regarding frequency analysis of incoming sound are the place theory and the temporal theory. The place theory proposes that frequency analysis and sound coding is made according to the area of the basilar membrane which has been excited by the incoming sound. This means that for a determined frequency, the basilar membrane, due to its mechanical properties will respond with a specific pattern of vibration. This pattern will have a maximum in a single area. Békésy in 1960 was the first among a number of researchers who confirmed the fact that the point of maximum displacement along the basilar membrane changes as the input frequency is varied. It has also been shown that the linear distance from the point of maximum displacement to the apex is approximately proportional to the logarithm of the input frequency. The specific movement originated by certain frequencies will make the hair cells in the organ of Corti attached to a corresponding area of the basilar membrane to bend. This will generate neural firings that will run through the nerves that form the auditory nerve to superior neural centres.

The temporal theory indicates that frequency information is coded by phase locking to basilar vibration. This is carried out by specialized neurons that are able to combine their performance. This allows the hearing system to code higher frequencies than the corresponding to the minimum relaxation period of a single neuron. It is highly probable that both systems are combined. This may take place in specific frequency ranges and will lead to a better frequency resolution below around 4000 Hz, where there is not temporal coding anymore.

Effects of Cochlear Damage on Loudness Perception

Many of individuals with cochlear hearing loss show a phenomenon called *loudness recruitment*, also called *recruitment* (Fowler, 1936; Steinberg and Gardner, 1937). This may be described as follows. The absolute threshold is higher than normal. When the level of a sound is more than 4–10 dB above the elevated absolute threshold, the rate of growth of loudness level with increasing sound level is greater than normal (Moore, 2004). When the level is sufficiently high, usually around 90–100 dB SPL, the loudness reaches its ‘normal’ value; the sound appears as loud to the person with impaired hearing as it would to a normally hearing person. With further increases in sound level above 90–100 dB SPL, the loudness grows in an almost normal manner

A complementary way of describing this effect is in terms of *dynamic range*. This refers to the range of sound levels between the absolute threshold and the level at which sounds become uncomfortably loud. Typically, in individuals with cochlear hearing loss, the absolute threshold is elevated, but the level at which sounds become uncomfortably loud is about the same as normal (except in cases of severe or profound loss). Hence, the dynamic range is reduced compared to normal.

Loudness recruitment also influences the perception of sounds whose amplitude fluctuates from moment to moment. Examples of such sounds include speech and music. For speech, the most prominent fluctuations occur at rates from about 0.5 Hz up to 32 Hz (Steeneken and Houtgast, 1980). For sounds that are amplitude modulated at rates up to 32 Hz, recruitment results in a magnification of the perceived amount of fluctuation (Moore, Wojtczak and Vickers, 1996); the sound appears to 'wobble' more than it would for a normally hearing person. The magnification of the perceived fluctuation is roughly independent of the modulation rate over the range 4–32 Hz. The fact that the magnification does not decrease with increasing modulation frequency is consistent with the idea that recruitment results mainly from the loss of fast-acting compression on the basilar membrane (BM).

The perception of loudness may be affected by at least four changes that occur with cochlear hearing loss:

1. Elevation in absolute threshold, which may be caused by loss of function of OHCs or IHCs, neural degeneration or a combination of all of these (Schuknecht, 1993). Reduced functioning of the stria vascularis may also be involved (Schmiedt, 1996), but it is assumed that this can be modeled indirectly as reduced functioning of the OHCs and IHCs. On average, the rate at which loudness grows with increasing intensity goes up with increasing absolute threshold at the test frequency (Glasberg & Moore, 1989; Hellman & Meiselman, 1990 and 1993; Kiessling, Steffens, & Wagner, 1993; Miskolczy-Fodor, 1960). This is consistent with the idea that threshold elevation and loudness recruitment are both linked to the loss of the active mechanism in the cochlea. When the absolute threshold is high, the dynamic range can be very small indeed.

2. A reduction in or loss of the compressive nonlinearity in the input-output function of the BM, which is mainly associated with OHC dysfunction. If the input-output function on the BM is steeper (less compressive) than normal in an ear with cochlear damage, it would be expected to lead to an increased rate of growth of loudness with increasing sound level. However, at high sound levels, around 90 to 100 dB SPL, the input-output function becomes almost linear in both normal and impaired ears. The magnitude of the BM response at high sound levels is roughly the same in a normal and an impaired ear. This can explain why the loudness in an impaired ear usually "catches up" with that in a normal ear at sound levels around 90 to 100 dB SPL (Moore, 2007).

3. Loss of frequency selectivity, which results in broader excitation patterns, which is again associated mainly with loss of OHC function. Kiang, Moxon, and Levine (1970) and Evans (1975) suggest that reduced frequency selectivity might be the main factor contributing to loudness recruitment. They suggest that, once the level of a sound exceeds threshold, the excitation in an ear with cochlear damage spreads more rapidly than normal across the array of neurons, and this leads to the abnormally rapid growth of loudness with increasing level. However, both experimental studies (Hellman, 1978; Hellman & Meiselman, 1993; Moore, Glasberg, Hess, & Birchall, 1985; Zeng & Turner, 1991) and theoretical analyses (Moore, 1995) suggest that reduced frequency selectivity plays only a minor role.

4. Complete loss of IHCs or functional neurons at certain places within the cochlea (*dead regions*). The effects of a dead region are included in the loudness model by setting the excitation to a very low value (effectively zero) over the frequency range corresponding to the region assumed to be dead. This has the effect that the specific loudness evoked from that region is zero (Moore, 2007).

Effects of Cochlear Damage on Frequency Selectivity

Frequency selectivity refers to the ability of the auditory system to separate or resolve (to a limited extent) the components in a complex sound. It seems likely that frequency selectivity depends to a large extent on the filtering that takes place in the cochlea (Evans, Pratt, & Cooper, 1989). It is also found that, reduced frequency selectivity would lead to degradation of the recognition of complex sound images; particularly speech sounds (Dreschler & Plomp, 1980; Festen & Plomp, 1983; Gorga & Abbas, 1981; Patterson, Minno-Smith, Weber, & Milory, 1982; Stelmachowizc, Jesteadt, Gorga, & Mott, 1985; Tyler, 1979). Hence, it would be expected that frequency selectivity as measured behaviorally would be poorer than normal in people with cochlear hearing loss.

In an ear with cochlear damage, frequency selectivity is usually reduced. Hence, the excitation pattern contains less detail about the spectrum than would be the case for a normal ear. This leads to a reduced ability to distinguish sounds on the basis of their spectral shape (Summers and Leek, 1994). Also, the internal representation of spectral shape may be influenced by suppression. Suppression in a normal ear may enhance the contrast between the peaks and dips in the excitation pattern evoked by a complex sound (Tyler and Lindblom, 1982; Moore and Glasberg, 1983). This may make it easier to pick out spectral features, such as *formants* in vowel sounds. The loss of suppression associated with cochlear damage would thus lead to a greater difficulty in picking out such features.

Frequency selectivity is often determined using psychophysical tuning curves (PTCs) (Chistovich, 1957; Small, 1959) or to estimate auditory filter shapes using rippled noise or notched noise (Glasberg & Moore, 1990; Glasberg, Moore, & Nimmo-Smith, 1984; Houtgast, 1977; Moore & Glasberg, 1983; Moore & Glasberg, 1987; Patterson, 1976; Patterson & Moore, 1986; Patterson & Nimmo-Smith, 1980; Pick, Evans, & Wilson, 1977). It seems likely that frequency selectivity depends to a large extent on the filtering that takes place in the cochlea (Evans, Pratt, & Cooper, 1989). Hence, it would be expected that frequency selectivity as measured behaviourally would be poorer than normal in people with cochlear hearing loss.

Methods for determining the frequency selectivity

1. Band limiting methods

This is the classic method following an experiment performed by Fletcher (1940). The threshold of a sinusoidal signal was measured as a function of bandwidth of a bandpass noise masker. The noise was centered at the signal frequency and the noise power density was kept constant. This means the overall noise power increased as bandwidth increased. As the noise bandwidth increases the threshold of the signal increases until a certain bandwidth were it flattens off. From this point, further increases in bandwidth did not alter the signal threshold significantly. From the power spectrum model of masking assumptions, increases in the noise bandwidth will result in more noise passing through the auditory filter, as long as the noise bandwidth is less than the filter bandwidth. When the noise bandwidth exceeds the filter bandwidth, further increases in the noise bandwidth do not increase the noise passing through the filter. The bandwidth at which the signal threshold ceased to increase was called *critical bandwidth* (Fletcher, 1940).

The results of critical bandwidth estimates using this method are not in complete agreement with measures using the same method, or with measures using other methods, specifically in low frequencies. In the mid and high frequencies different experiments have given reasonable similar estimates of the critical band values as when using this method. However, there are also contradicting results using this method, where no break point in the threshold curve has been reported by Spiegel (1981). This suggests the ear is capable of integration over bandwidths much greater than the critical bandwidth.

If the filter is assumed to be symmetrical, this method could in principle be used to determine the filter shape (Moore, 1986). However, once the noise bandwidth is as wide as the filter passband, further increases in the noise bandwidth produce negligible increases in the total noise power passing through the filter. Therefore, the method is considered not sensitive enough to determine the filter shape in a useful range (Moore, 1986).

2. Two-tone masking methods and psychophysical tuning curves

Two-tone masking methods are based on the measurement of the perception thresholds of a tonal test signal on the background of another tone, which serves as the masker. As the separation of the tones is increased the threshold of the signal drops. The tones can be placed either symmetrically or asymmetrically around the signal. The method is based on the fact that the effect of masking becomes weaker as the difference between the frequencies of the tonal masker and the tonal test signal increases. The relationship between the masking threshold and the frequency difference between the masker and the test signal (the tonal masking curve or tuning curve) directly reflects the acuity of frequency tuning: the narrower the masking curve, the more acute the tuning. In principle, this method can directly estimate the filter shape (Zwicker & Fastl, 1999).

Psychophysical tuning curves are obtained in a similar fashion but they use only one second tone and are restricted to low signal levels. The curves aim to represent the output of neurons of similar centre frequencies. The results are shown as the power of the second tone required to mask the signal as a function of frequency. The masker can be either a sinusoid or a narrow band of noise. However, when measuring a PTC, the preferred masker is a band of noise rather than a sinusoid, as the use of a noise reduces the influence of *beats* (fluctuations in amplitude) caused by the interaction of the signal and the masker (Kluk and Moore, 2004). When a sinusoid is used as a masker, beats can provide a cue to indicate the presence of the signal when the masker frequency is close to the signal frequency, but not when the masker frequency is equal to the signal frequency. This can lead to a PTC with a very sharp tip (Kluk and Moore, 2004), which may be much sharper than the tip of the auditory filter.

When measuring PTCs, signal is at a low level; therefore it is assumed that it will produce activity primarily at the output of one auditory filter. It is assumed further that, at threshold, the masker produces a constant output from that filter in order to mask the fixed signal. If these assumptions are valid, then the PTC indicates the masker level required to produce a fixed output from the auditory filter as a function of frequency and the curve can be seen as an inverted auditory filter shape (Moore, 1986); this is why the procedure is analogous to the determination of a tuning curve on the BM or a neural tuning curve.

Nelson (1991) compared frequency selectivity in 26 ears with normal hearing and 24 ears with cochlear hearing loss using PTCs. The results showed that individuals with cochlear hearing loss had significantly poorer frequency resolution compared to normal listeners.

Moore (1985) measured PTCs and reported that frequency selectivity is usually reduced in people with cochlear hearing losses. It is concluded that the loss of frequency resolution accompanying cochlear hearing loss is a major cause of the difficulties encountered by the hearing impaired in understanding speech in noisy situations.

3. Rippled-noise method

In this method the filter shapes can be derived by the use of masking experiments with rippled noise. This noise has a long term spectrum that varies sinusoidally on a linear frequency scale. It is produced by adding a white noise to a copy of itself that has been delayed *T* seconds. The delayed noise can be added in phase or out of phase. The in phase adding produces peaks at 0 Hz and at every multiple 1/*T* Hz, while the out of phase adding produces deeps at those locations. The spectrum of the rippled noise is:

$$N(\mathrm{f}) = N_0(1 \pm M\cos 2\pi f\,T)$$

Where N_0 is the original noise spectrum and M is the modulation depth determined by the attenuation of the delayed noise (Moore, 1986).

Based on threshold determination of a pulse sinusoid in the presence of this noise masker according to delay time, *T*, and polarity of the rippled noise, the filter shape has been determined (Houtgast, 1977). This was done using the general masking model and a trigonometric Fourier series application.

The filter shapes that have been determined using this procedure are similar to the ones obtained with other methods. However, there can be practical difficulties in the derivation of the filter shapes when the ripple densities are high due to threshold differences being comparable to measurement errors (Moore, 1986). Furthermore, even in low ripple densities, small measurement errors can give noticeable irregularities in the filter shape. The dynamic range of the measurements is considerably less than in other methods and it requires more subject time in general (Moore, 1986).

Pick, Evans, and Wilson (1977) estimated auditory filter bandwidths using a rippled-noise masker for both normal-hearing subjects and subjects with cochlear hearing loss. They found that when the absolute thresholds of the hearing-impaired subjects were less than about 20 dB HL at the test frequency, the filter bandwidths were generally within the normal range. With increasing hearing loss above 20 dB, the filter bandwidths generally increased. On average, bandwidths reached about twice the normal values for absolute thresholds in the range 40 to 50 dB HL, although considerable scatter was evident in the data. Broadly comparable results were obtained by Hoekstra (1979) and by Festen and Plomp (1983).

4. Notched noise method

This method allows the determination of the auditory filter shape and was designed to prevent off frequency listening (Patterson, 1976). A signal fixed in frequency is located symmetrically at the center of a noise masker with a bandstop or notch centered at the signal frequency.

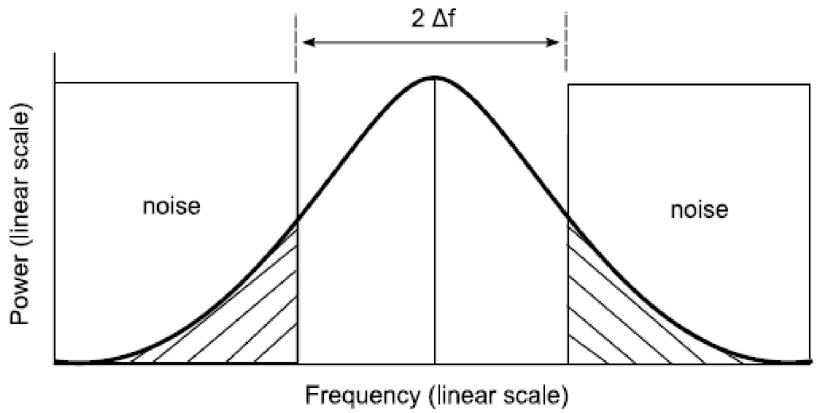

Figure 2. Notched noise method schematic illustration Spring (2007).

The deviation from each of the noise edges to the signal frequency is denoted by Δf. The measurement consists of determining the signal threshold for different notch widths, while maintaining the level of the noise masker constant. Since the signal is symmetrically placed at the center of the notch, the method cannot reveal any filter asymmetries. As the width of the notch is increased, less and less noise leaks through the filter skirts and the threshold is reduced. The variation in threshold with notch width can be seen as a measure of the area of the noise leaking through the filter skirts.

Then, assuming that threshold corresponds to a constant signal to masker ratio, the filter function can be obtained by differentiating the threshold function respect to Δf, given that the integral of a function between certain limits corresponds to the area under that function. This is the basic idea that has been used to determine the filter shapes using this method.

Auditory filter shapes of subjects with cochlear impairments have been estimated in several studies using notched-noise maskers (Dubno & Dirks, 1989; Glasberg & Moore, 1986; Laroche, Hetu, Quoc, Josserand, & Glasberg, 1992; Leek & Summers, 1993; Leeuw & Dreschler, 1994; Peters & Moore, 1992; Sommers & Humes, 1993; Stone, Glasberg, & Moore, 1992; Tyler, Hall, Glasberg, Moore, & Patter son, 1984). The results generally agree in showing that auditory filters are broader than normal in hearing-impaired subjects and that, on average, the degree of broadening increases with increasing hearing loss.

Sommers & Humes (1993) to dissociate the effects of age and hearing impairment on changes in frequency selectivity, auditory filter shapes were measured at 2 kHz in four groups of subjects: (1) normal-hearing young listeners consisted of four subjects; (2) normal-hearing elderly listeners consisted of three older subjects with 2-kHz thresholds less than 15 dB HL; (3) elderly hearing-impaired listeners contained four older listeners with absolute thresholds between 20 and 30 dB HL and moderate impairment, had four listeners with thresholds between 40 and 50 dB HL at 2 kHz; and (4) young normal-hearing listeners with simulated hearing losses consisted of four subjects. Filter shapes were derived using a modified version of the notched-noise procedure (Glasberg and Moore, 1990). Equivalent rectangular bandwidths (ERBs) of auditory filters were not significantly different in young and elderly subjects with normal 2-kHz hearing. Furthermore, filter widths for young subjects with 20- and 40-dB simulated hearing losses overlapped with those obtained from elderly subjects with corresponding degrees of actual hearing loss. One measure that did show significant differences between actual and simulated hearing losses was the degree of filter asymmetry; auditory filters in hearing-impaired listeners were more asymmetrical than those obtained from noise-masked normal-hearing subjects. These findings suggest that the reduced frequency selectivity often reported for older listeners can be attributed, primarily, to hearing loss rather than increased age.

Limitations of masking methods for measuring frequency selectivity

Some practical problems arise when using tonal masking methods. If the signal is a sinusoid, when the masking tones are close to the signal they will beat with the signal. This may provide an additional cue to the subject and the resulting filter will seem to have a notch at its centre frequency Moore (1986). Narrowband noises can be used to avoid the beats. Off frequency listening can affect both of the methods Moore (1986). In the two-tone masking experiments, when the masking tones are narrowly spaced, it is not clear whether the subject is using a filter centred at the frequency of the signal or not, because all of the filters in the region of the signal will have similar signal to masker ratios. Besides, the presence of combination tones produced by interaction of the maskers and the signal can also introduce irregularities in the threshold functions. If precautions are taken care off to reduce this problems (i.e. choosing narrowband signals and appropriate maskers for combination tones), the filters obtained with the two-tone masker method are very similar to the ones obtained with other methods Moore (1986).

Although laboratory methods for measuring the acuity of the frequency tuning of hearing have been developed in detail and are widely used in basic studies, measurements of this type have thus far received almost no clinical application. This is because methods for identifying the acuity of frequency tuning based on frequency-dependent masking are very laborious and require a large investment of time. The duration of the measurement procedure arises because these methods are “multi-point” methods, i.e., determination of one value for the acuity of frequency tuning requires many measurements of masking thresholds using maskers with different frequency characteristics (at different tonal masker frequencies using the masking curves experimental paradigm or different widths for narrowband noise in the critical bands method, and different spectral widths for spectral notches, etc). The results of these measurements are then used to construct curves reflecting the relationship between the masking threshold and the corresponding masking parameter; the width of this curve allows assessment of the acuity of tuning frequency. This complex of measurements can be performed in laboratory conditions, when a single subject can take part in large numbers of sessions. However, this situation is not applicable to diagnostic situations.

Even if the technical complexity associated with the duration of the masking procedure for measuring frequency selectivity is overcome, there remain a number of other fundamental difficulties: when measurements use test signals with narrow frequency spectral tones or narrowband noises, it is difficult to predict how well or poorly the auditory system will be able to discriminate complex spectral patterns. This results from the apparent non-linearity of signal transformations in the auditory system.

Methods for the direct measurement of the ability to discriminate spectral patterns

A wide range of tasks associated with both basic studies of the mechanism of auditory analysis and practical applications need methods which can be used to make direct rather than indirect (via assessment of the acuity of frequency-selective filters) measurements of the ability of the auditory system to discriminate complex spectral patterns. The test signals have to consist of sounds with complex spectra such that the degree of divisibility of the spectral pattern can be unambiguously characterized in precise physical terms. This limiting divisibility of the spectral pattern which can be discriminated by the auditory system will provide a measure of the spectral resolving ability. Two important questions need to be resolved:

1. Identification of the type of spectral pattern most suitable for measuring spectral resolving ability
2. Development of an experimental protocol able to show which spectral pattern can be discriminated and which cannot be discriminated by the auditory system.

An important step towards direct investigation of the spectral resolving ability of hearing was provided by studies reported in (Green, 1983, 1986, 1987, 1988 & 1992; Green, Mason & Kidd Jr, 1984; Green, Onsan, & Forrest, 1987) in which the concept of "auditory profile analysis" was developed. Testing used signals consisting of some number (usually 21) pure tones uniformly distributed on a logarithmic frequency scale. Altering the intensity of individual tones allows "construction" of a spectral pattern: a uniform spectrum (identical amplitudes for all tones), or with "peaks" or "troughs" at certain frequencies, or with periodically alternating increases and decreases. However, construction of spectral profiles from individual pure tones provides only a relatively crude spectral pattern. “Auditory profiles” reflect the nature of interactions between neighboring auditory filters (critical bands) more than the acuity of frequency selectivity (Durlach, Braida, & Ito, 1986).

Another type of spectral pattern suitable for testing the spectral resolving ability of hearing is provided by noise with a rippled spectrum (rippled noise), i.e., a spectrum containing periodically alternating spectral power maxima and minima within a given spectral band. Signals of this type have been used a number of times in psychoacoustics as maskers for measurement of frequency selectivity (Houtgast, 1974 & 1977; Pick, 1980; Pick, Evans, & Wilson, 1977), for studies of the physiological mechanisms of analysis of complex signals (Shamma, Kowalski, & Versnel, 1995; Shamma & Versnel, 1995; Shamma, Versnel, & Kowalski 1995; Versnel, Kowalski, & Shamma, 1995) and for studies of the perception of the peaks of complex signals associated with the frequency interval between the spectral ripples (Bilsen & Wieman, 1980; Yost, 1982, 1996a & 1996b; Yost & Hill, 1979; Yost, Hill, & Perez-Falcon, 1977; Yost, Patterson, & Sheft, 1996). Rippled spectra are suitable for testing the ability to discriminate complex spectral patterns. These signals have relatively complex and fractionated spectral structures, but can nonetheless be completely and precisely characterized by a small number of physical parameters: ripple density (number of ripples per unit frequency interval, i.e., a value inversely proportional to the frequency interval between ripples) and ripple depth (from 0% to 100%). Variation in these parameters can yield spectra of any discreteness and contrast. With the aim of identifying the cases in which spectral patterns can and cannot be discriminated by the auditory system, a spectral ripple phase reversion test was proposed (Popov & Supin, 1984; Supin & Popov, 1987). A noise with a rippled spectrum is replaced at some point in time with a noise of the same intensity and the same spectral bands, the same ripple density and depth, but with peaks and troughs on the frequency scale reversed. If the fine rippled structure of the spectrum can be discriminated, the subject hears a change in the timbre of the sound at this time; if the rippled structure cannot be discriminated (because the ripple density is too high and/or ripple density is too low), the subject hears no change at the time of the change because the signals are identical in all octal parameters other than the positions of the peaks and troughs on the frequency scale before and after the change.

Results obtained from direct measurements of frequency- resolving ability allowed them to be compared with assessments of the acuity of frequency tuning obtained by masking methods. Calculations showed (Supin, Popov, Milekhina, & Tarakanov, 1998) that when the system was linear, auditory filters with equivalent rectangular breadths of 11–12% should support the discrimination of the rippled structure of spectra with densities of up to 6–8 units; in the same way, a spectral structure with double the fractionation level was discriminable. It would appear that non-linearity in transmission of the signal to the auditory system significantly sharpens frequency-resolving ability.

It is more realistic to suggest that analysis of the spectral structure is preformed by a set of peripheral frequency selective filters, though further sharpening of the "internal spectrum” occurs because of lateral suppression or inhibition. These processes are characteristic for all sensory systems: lateral inhibition emphasizes the contrast in the distribution of excitation in sets of neurons; the auditory system is no exception (Evan, 1992). Lateral inhibition in the auditory system yields a number of psychophysical effects (Dubno & Alstrom, 2001; Gifford & Bacon, 2000; Houtgast, 1972; Moore & Vickers, 1997; Oxhenham & Plack, 1998). It probably causes similar contrast-enhancing effects as in other sensory systems.

Spectral peak resolution was investigated in normal hearing, hearing impaired, and cochlear implant listeners. The task involved discriminating between two rippled noise stimuli in which the frequency positions of the log-spaced peaks and valleys were interchanged. The ripple spacing was varied adaptively from 0.13 to 11.31 ripples / octave, and the minimum ripple spacing at which a reversal in peak and trough positions could be detected was determined as the spectral peak resolution threshold for each listener. The results revealed that the normal listeners had the best spectral peak resolution, with an average threshold across listeners of 4.84 ripples / octave and a range of 2.03–7.55 ripples / octave, while cochlear implant listeners had the poorest spectral peak resolution, with an average threshold across listeners of 0.62 ripples / octave, and a range of 0.13–1.66 ripples / octave. The average spectral peak resolution threshold of 1.77 ripples / octave for the hearing impaired listeners was between those of the normal and the cochlear implant listeners.

The role of Envelope and Temporal fine structure (TFS) in speech perception

When a complex broadband sound such as speech is analyzed in the cochlea, the result is a series of bandpass filtered signals, each corresponding to one position on the basilar membrane (BM). Each of these signals contains two forms of information: fluctuations in the envelope (the relatively slow variations in amplitude over time) and fluctuations in the temporal fine structure (the rapid oscillations with rate close to the centre frequency of the band).

The role of envelope and temporal fine structure cues in speech perception has been studied by filtering the speech into several contiguous frequency bands (each like one of the bands) and processing the signal in each band so as to preserve only envelope or temporal fine structure cues. Envelope cues alone can be presented by deriving the envelope in each band and using the envelope to amplitude modulate a noise band or sinusoid centered at the same frequency as the band from which the envelope was derived (Dudley, 1939). With a moderate number of bands (4–16), envelope cues alone can lead to a high intelligibility for speech in quiet both for normally hearing people (Shannon *et al.*, 1995; Loizou, Dorman and Tu, 1999) and for hearing-impaired people (Turner, Souza and Forget, 1995; Souza and Boike, 2006; Baskent, 2006). However, for normally hearing listeners, the intelligibility of speech based on envelope cues alone is very poor when a fluctuating background sound such as a single talker or an amplitude-modulated noise is present (Nelson *et al.*, 2003; Qin and Oxenham, 2003; Stone and Moore, 2003); presumably, the same would apply for hearing-impaired listeners. This suggests that envelope cues alone do not allow effective listening in the dips of a background sound.

The perception of speech in noise, especially modulated noise, seems to depend at least partly on the use of TFS information (Lorenzi et al, 2006a; Gilbert et al, 2007; Hopkins et al, 2008; Hopkins & Moore, 2008; Lorenzi & Moore, 2008).

To effectively make use of the dips in a fluctuating background sound, the auditory system must have some way of determining whether the signal that is present during the dips is dominated by the target speech or by the background sound. The normal auditory system may do this by using information derived from neural phase locking to the temporal fine structure; changes in phase locking of auditory nerve discharges when a dip occurs indicate that the target speech is present in the dip. The difficulties experienced by hearing-impaired people when trying to listen in the dips may reflect a loss of ability to extract or use information from the temporal fine structure (Lorenzi et al 2006)

The effect of hearing loss on the ability to use TFS information

Differences across listeners in the ability to understand speech in noisy environments may relate to inter-individual variations in TFS-processing capacities (Lorenzi et al, 2006; Hopkins et al, 2008; Hopkins & Moore, 2009; Strelcyk & Dau, 2009; Lorenzi & Moore, 2008; Moore, 2008a, 2008b). Consistent with this idea, Lorenzi et al (2006) reported that the hearing impaired listeners showing the worst ability to use TFS cues in speech sounds processed to remove envelope cues also showed the worst ability to understand speech against an amplitude-modulated noise background. The authors speculated that the poor speech intelligibility in fluctuating noise was due to a reduced ability to use TFS cues signaling the presence of speech in the low-level portions or dips of the background noise, impairing ‘ glimpsing ’ of the target speech. With a different approach, Hopkins et al (2008) showed that most hearing impaired listeners (i.e. eight among nine listeners) could not benefit from TFS cues present in the low-frequency bands (<1.6 kHz) of speech stimuli when speech was presented against a competing-talker background, whereas one hearing impaired listener showed near-normal benefit. The reasons for the individual differences were not clear. Hopkins et al (2008) indicated that the benefit gained from the provision of TFS information was not correlated with the average audiometric threshold over the range 0.25 to 4 kHz.

Lorenzi et al. (2006) studied the role of temporal fine structure cues in speech perception by processing speech so as to remove envelope cues as far as possible while preserving temporal fine structure cues. The ability to identify nonsense syllables presented in quiet was measured for three groups of subjects: normally hearing, and young and elderly with mild-to moderate flat hearing loss. For the intact speech, the normally hearing group achieved perfect scores and both groups of hearing-impaired subjects performed nearly as well as the normally hearing group. After moderate training, all groups achieved high scores (about 90 % correct) for the speech with envelope cues. After more extensive training, the normally hearing group also achieved about 90 % correct for the speech with mainly temporal fine structure cues. These results indicate that moderate hearing loss causes a dramatic deterioration in the ability to use temporal fine structure cues for speech perception.

Current cochlear implant systems convey mainly envelope information in different frequency bands, and this may partly account for the relatively poor ability of cochlear implantees to understand speech when background sounds are present. Nie et al. (2005) evaluated the potential contribution of TFS information to speech recognition in noise via acoustic simulations of a cochlear implant. They transformed the rapidly varying TFS into a slowly varying FM signal which was applied to the carrier in each band (which was already amplitude modulated by the speech envelope in that band). They found that, for sentence recognition in the presence of a competing voice, adding this FM signal improved performance by as much as 71 percentage points. This illustrates the potentially large benefit of providing TFS information in a cochlear implant.

Possible reasons for the effect of cochlear hearing loss on sensitivity to TFS

There are several possible reasons why cochlear hearing loss might lead to a reduced ability to process TFS information. Such a loss may lead to:

1. Reduced precision of phase locking. Physiological studies disagree about whether or not this occurs for pure-tone stimuli (Harrison and Evans 1979; Miller et al. 1997; Woolf et al. 1981). For complex sounds, such as synthesized vowels, noise-induced hearing loss can lead to reduced synchrony capture (phase locking to the formant peaks), possibly as a result of diminished two-tone suppression (Miller et al. 1997)
2. A change in the relative phase of response at different points along the basilar membrane (Ruggero, 1994). This would affect mechanisms for decoding TFS based on correlation of the outputs of adjacent places (Carney et al. 2002;

Deng and Geisler, 1987; Loeb et al. 1983; Shamma and Klein 2000; Shamma, 1985)

3. More complex and more rapidly varying TFS resulting from broader auditory filters, which might make it more difficult for central mechanisms to decode the TFS information

4. A mismatch between TFS information and the place on the basilar membrane that would "normally "respond to that information. TFS information may be decoded on a place-specific basis (Huss and Moore 2005; Moore 1982; Oxenham et al. 2004; Srulovicz and Goldstein, 1983). For example, the TFS at a 1-kHz rate may be decoded best by the central neurons that are tuned (in a normal ear) close to 1 kHz. Hearing loss may produce a shift in frequency-place mapping (Liberman and Dodds, 1984; Sellick et al. 1982) and disrupt the decoding process. Note that this explanation depends on the assumption that the central decoding mechanism is, to some extent, "hard wired"

5. There may be central changes that occur following cochlear hearing loss, such as loss of inhibition, and such changes might disrupt the mechanisms for decoding the TFS

Compression or Automatic gain control (AGC)

People with cochlear hearing loss usually experience loudness recruitment and the associated reduced dynamic range (Fowler, 1936; Moore, 2004, 2007; Steinberg & Gardner, 1937). Most modern hearing aids incorporate some form of compression or automatic gain control (AGC) to deal with this. In principle, AGC can provide high gain for low-level sounds, overcoming the loss of sensitivity of the hearing-impaired individual, and reduced gain for high-level sounds, preventing such sounds from becoming uncomfortably loud. However, controversy continues about the “best” way to implement AGC, and particularly whether it should be fast acting or slow acting.

The basic characteristics of an AGC system can be described in terms of the input—output function, which is a plot of the output level as a function of the input level. For low input levels, the gain (output level in dB minus input level in dB) is independent of input level, and the input–output function has a slope of one. At higher input levels, the gain decreases with increasing input level, and the input–output function has a slope less than one. The compression threshold is defined as the input level at which the gain is reduced by 2 dB, relative to the gain applied in the region of linear amplification (American National Standards Institute [ANSI], 2003). For example, if the gain were 25 dB for input levels well below the compression threshold, the compression threshold would be the input level at which the gain was reduced to 23 dB. One reason for having a compression threshold, with linear amplification for lower levels, is that it is impractical to continue to increase the gain indefinitely as the input level decreases. A second reason is that the use of high gain for very low-level inputs can make microphone noise or low-level environmental noise sound intrusive. Indeed, for very low-level inputs, some hearing aids reduce the gain to prevent such noises from being audible; this is called expansion as opposed to compression.

The “amount” of compression is specified by the compression ratio, which is the change in input level (in dB) required to achieve a 1-dB change in output level (for an input level exceeding the compression threshold); the compression ratio is equal to the reciprocal of the slope of the input–output function in the range where the compression is applied. For example, a compression ratio of 3 means that the output grows by 1 dB for each 3-dB increase in input level. The compression ratio is usually measured using a steady signal, such as a continuous sine wave. When the level of the sine wave is changed, the gain is allowed to stabilize before a measurement is taken. The compression ratio measured in this way will be referred to as CR_{static}.

Choosing compression speed in hearing aids

Moore (2008a, b) has suggested that the ability to process TFS information may have implications for the choice of compression speed in hearing aids. Compression is used in most commercially available hearing aids to fit the wide range of signal levels occurring in everyday life (Levitt, 1982) into the typically small dynamic range of the hearing-impaired person (Miskolczy-Fodor, 1960).

An individual who has little or no ability to process TFS information will rely largely on temporal envelope cues in different frequency channels to understand speech. Stone and Moore (2003, 2004 & 2008) have shown that fast-acting compression can disrupt the ability to use envelope cues. In particular, when the input signal is a mixture of voices from different talkers, fast-acting compression can introduce cross-modulation between the voices, because the time-varying gain of the compressor is applied to the mixture of voices (Stone & Moore, 2007). Voices which are independently amplitude modulated at the input to the compressor acquire a common component of modulation at the output. This decreases the ability to perceptually segregate the voices, and leads to reduced speech intelligibility under conditions when TFS cues are removed by the use of a noise or tone vocoder (Stone & Moore, 2004, 2008). Hence, for an individual with little or no ability to process TFS information, slow-acting compression might be more effective than fast-acting compression.

For a hearing-impaired individual who retains some ability to process TFS, the situation is different. Fast-acting multi-channel compression can help to restore the audibility of low-level portions of signals (the dips), and information derived from TFS can be used to extract ‘glimpses’ of the target speech during dips in a background sound. Thus, fast-acting multi-channel compression may lead to improved intelligibility of speech in the presence of sounds with spectral and/or temporal dips for such an individual (Moore, Peters, & Stone, 1999).

The conclusion from all this is that measures of the ability to use TFS information might be useful in determining the most appropriate speed of compression for a hearing-impaired individual. It is possible that the ability to process TFS is related in a more general way to the speed and accuracy of neural processing in the brain. If this were the case, the ability to process TFS could be related to cognitive abilities. This might explain the link between cognitive abilities and the benefit of fast compression for listening in the dips (Gatehouse, Naylor, & Elberling, 2003).

Chapter 3

METHOD

The aim of the study was to determine the spectral ripple discrimination thresholds and to correlate it with speech identification abilities measured in terms of SNR loss using QuickSIN protocol (Killion, 1997) with and without compression by individuals with cochlear hearing loss. The following method was adopted to realize the aim.

Participants

20 participants were recruited for the current study. The participants were divided into two groups namely, control and clinical group. The control group comprised of 8 participants (N=15 ears) with normal hearing sensitivity. They were age matched to compare spectral ripple discrimination threshold. The clinical group comprised of 12 participants (N=17 ears) with hearing impairment.

Control group

1. The age range was from 15 to 55 years.
2. The pure tone average (PTA) was less than 15 dB HL in the frequencies between 250 Hz to 8000Hz.
3. They had normal middle ear status.
4. They had no history of psychological or neurological or cognitive problem.

Clinical group

1. The age range was from 15 to 55 years.
2. The pure tone average (PTA) was between 26 and 55 dB HL.
3. They were diagnosed as having sensorineural hearing loss, with elevated air and bone conduction thresholds and within 10 dB of each other.
4. The test ear had flat configuration of hearing loss. The highest and the lowest threshold across frequency from 250 Hz to 8000 Hz were less or equal to 20 dB.
5. They had normal middle ear status.
6. The latency difference of V peak auditory brainstem response was less than 0.8 ms between the repetition rates of 11. 1/ sec and 90.1 /sec.
7. They had acquired hearing loss.
8. They had no history of psychological or neurological or cognitive problem.

Instrumentation

1. Calibrated two channel diagnostic audiometer Madsen Model Orbiter 922 version 2 coupled with acoustically matched TDH 39 headphones housed in MX – 41/AR was used to estimate the Pure-tone thresholds, Speech Recognition Thresholds (SRT) and Speech Identification Scores (SIS). Radio ear B - 71 bone vibrator was used to estimate the bone conduction threshold.

2. Calibrated middle ear analyzer GSI Tympstar version 2 was used for tympanometry and reflexomety.

3. The otoacoustic emissions were measured using ILO 292 Echoport Plus.

4. To record and analyze ABRs IHS Smart EP version: 3140 (Intelligent Hearing Systems, Florida, USA) was used with Eartone 3A insert earphones to deliver the stimuli.

5. Computer software: Mathworks (MATLAB, v-7) was used to develop the spectral ripple discrimination test and Adobe Audition 3 was used to simulate compression on QuickSIN sentences in Kannada developed by Methi, Avinash, & Kumar (2009).

Test environment

All the testing was carried out in a sound treated room and noise levels within permissible limits as per ANSI (1991).

Preparation of stimuli

1. **Spectral ripple discrimination:** Ripple noises were generated using MATLAB as described by Won, Drennan & Rubinstein (2007). Two hundred pure-tone frequency components with the duration of 500 ms were summed to generate the rippled noise stimuli. The starting phases of the components were randomized for each presentation. The amplitudes of the components were determined by a full-wave rectified sinusoidal envelope on a logarithmic amplitude scale. The ripple peaks had equal space on a logarithmic frequency scale. The overall bandwidth of rippled stimuli was 100-5,000 Hz with a peak-to-valley ratio of 30 dB. The ripple stimuli were generated with 8 different densities, measured in ripples per octave, those were 1.000, 1.414, 2.000, 2.828, 4.000, 5.657, 8.000 & 11.314. For standard ripples, the phase of the full-wave rectified sinusoidal spectral envelope was

created using ‘sin’ function and for inverted ripples, it was ‘cos’ function (Fig 3). The stimuli were ramped with 150 ms rise/fall times.

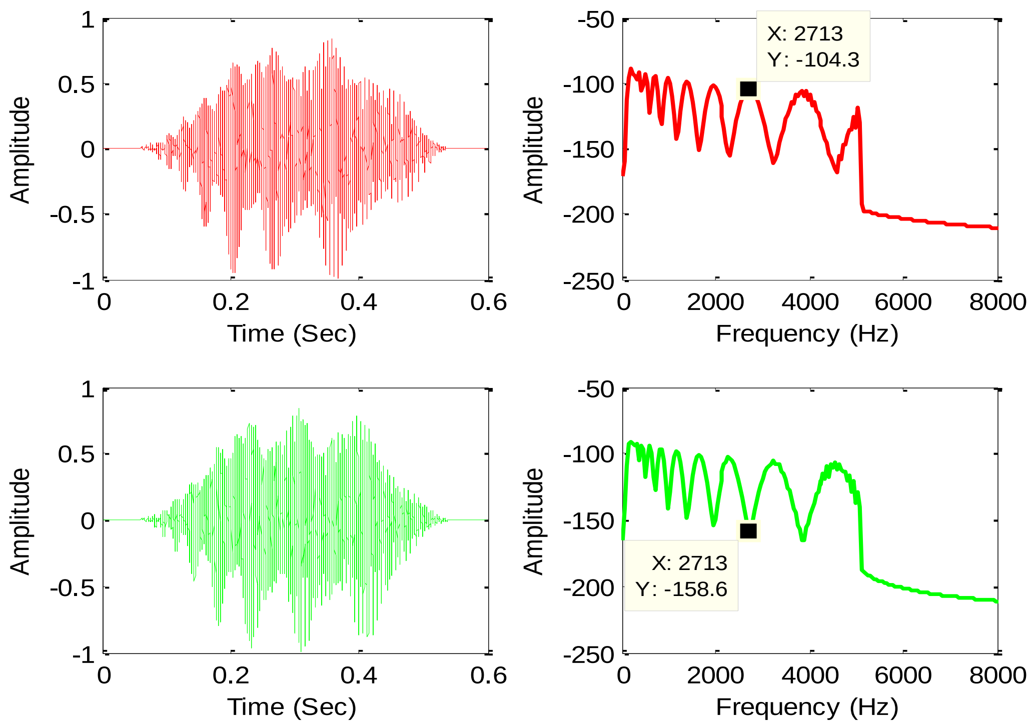

Figure 3. Time and spectral representation of rippled noise

2. **Speech Identification task:** Compression algorithms were implemented using Adobe Audition 3 software with following parameters: A compression ratio of 3:1 was used for both fast-acting and slow-acting compression. An attack time and release time of <5ms and 50ms respectively were used for fast-acting compression (e.g., Walker & Dillon, 1982). Similarly an attack time and release time of around 500ms (Plomp, 1988; Festen et al 1990; Moore et al 1991) were used for slow-acting compression.

The quick speech-in-noise sentence lists developed by Methi, Avinash & Kumar (2009) were used. Each list consisted of seven sentences recorded at +20, +15, +10, +5, 0, -5, and -10 respectively. The present study consisted of six lists of which four lists were digitally compressed. Among compressed lists two lists simulated slow-acting compression and two simulated fast-acting compression. The remaining two lists were retained (uncompressed). The Speech identification was tested in different conditions as listed below:

a. Uncompressed condition

b. Compressed condition

    - Slow-acting compression
    - Fast-acting compression

Procedure

1. Air conduction and bone conduction threshold were obtained using modified Hughson and Westlake procedure S3.21-*1978* (R-1992) using + 5 dB and -10 dB step-sizes.
2. The speech identification test was administered using phonetically balanced Kannada words which were presented through the auxiliary input of the audiometer at a level of 40 dB SL (ref: SRT).
3. The tympanogram was obtained using 226 Hz probe tone frequency with pump rate of 50 daPa / unit time. Ipsilateral and contra lateral acoustic reflex thresholds will be obtained at 500 Hz, 1 kHz, 2 kHz and 4 kHz by varying the intensity of stimulus in 5 dB-steps to observe changes in acoustic admittance.

4. To confirm OHC dysfunction TEOAE's were recorded. Transient otoacoustic emissions evoked by clicks presented at 85 dB SPL for the linear clicks were recorded. The probe with an appropriate sized tip was positioned in the external ear canal and was adjusted to give flat stimulus spectrum across the frequency range. The response was acquired using the linear averaging method. The two averaged TEOAE waveforms of each memory buffer composed of 260 click trains; they were automatically cross-correlated and used to determine the reproducibility of the measured TEOAEs by the software. The response was considered to be present when the emission amplitude was 3 dB more than the noise floor and had reproducibility more than 70%. The absence of the TEOAEs or reproducibility less than 70% and reduced overall amplitude in the presence of hearing loss were considered as indicators of hearing loss.
5. The auditory brainstem response was obtained at two repetition rates at 90 dB nHL. The latency difference of V peak shall be less than 0.8 ms between the waveforms obtained at two repetition rates will rule out retro cochlear pathology.
6. Spectral ripple discrimination threshold: Ripple resolution thresholds were determined using a three interval forced-choice adaptive procedure, based on the method developed by Henry & Turner (2003). One interval contains stimuli with standard & reverse phase separated by 10 ms (we refer this as variable interval) another interval contains stimuli with standard & standard or reversed & reversed phase (we refer this as standard interval). The position of the standard & variable interval will be randomized across the presentation and within variable interval also position of standard & reverse phase stimuli was randomized. Highest ripple

density at which phase reversal could be perceived by participants were estimated using simple up-down procedure (Levitt, 1971). Three numerically labeled buttons were displayed on the computer monitor, corresponding to the three intervals, and subjects were instructed to press the button corresponding to the interval that sounded “different” (i.e., that contained the test stimulus), ignoring any loudness variation between intervals. Correct answer feedback was provided throughout the experiment. Each test run commenced at a ripple frequency of 1.000 ripples / octave, and the ripple frequency was varied in a one-down, one-up procedure. After each incorrect response the ripple frequency was decreased by a step, and it was increased after a correct response and thresholds corresponds to 50% point on psychometric function.

7. Speech Identification task: In all the conditions, speech materials were routed through the speech channel of standard clinical audiometers, and testing took place in sound treated test booths. The sentences were presented at 40 dB SL via TDH-39 headphones. All the subjects were tested monaurally. Prior to the test session, three prototype lists were administered at 40 dB SL to familiarize the subjects with the task.

Scoring: one point was given for each of five key words repeated correctly in each sentence. Half credit was given for words close to the target word. The SNR-50 was calculated for each sentence using a formula as recommended by Methi, Avinash, & Kumar (2009) for obtaining spondee thresholds:

$$\text{SNR loss} = 28.67 - (\text{total words correct})$$

Analyses

The mean data obtained for spectral discrimination thresholds in individuals with normal hearing and hearing impairment is represented on a bar graph with ±1 standard deviation. Shapiro-Wilk’s test was administered to test whether data of spectral ripple discrimination thresholds from both the groups were normally distributed. A parametric independent sample ‘t’ test was chosen to investigate the main effect of hearing loss on spectral ripple discrimination threshold. Gaussian nature of the data was assessed using Shapiro-Wilk’s test for normality for SNR loss in all the three conditions. Pair wise comparisons between the three conditions were performed using Wilcoxon signed rank test. Bonferroni’s adjustments were made for each pair wise comparisons to account for the multiple comparisons. To investigate the possible association between the types of compression and SNR loss Spearman’s rank correlation analysis was performed. All the statistical analyses were carried out using Statistical Package for Social Sciences (SPSS) software version 17.0.

Chapter 4

RESULTS

The present study was designed to determine the spectral ripple discrimination thresholds and to correlate it with speech identification abilities measured in terms of SNR loss using QuickSIN protocol (Killion, 1997) with and without compression by individuals with cochlear hearing loss. Spectral ripple discrimination thresholds were measured using standard phase reversal test. Highest ripple density at which phase reversal could be perceived by participants were estimated using simple up-down procedure (Levitt, 1971). Since, ripple density was adjusted in 1up & 1down manner, the thresholds corresponds to 50% point on psychometric function. Spectral ripple discrimination thresholds were measured in both groups; individuals with hearing loss (experimental group) & individuals with normal hearing sensitivity (control group).

Shapiro-Wilk's test was administered to test whether data of spectral ripple discrimination thresholds from both the groups were normally distributed. Shapiro-Wilk's test for normality compared the distribution of current data against the normal distribution. Results revealed that data of spectral ripple discrimination threshold from both groups are normally distributed (Experimental group; $W=0.98$, $p=0.93$ & Control group; $W=0.88$, $p=0.06$). So, a parametric independent sample 't' test was chosen to investigate the main effect of hearing loss on spectral ripple discrimination threshold. Independent sample 't' test revealed that spectral ripple discrimination thresholds obtained from both groups are significantly different($t(30)=-0.85$, $p < 0.05$). Levene's test for equality of variances indicated an equal variances between both groups ($F = 0.01$, $p =$

0.92). So, no adjustments were done to degrees of freedom. Spectral ripple discrimination thresholds were significantly better in normal hearing individuals (Mean = 3.5 ripples/octave) when compared to individuals with cochlear hearing loss (Mean = 1.6 ripples/octave) which can be observed from figure 4. This result confirms the previous studies by Hopkins & Moore (2006), that individual with cochlear hearing loss has poor sensitivity to spectral fine structure.

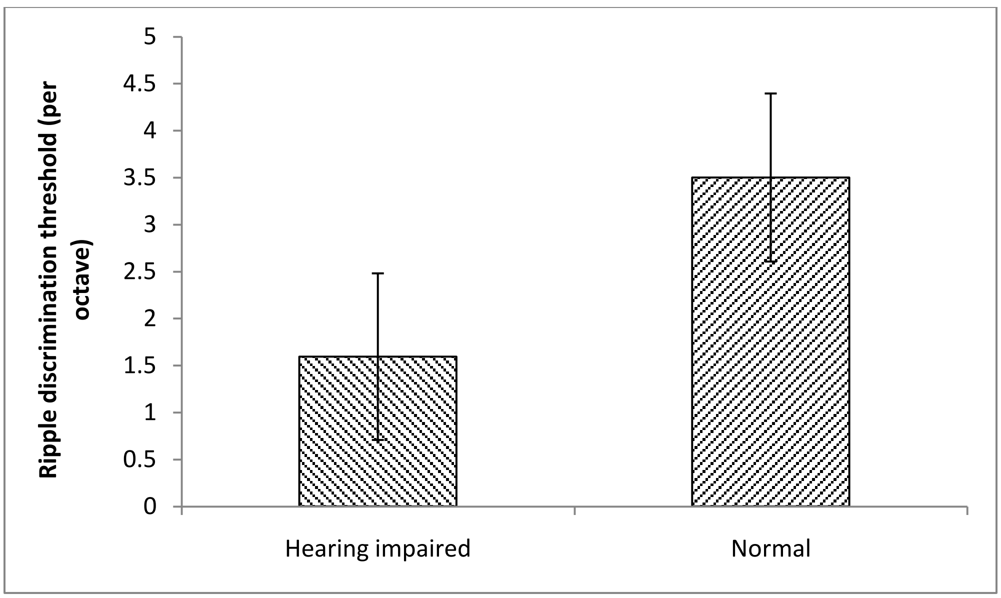

Figure 4. Bars represent mean ±1 SD Spectral ripple discrimination thresholds in individuals with normal hearing and hearing impairment.

Speech identification abilities by individuals with cochlear hearing loss were assessed using QuickSIN protocol (Killion, 1997). QuickSIN does not measure speech identification scores instead it measures SNR loss. SNR loss indicates loss in ability to understand speech at the SNR used by those with normal hearing (Killion, 1997). SNR loss was calculated for each individual using following formula as recommended by Methi, Avinash, & Kumar (2009). SNR loss was measured for three conditions which are; (i) Speech stimuli compressed using fast-acting compressor hence forth this condition will be regarded as 'fast-acting compression' (ii) Speech stimuli compressed using slow-acting compressor, here after this condition will be regarded as 'slow-acting compression' and (iii) original speech which will regarded as 'original' in following section.

Gaussian nature of the data was assessed using Shapiro-Wilk's test for normality and the results revealed that SNR loss for all the three conditions are not normally distributed. Fast-acting compression ($W=0.85$, $p=0.01$), Slow-acting compression ($W=0.87$, $p=0.02$) and original ($W=0.74$, $p<0.001$). Since, data from all the three conditions are not normally distributed the non parametric Friedman's test was used to investigate the main effect of compression on SNR loss. Friedman's test revealed that compression had significant main effect ($X^2(2) = 6.89$, $p = 0.03$) on SNR loss.

Pair wise comparisons between the three conditions were performed using Wilcoxon signed rank test. Bonferroni's adjustments were made for each pair wise comparisons to account for the multiple comparisons. Results of the Wilcoxon signed rank test was considered to be significant when, $p<0.016$ as the significance level was adjusted for Bonferroni's correction factor. Pair wise comparison revealed that SNR loss for original signal was significantly different ($Z=-2.99$, $p=0.001$) from the SNR loss for fast-acting compression. SNR loss for original signal was lower when compared to fast-acting compression (see table 1). Even though median SNR loss for slow-acting compression was lower than the SNR loss for fast-acting compression (see table 1), the difference was not statistically significant (Z=-0.98, p=0.34). Similarly, median SNR loss for original signal was lower than the SNR loss for slow-acting compression (see table 1) but, statistically the difference was not significant (Z=-1.87, p=0.062).

Table.1

Median, range and inter quartile range for SNR loss in original, slow-acting compression and fast-acting compression conditions.

Conditions	Median (dB)	Range (dB) (Min-Max)	Inter quartile range (dB) (Q1-Q3)
Original	4.67	0.67-23.67	3.67-8.1 7
Slow acting compression	5.67	1.67-22.67	3.67-13.67
Fast acting compression	8.67	2.67-28.67	4.67-13.17

To investigate the possible association between the types of compression and SNR loss Spearman's rank correlation analysis was performed. Results of the Spearman correlation was considered to be significant when p<0.016 due to multiple correlations. Correlation analysis revealed that there is no association (r_s=-0.33, p=0.097) between spectral ripple discrimination threshold & SNR loss for original speech, which means that spectral fine structure sensitivity or frequency selectivity did not play a major role in perception of original speech in the presence of noise.

There was a negative correlation observed between spectral ripple discrimination threshold & SNR loss for fast compression (r_s=-0.54, p= 0.013) as well as between spectral ripple discrimination threshold & SNR loss for slow-acting compression (r_s= -

0.69, p= 0.001). As the spectral ripple discrimination threshold increases SNR loss decreases for compressed speech. In other words, if the frequency selectivity or spectral fine structure sensitivity is better, SNR loss will be smaller. Statistically significant correlation suggests that, spectral fine structure sensitivity had played a role in perception of speech in the presence of noise under compressed conditions.

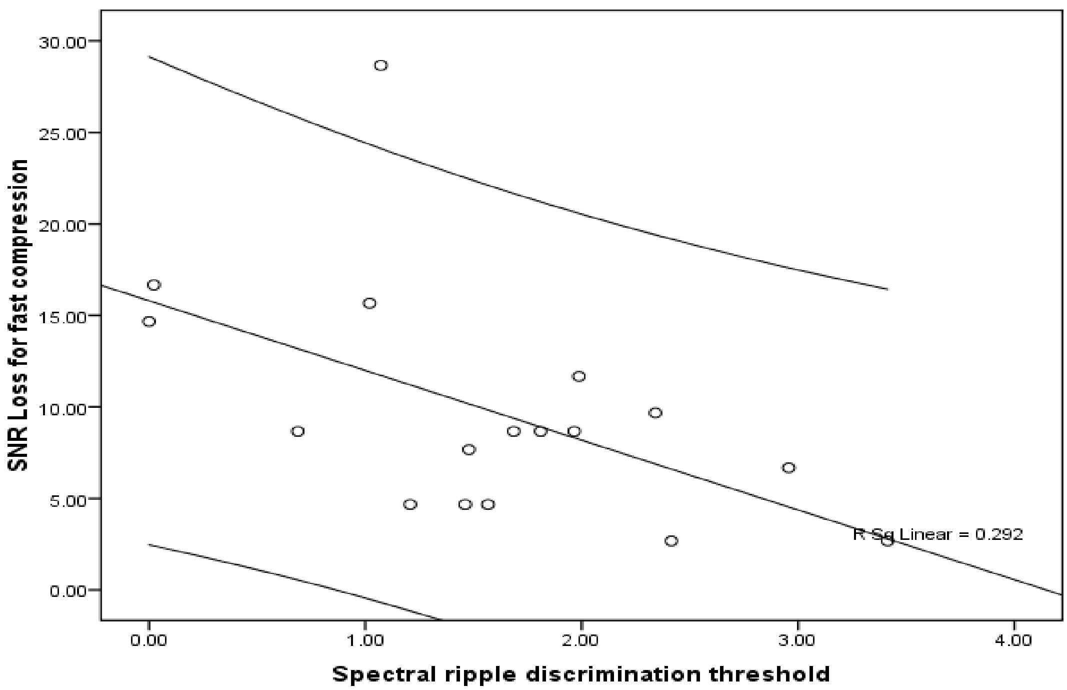

Figure 5. Scatter plots showing a linear relationship between spectral ripple discrimination threshold and SNR loss for fast-acting compression conditions.

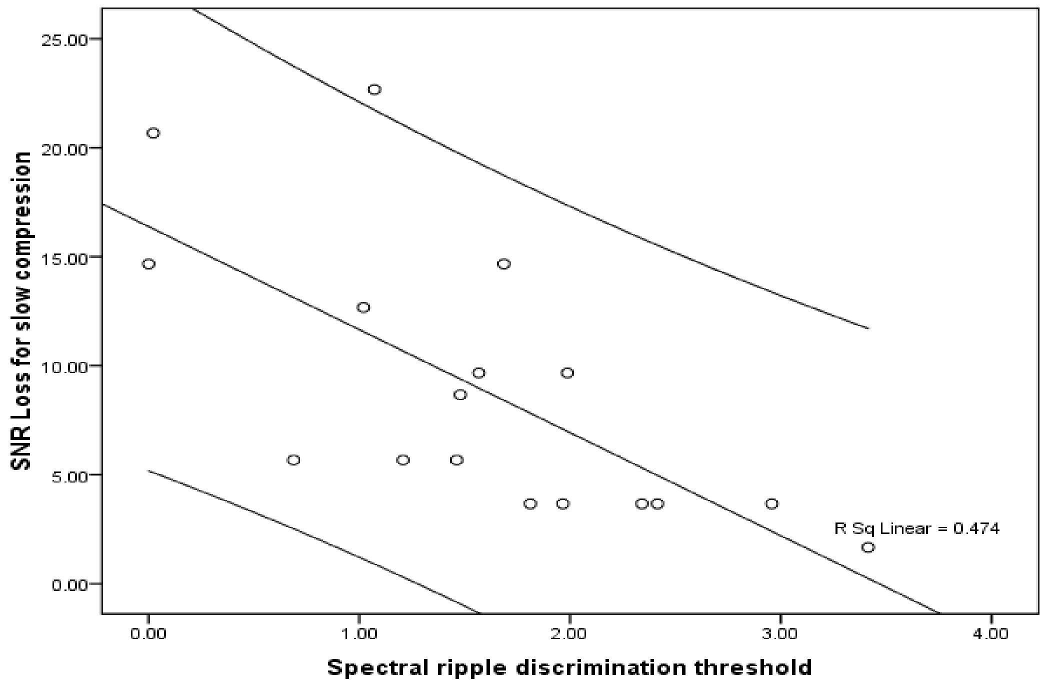

Figure 6. Scatter plots showing a linear relationship between spectral ripple discrimination threshold and SNR loss for slow-acting compression conditions.

Linear regression analysis was performed to investigate whether SNR loss can be predicted from spectral ripple discrimination threshold. Linear model well suited to describe the relationship between spectral ripple discrimination threshold & SNR loss for fast-acting compression ($F(1,15)=6.19$, $p<0.05$). Similarly, regression analysis revealed that SNR loss for slow-acting compression can be predicted from spectral ripple discrimination threshold using linear model ($F(1,15)=13.50$, $p<0.05$). Regression equations are as follows:

$$\text{SNR loss } fc = 15.80-3.81*Srdt$$

$$\text{SNR loss } sc = 16.38-4.72*Srdt$$

In the above equations '*fc*' stands for fast-acting compression, '*sc*' stands for slow-acting compression and '*Srdt'* stands for spectral ripple discrimination threshold.

Observation of R^2values suggests that 47% variance in SNR loss for slow-acting compression could be attributed to the variability in spectral ripple discrimination threshold. Similarly, 29% variance in SNR loss for fast-acting compression could be attributed to the variability in spectral ripple discrimination threshold.

Chapter 5

DISCUSSIONS

Spectral ripple discrimination thresholds revealed that individuals with cochlear hearing loss required less ripple density to perceive the phase reversal when compared to individuals with normal hearing. Mean spectral ripple discrimination threshold for normal hearing individuals is 3.5 ripples / octave and mean spectral ripple discrimination threshold for hearing impaired individuals is 1.6 ripples / octave. This result suggests that frequency resolving ability of individual with cochlear hearing loss is worse than normal hearing individuals. In the phase reversal task both the stimuli have same spectral band but opposite positions of spectral maxima and minima on the frequency scale. The phase reversal effect will be detected only if the rippled structure of the spectrum can be resolved. If the fine structure of the spectrum is unresolved phase reversal cannot be detected (Supin, Popov, Milekhina & Tarakanov, 2002). As the ripple density increases position between the maxima and minima decreases hence, becoming irresolvable at cochlear level. Since cochlear hearing loss results in broadening of auditory filter (Tyler et al., 1984; Glasberg & Moore, 1986; Dubno & Dirks, 1989; Larocheet al., 1992; Peters & Moore, 1992; Stone, Glasberg & Moore, 1992; Leek & Summers, 1993; Sommers & Humes, 1993; Leeuw & Dreschler, 1994). They require less ripple density to perceive the phase reversal. Similarly, Henry, Turner & Behren (2005) also reported that spectral ripple thresholds were poor in hearing impaired individuals when compared to normal hearing listeners. Normal hearing individuals obtained threshold of 4.84 ripples / octave and hearing impaired individuals obtained threshold of 1.77 ripples / octave. The values obtained in the current study are slightly worse when compared to results of Henry, Turner & Behren (2005). This might be due to the age effect; participants in the current study are slightly older than the previous study. Other reason could be technique used to generate spectral ripples. Current study used the spectral ripples which were sinusoidal in decibel amplitude space, whereas the former study used the spectral ripples which were sinusoidal in a linear amplitude space.

SNR loss in hearing impaired individual was measured for following three conditions; original speech, fast-acting compression and slow-acting compression. For original speech, hearing impaired subjects required 4.67dB (median) more SNR than normal hearing subjects. This finding confirms previous several other studies (Plomp, 1978, 1986; Dreschle & Plomp, 1980, 1985; Humes, Dirks & Kincaid, 1987; Zurek & Delhorne, 1987; Lee &Humes, 1993; Glasberg & Moore, 1989) that individual with cochlear hearing loss perform poor in the presence of background noise. Spectral differences especially the difference in F_0 help the individual to perceptually segregate the target speech and competing speech maskers. When the individual is unable to utilize the spectral differences between target speech and competing speech, he/she may not form separate perceptual streams for target speech and masker (Oxenham, 2008). Poor spectral ripple perception by the participants of the study indicates that, they had poor spectral resolution which would have disabled them from utilizing the spectral difference between the target and masker.

SNR loss for compressed speech was greater than for original speech. Median SNR loss for slow-acting compression was 5.67 dB and for fast-acting compression is

8.67 dB indicating, worst performance with fast-acting compression among the three conditions. In the present study, it was found that SNR loss for slow-acting compression was less, indicating good speech intelligibility compared to fast-acting compression. Poor performance with compression could be attributed to the reduced dip listening ability. Listener ability to take advantage of dips in the background sound when trying to understand a target signal is denoted as dip listening (Gatehouse, Naylor, & Elberling, 2003). Dip listening is important in situations where communication takes place in the presence of modulated background noise, like current study where multi-talker babble was used. Use of compression reduces the temporal contrast or modulations thus resulting poor dip listening (Stone & Moore, 2004, 2008). It reduces intensity contrasts and the modulation depth of speech, which may have an adverse effect on the perception of certain speech cues, especially when high compression ratios are, used (Plomp, 1988).

Recent evidence suggests that the benefit obtained from listening in the dips may be related to the ability to process the temporal fine structure of sounds. Changes in the TFS during dips in the background help the listener to determine that target speech is present and to determine what the properties of the target speech are (Moore, 2008). Difference in TFS cues enables the listener to form separate perceptual streams for target speech and competing speech (Nie, Stickney & Zeng, 2005). There is evidence that moderate cochlear hearing loss reduces or abolishes the ability to process TFS (Hopkins & Moore, 2007; Moore, Glasberg, & Hopkins, 2006). Most important information that TFS carries is harmonicity of the signal (Moore, Glasberg & Hopkins, 2006). Perception of harmonics are important for perception of pitch and thus for source segregation

(Oxenham, 2008). Auditory frequency selectivity and the resolvability of harmonics can predict pitch discrimination, suggesting that peripheral filtering is important for pitch coding (Bernstein & Oxenham, 2006). Resolvability of harmonics in hearing impaired participants would have been affected by their poor frequency selectivity, thus resulting in poor speech perception in the presence of noise.

Better performance under slow-acting compression could be attributed to the fact that, the deleterious effect on temporal envelope is minimal. This finding was in accordance with a study done by Drullman, Festen, & Plomp (1994), where they reported limited distortion in temporal envelope and also the envelope fluctuations at syllabic rates are preserved when using slow-acting compression systems. Thus may be important for maintaining speech intelligibility. Poor performance of fast-acting compression could be attributed to the fact that, it induces greater distortion in temporal envelope while preserving some amount of fine structure (FS) information. This was confirmed by previous studies which suggested that fast-acting compression reduces the temporal contrast to greater extent than slow-acting compression resulting in impaired speech perception (Plomp, 1994; Noordhoek & Drullman, 1997). It was also reported that it could introduce spurious changes in the shape of the temporal envelope of sounds (e.g., overshoot and undershoot effects; Stone & Moore, 2008). Therefore, it can be speculated that, in individuals with moderate cochlear hearing loss the ability to use fine structure information is reduced and hence they rely more on information carried in the temporal envelope of speech signal. A recent study by Moore, Glasberg & Hopkins, (2006) confirms reduced ability to process FS information in individuals with moderate cochlear hearing loss. Hence, in the current study SNR loss for slow-acting compression is relatively better than fast-acting compression. The role of FS information in speech perception remains somewhat controversial. While envelope information in a few frequency bands appears sufficient to give reasonably high intelligibility for speech presented in quiet (Shannon et al, 1995), the perception of speech in noise, especially modulated noise, seems to depend at least partly on the use of FS information (Lorenzi et al, 2006a; Gilbert et al, 2007; Hopkins et al, 2008; Hopkins & Moore, 2008; Lorenzi & Moore, 2008).

Results of the study revealed that spectral ripple discrimination method can be reliably used to study individual's frequency resolution ability and be used to predict SNR loss. Individual with good spectral ripple threshold can be fitted with fast-acting compression since they can utilize spectral differences to differentiate speech and noise. But individual with poor spectral ripple threshold slow-acting compression may be preferred as they may have to rely more on information provided by the envelope rather than the spectral fine structure. Rippled noise has been used to estimate frequency selectivity in neurophysiological single-unit studies (Wilson and Evans, 1971; Evans and Wilson, 1973; Bilsen et al., 1975; Evans, 1977), in psychophysical studies using a masking paradigm (Houtgast, 1974, 1977; Pick et al., 1977; Pick, 19801, and in studies of pitch perception of complex sounds (Yost et al., 1977; Yost and Hill, 1979; Bilsen and Wieman, 1980; Yost, 1982). However, to study the frequency resolving power (FRP) dependence on frequency, measurements can be made using narrow-band noises of various central frequencies. Thus, FRP data can be used to derive the auditory filter bandwidth. According to Supin, Popov, Milekhina & Tarakanov, (1994), equivalent rectangular bandwidth (ERB) is given by the simple expression:

$$ERB \approx 0.71 \times F0/D$$

where the ERB is given in Hz, ‘F0’is the central frequency, kHz and ‘D’ is the rippled density expressed in number of ripples per kHz, which determines the limit of resolvable ripple density.

Chapter 6

SUMMARY AND CONCLUSIONS

Human speech is highly redundant with spectral and temporal cues. Speech signals contain two forms of information, envelope & temporal fine structure (TFS). Envelope cues (also called as amplitude modulations) correspond to the slow amplitude variations that rate below 50 Hz and fine structure cues correspond to rapid frequency fluctuations that rate above 250 Hz (Rosen, 1992). Recent research evidences suggest that cochlear hearing loss adversely affects the ability to use TFS information for speech perception (Qin & Oxenham, 2003; Stickney et al, 2005; Zeng et al, 2005; Lorenzi et al, 2006; Hopkins et al, 2008; Gnansia et al, 2008; Lorenzi & Moore, 2008). This seems likely to be one factor that contributes to the difficulty experienced by cochlear hearing loss individuals when trying to understand speech in the presence of background especially when the noise is also modulated (Festen & Plomp, 1990; Peters et al, 1998; Hopkins et al, 2008).

Moore (2008) established that, the ability to process TFS may have implications for the choice of compression speed in hearing aids. Hearing-impaired individual who has little or no ability to process temporal fine structure (TFS) information, slow-acting compression might be more effective than fast-acting compression. However, in individuals who restore some ability to process TFS, the situation is different. Fast-acting multi-channel compression can help to restore the audibility of low-level portions of signals (the dips), and information derived from TFS can be used to extract ‘glimpses’ of the target speech during dips in a background sound. Perception of temporal fine structure largely depends on frequency resolution ability of the auditory system.

The present study determined the extent to which individuals with normal hearing and individuals with cochlear hearing loss can use temporal fine structure information, by measuring the highest ripple density at which the fine spectrum structure is resolvable using noise with a rippled spectrum (rippled noise), i.e., a spectrum containing periodically alternating spectral power maxima and minima within a given spectral band. The ripple spacing was varied adaptively from 1.000 to 11.31 ripples / octave, and the minimum ripple spacing at which a reversal in peak and trough positions could be detected was determined as the spectral ripple discrimination threshold for that listener. The study also aimed to correlate the spectral discrimination threshold with speech identification abilities measured in terms of SNR loss using QuickSIN protocol (Killion, 1997) with and without compression by individuals with cochlear hearing loss

A total of 20 participants were involved in the study. The participants were age matched and divided into two main groups: Control group having normal hearing individuals (N=15 ears) and experimental group having individuals with mild to moderate cochlear hearing loss (N=17 ears).

The results obtained in the current study are summarized as follows:

1. Spectral ripple discrimination thresholds were significantly better in normal hearing individuals (Mean = 3.5 ripples / octave) when compared to individuals with cochlear hearing loss (Mean = 1.6 ripples / octave).

2. SNR loss in hearing impaired individual was measured for following three conditions; original speech, fast-acting compression and slow-acting compression. For original speech, hearing impaired subjects required 4.67 dB (median) more SNR than normal hearing subjects.
3. As the spectral ripple discrimination threshold increases SNR loss decreases for compressed speech. In other words, if the frequency selectivity or spectral fine structure sensitivity is better, SNR loss will be smaller. Statistically significant correlation suggests that, spectral fine structure sensitivity had played a role in perception of speech in the presence of noise under compressed conditions.
4. SNR loss for compressed speech was greater than for original speech. Median SNR loss for slow-acting compression was 5.67 dB and for fast-acting compression is 8.67 dB indicating, worst performance with fast-acting compression among the three conditions

Overall, it can be concluded that normal hearing individuals perform spectral ripple discrimination task using TFS cues implying the superior ability to process TFS information. On the other hand, individuals with cochlear hearing loss show difficulty in processing TFS information. Therefore, it can be concluded that hearing loss significantly reduces the ability to analyze and utilize TFS cues to perform spectral ripple discrimination task. Hence, individuals with cochlear hearing loss rely more on temporal envelope cues rather than TFS for understanding speech. Fast-acting compression induces greater distortions in the temporal envelope and preserves TFS information which results in significant difficulty in speech perception compared to slow-acting compression.

In summary, the results demonstrate a dramatic loss of the ability of hearing-impaired subjects to use TFS cues for speech perception. The conclusion from all this is that measures of the ability to use TFS information might be useful in determining the most appropriate speed of compression for a hearing-impaired individual. It is possible that the ability to process TFS is related in a more general way to the speed and accuracy of neural processing in the brain. If this were the case, the ability to process TFS could be related to cognitive abilities. This might explain the link between cognitive abilities and the benefit of fast-acting compression for listening in the dips (Gatehouse, Naylor, & Elberling, 2003).

Clinical implications

1. Spectral ripple threshold can be used to predict speech perception difficulties by cochlear impaired individuals.
2. Spectral ripple threshold has an implication for the choice of compression speed in hearing aids (Moore, 2008)
3. The ability utilize fine structure information helps to improve spectral peak resolution for hearing impaired and cochlear implant users which may lead to improved speech recognition (Henry, Turner & Behrens, 2005).

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