Less is more?
Loudness aspects of prescriptive methods for nonlinear hearing aids

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Abstract

In Sweden, about 10% of the adult population experiences hearing problems that cause them difficulties in everyday communication, and approximately 60,000 people are provided with hearing aids each year. Despite the fact that modern hearing aids can facilitate speech communication in a wide range of listening environments, many hearing-aid users are dissatisfied with their hearing aids. It is likely that the clinical methods used for individual fitting of the hearing aids are not optimal.

The current study investigates prescriptive methods for nonlinear, wide dynamic range compression (WDRC) hearing instruments. The goal is to draw general conclusions about the preferences of hearing aid users. Therefore, the prescriptions are evaluated using well-established models of loudness and speech intelligibility.

Current methods differed considerably in prescribed gain. Evaluations in a laboratory test, with 20 hearing-impaired listeners, showed that these differences led to large differences in perceived and calculated loudness, but only to minor differences in measured and predicted speech recognition scores.

The difference in loudness was explored in a study where 21 first-time hearing-aid users compared two prescriptions. One method led to normal and the other to less-than-normal overall calculated loudness (according to the loudness model of Moore and Glasberg (1997)). The prescription that led to less-than-normal overall loudness was clearly preferred in field and in laboratory tests.

Preferred overall loudness was then quantified. Hearing-impaired participants with mild to moderate hearing loss preferred considerably less-than-normal overall calculated loudness in both field and laboratory tests. There were no significant differences between inexperienced and experienced hearing aid users. Normal-hearing participants, on the other hand, preferred close-to-normal overall calculated loudness. In addition, a potential problem with the loudness model was encountered: despite the fact that the hearing-impaired listeners were provided with less than normal overall calculated loudness, they rated loudness higher than the normal-hearing listeners.

The results refute the most commonly adopted rationale for prescriptive methods for WDRC hearing aids—that overall loudness should be restored to normal. Hearing-impaired listeners with mild to moderate hearing loss preferred considerably less than normal overall loudness. This should be taken into account when deriving new prescriptive methods, and when providing clients with hearing aids.

Key words: hearing impairment, hearing aid, nonlinear, WDRC, hearing aid experience, prescription, loudness, loudness model, speech intelligibility, preference.
Cirka 10% av befolkningen i Sverige uppgör att de har en hördevirksamhet av en sådan grad att de har problem att deltaga i vanliga samtal med flera talare. ungefär 60 000 personer får varje år hörapparater utprovade. Trots att studier visat att moderna hörapparater kan underlätta kommunikation, också i svåra lyssnings-situationer, tycker många hörapparatanvändare att deras hörapparater fungerar dåligt. De metoder som används vid hörapparatutprovningen är förmodligen inte tillräckligt bra.

Studien undersöker preskriptiva metoder för utprovning av olinjära hörapparater. Målet är att dra generella slutsatser om hörapparatanvänndes föredragna förstärkning. Därför används vid utvärdering av metoderna etablerade modeller för hörstyrka och taluppfattning.

Modern hörapparater visade sig föreskriva mycket olika förstärkning för en och samma hördevirksamhet. När preskriptionsmetoderna utvärderades i ett laboratorieförsöck med 20 hörskadade försökspersoner kunde man se att skillnaderna i föreskriven förstärkning resulterade i stora skillnader i uppmätt och beräknad hörstyrka, men bara i små skillnader i uppmätt och skattad taluppfattning.

Skillnaderna i hörstyrka undersöcktes i en annan studie där 21 hörskadade personer utan tidigare hörapparativna jämförde två preskriptionsmetoder, en som ledde till normal total hörstyrka och en annan som ledde till lägre än normal hörstyrka (beräknat med en hörstyrkemodell av Moore och Glasberg (1997)). Metoden som gav lägre än normal hörstyrka föredrogs såväl i ett fältförsök som i efterföljande laboratorieförsök.


Included papers

The thesis is based on the following papers:


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Contents

Contents ix

Acronyms xi

1 Background 1

2 Introduction 3
  2.1 The Ear ............................................. 3
  2.2 The Nonlinear Cochlea .............................. 4
  2.3 Cochlear Hearing Loss .............................. 7
  2.4 Loudness Models .................................... 9
  2.5 Speech Intelligibility Index ....................... 11
  2.6 Hearing Aids ....................................... 12
  2.7 Hearing Aid Fitting ............................... 14

3 Aims and Outline of the Current Work 19

4 Prescriptive Methods—Same or Different? (Paper I) 21
  4.1 Methods ........................................... 21
  4.2 Results ........................................... 22

5 Laboratory Comparison between Prescriptive Methods (Paper II) 23
  5.1 Methods ........................................... 23
  5.2 Results ........................................... 24

6 Do First-Time Hearing Aid Users Prefer Normal Overall Loudness? (Paper III) 27
  6.1 Methods ........................................... 28
  6.2 Results ........................................... 28


7 Preferred Overall Loudness (Papers IV and V) 31
  7.1 Methods .................................................. 31
  7.2 Results .................................................. 32

8 Discussion 35
  8.1 Paper I .................................................... 35
  8.2 Paper II ................................................... 36
  8.3 Paper III .................................................. 36
  8.4 Papers IV and V ......................................... 37
  8.5 Hearing Aid Experience ............................... 38
  8.6 Prescriptions for Narrowband and Broadband Sounds . . . 40

9 Conclusions 43

10 About the Papers 45

11 References 47

Included papers


Acronyms

AGC automatic gain control
BTE behind-the-ear (hearing aid)
CAMEQ Cambridge fitting method, equalising loudness density
CAMREST Cambridge fitting method, resoring loudness density to normal
CR compression ratio
CT compression threshold
DSLi/o desired sensation level (input/output), a prescriptive method for nonlinear hearing aids
FIG6 a prescriptive method for nonlinear hearing aids
HA hearing aid
HI hearing impairment, hearing-impaired
HI Exp hearing-impaired listener with hearing aid experience
HI Inexp hearing-impaired listener without hearing aid experience
HL hearing loss; when the acronym is used in the combination dB HL, it means hearing level, i.e., “relative to normal thresholds”
HRTF head-related transfer function
HTL hearing threshold level
ICRA International Collegium for Rehabilitative Audiology, the acronym is used for a particular speech-shaped noise developed within the collegium
IHC inner hair cell
ITE in-the-ear (hearing aid)
LDF level distortion factor
LDL loudness discomfort level
NAL-NL1 National Acoustic Laboratories – nonlinear 1, a prescriptive method for nonlinear hearing aids
NAL-RP National Acoustical Laboratories – revised, profound, a prescriptive method for linear hearing aids
NH normal hearing
OHC outer hair cell
REAG real ear aided gain
REIG real ear insertion gain
REUG real ear unaided gain
RMS root-mean-square
SHARP stereo hearing aid research processor, hearing aid used in the study reported in Paper V
SII speech intelligibility index
SNR signal-to-noise ratio
SPL sound pressure level; when the acronym is used in the combination dB SPL, it means “relative to 20μPa”
WDRC wide dynamic range compression
Chapter 1

Background

Hearing loss is a common disorder. About 10% of the adult population in Sweden experiences hearing problems that cause them difficulties in everyday communication. For the elderly population (75–84 years), the proportion of people that report hearing problems is about 30% (Rosenhall, Jönsson, & Söderlind, 1999).

The term *aural rehabilitation* refers to a number of rehabilitative efforts that attempt to alleviate the hearing-impaired person’s disability. A wide range of professionals address this issue, including audiologists, engineers, psychologist, and physicians. One important component of aural rehabilitation is hearing aid fitting. In Sweden year 2002, approximately 58,000 people were provided with hearing aids. About 55% of these were first-time hearing aid users. In total, approximately 77,000 hearing aids were fitted (Arlinger et al., 2003). The hearing aid fitting process is summarised in Figure 1 (Smells & Leijon, 2000).

The process starts when the hearing-impaired person’s needs, prerequisites, and expectations are evaluated. Appropriate hearing aids are selected based on available information. This selection includes choices such as monaural or binaural hearing aids, in-the-ear or behind-the-ear hearing aids, linear or nonlinear signal processing, and a choice of various accessories. Sometimes various demands are conflicting, and it is important to find a compromise that the hearing aid user can accept.

In the next step of the process, the selected hearing aids are adjusted to suit the hearing aid user. A prescriptive method is used. The prescription is a formula by which the audiologist can find a suitable gain-frequency response based on the hearing-impaired person’s hearing characteristics and the listening environments that are particularly important to the user. The hearing aid is adjusted to the prescribed gain, usually by measuring the actual hearing aid gain in the user’s ear canal.

The hearing aids are then used during a training period of usually a
1. Background

Figure 1: The hearing aid fitting process. The dashed lines indicate how information gained in later phases in the process can lead to modifications of earlier decisions. (From Smeds & Leijon (2000) with permission from the authors and CA Tegnér AB.)

couple of weeks. The hearing aid user’s experiences during this training period are evaluated together with the audiologist. Subjective and functional evaluations are used to fine-tune the hearing aids according to the user’s demands. A number of visits to the audiologist may be necessary.

Large consumer surveys have shown that about 35% of the hearing aid users are dissatisfied with their hearing aids (Kochkin, 1996). The most common complaint is that the hearing aids do not work well in noisy environments, where the hearing aid user usually needs them the most. Studies have shown that modern hearing instruments can facilitate speech communication in noisy environments (e.g., Larson et al., 2000). Why, then, are so many hearing aid users dissatisfied with their hearing aids?

It is likely that the clinical methods used for individual fitting of hearing aids are not optimal. This thesis evaluates existing prescriptive methods for nonlinear hearing aids and explores in depth the perhaps most basic rationale underlying prescriptions – the idea that overall loudness should be restored to normal.
Chapter 2

Introduction

2.1 The Ear

The auditory system is conceptually divided into one peripheral and one central part. The peripheral part consists of the outer, middle and inner ear. The outer ear consists of the pinna and the ear canal (Figure 2). The eardrum divides the outer from the middle ear. The ear canal and the eardrum act as a resonator for sounds with frequencies around 2.7 kHz; together with the rest of the outer ear, a broad resonance peak for frequencies between 2 and 4 kHz is created. Incoming sound travels through the ear canal to the eardrum, which is set in motion. The vibration created is

![Diagram of the ear](image)

Figure 2: The peripheral auditory system. Redrawn from information material from Orion Pharma AB, Sweden, with permission.
transmitted through three small bones in the middle ear to the membrane-covered oval window of the inner ear. The middle ear acts as an impedance transformer, mainly due to the large area difference between the eardrum and the oval window. Without this impedance transformation, less than 1% of the incoming sound energy would be transmitted into the inner ear. The transmission through the middle ear is described by a transfer function with a broad peak for frequencies between 300 and 4000 Hz.

The inner ear contains both the organ of hearing (organ of Corti) and the organ of balance. The organ of Corti is situated within the cochlea, a fluid-filled, snail-shaped part of the inner ear (Figure 3). Three chambers run along the cochlea, from the base to the apex. The basilar membrane divides two of these three chambers. The organ of Corti is located on the basilar membrane and is covered by the tectorial membrane. Stimulation of the cochlea results in a travelling wave that moves from the base of the cochlea to the apex. Pressure differences between the chambers displace the basilar membrane. The relative motion between the basilar and the tectorial membranes causes a bending of the hairs on the hair cells within the organ of Corti. These deflections affect transduction channels in the hair cells and allow an ion current to pass through the hair cells. A receptor potential is generated, the nerve fibres that innervate the hair cells are activated, and signals are sent to the central auditory system (e.g., Gelfand, 1998; Pickles, 1988).

2.2 The Nonlinear Cochlea

High sensitivity and good frequency selectivity are characteristics of a well-functioning auditory system. It has long been known that the basilar membrane, its surrounding structures, and the cochlear fluids together form a mechanical frequency analyser, which can separate the frequency components of a complex sound (e.g., Bekesy, 1949, 1989). The distance that the
travelling wave passes before it reaches its maximum amplitude will depend on the frequency of the stimulus. Low-frequency sounds show a maximum close to the apex of the cochlea, whereas high-frequency sounds show a maximum close to the base. However, this mechanical and passive process alone cannot explain our ability to detect very faint sounds nor our ability to separate the components of a complex sound with very good precision.

Kemp (1978a; 1978b) showed that the inner ear was capable of not only detecting, but also producing sound. He presented acoustic stimuli, and could measure emissions of acoustic energy in the ear canal in response to the stimuli. These findings point to some active processes within the cochlea. The acoustic emissions were shown to be biologically vulnerable, and he suggested that the hair cells in the cochlea were their origin.

Two types of hair cells are found in the cochlea: inner and outer hair cells. Afferent nerve fibres, which carry information from the cochlea to the central auditory system, mainly innervate the inner hair cells. The inner hair cells are the main sensory cells. Efferent nerve fibres, which carry information from the central auditory system to the cochlea, mainly innervate the outer hair cells. The organisation of the efferent system suggests that the outer hair cells can be selectively controlled by the central auditory system, and the outer hair cells play an important role for our ability to process sound with high sensitivity and good frequency resolution. They can change their shape in response to acoustic or electric stimuli. This produces a feedback force that interacts with the mechanics of the cochlea,

Figure 4: Basilar membrane vibration velocity as a function of input sound pressure level. The solid line shows the input-output function for a healthy cochlea. The dashed line shows the input-output function for a cochlea with a total loss of outer hair cells. Sketch redrawn from Ruggero & Rich (1991) (the drawing from Smeds & Leijon (2000) with permission from the authors and CA Tegnér AB).
thereby enhancing the cochlear response. This enhancement is only seen at relatively low input levels. When the outer hair cells are damaged, the basilar membrane motion in response to low-level sounds is reduced, and the frequency resolution impaired (e.g., Ulfendahl, 1997).

A schematic picture of how the vibration velocity of the basilar membrane varies as a function of stimulus level is shown in Figure 4 (Ruggero & Rich, 1991). The solid line shows the characteristics for a healthy cochlea. A compressive nonlinear relationship between input level and basilar membrane vibration velocity is seen. This nonlinear relationship is only present when the measurements are made at the point along the basilar membrane that corresponds to the frequency of the measurement stimulus, the so-called characteristic frequency. The dashed line shows the characteristics for a cochlea with a total loss of outer hair cells. The relationship between input level and basilar membrane vibration is now linear.

Neural frequency selectivity curves are shown schematically in Figure 5 (Sellick, Patuzzi, & Johnstone, 1982). The figure shows the sound pressure level that is needed to produce a constant neural response, the “threshold”, as a function of stimulus frequency. A healthy cochlea (solid line) is characterised by high sensitivity (low stimulus levels are enough to produce the given response) and high frequency selectivity at the characteristic frequency. For a cochlea with a large amount of outer hair cell loss (dashed line), both the sensitivity and selectivity are reduced.

![Figure 5: Neural selectivity depicted schematically. The sound pressure level needed to produce a constant neural response as a function of stimulus frequency is shown. The solid line shows the behaviour for a healthy cochlea. The dashed line represents a cochlea with a large amount of outer hair cell loss. Redrawn from Sellick et al. (1982) (the drawing from Smeds & Leijon (2000) with permission from authors and CA Tegnér AB).](image)
2.3 Cochlear Hearing Loss

There are different types of hearing losses. Cochlear hearing loss is the most common type, and the main reason for cochlear hearing loss is damage or loss of hair cells. The outer hair cells are most easily damaged (Borg, Canlon, & Engström, 1995), and the damage leads to a loss of the compressive nonlinearity described above and problems both at hearing threshold levels and at supra-threshold levels. Sometimes hearing losses caused by damage to the cochlea or the auditory nerve are classified as sensorineural hearing losses. It is, however, much more common that the cause of the hearing loss is cochlear (sensory) rather than neural.

A blocking of the transmission through the outer or middle ear causes a conductive hearing loss. This results in a frequency-dependent attenuation of the incoming signal. If the cochlea functions normally, the effect on, for example, speech recognition is not as severe for a conductive hearing loss as for a cochlear hearing loss.

Audibility

Most people with hearing impairment find their reduced ability to recognize speech, particularly in background noise, the most disturbing. The fact that the hearing loss reduces the audibility of the speech sounds is of course a reason for poor speech recognition abilities. For a person with a mild hearing loss (up to about 45 dB HL), Moore (1996) argues that audibility is the single most important factor that determines a hearing-impaired listener’s ability to understand speech. For greater hearing losses, deficits in supra-threshold perception reduce the speech recognition ability further. Deficits in supra-threshold perception include abnormal loudness perception, and reduced selectivity for frequency, intensity, and temporal characteristics (Moore, 1996).

The term hearing threshold refers to the lowest level at which a person can detect a sound. Absolute thresholds are determined when no background noise is present. For normal-hearing persons, the absolute threshold of hearing varies as a function of stimulus frequency. In the hearing clinic, an audiogram is used to describe some aspects of the hearing loss. In this type of graph, sound pressure levels are presented relative to the hearing thresholds for young, normal-hearing persons, represented as the horizontal line at 0 dB HL. Figure 6 shows a typical audiogram for a person with a cochlear hearing loss. This particular audiogram shows the median hearing thresholds for 649 first-time hearing aid users seen at a Swedish clinic 1998 (Holst, 2002).

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1The acronym HL used in this context means hearing level and dB HL should be interpreted as “dB relative to normal thresholds.”
2. Introduction

Figure 6: A typical audiogram for a person with a mild to moderate cochlear hearing loss. Absolute hearing thresholds (×) and loudness discomfort levels (✓) are presented. The hearing threshold data are the median values for 649 first-time hearing aid users seen at a Swedish clinic 1998 (Holst, 2002).

Loudness Perception

*Loudness* is the magnitude of the auditory sensation. The measure is most closely correlated to the intensity of a sound. However, loudness depends not only on the intensity of a sound, but also on its frequency, spectral contents, and duration. Loudness can be measured in various ways. In loudness-scaling experiments, the participant can use labelled categories (e.g., “too quiet”, “too loud”), numbers, or some other way to describe how loud they experience a sound. In a loudness-balancing task, two sounds are compared and adjusted to be equally loud. Paired comparisons, where the participant judges which one of two stimuli is louder, can be used to rank order sounds with regard to their loudness.

The *auditory dynamic range*, the range between the hearing threshold and the loudness discomfort level (LDL, the sound pressure level at which a sound becomes uncomfortably loud) is usually reduced for people with cochlear hearing loss. This is depicted in Figure 6. Normal hearing thresholds are 0 dB HL, and normal LDLS around 100 dB HL. The hearing loss represented in Figure 6 increases with frequency, whereas the LDLS are close to normal and the variation across frequency is small. The auditory dynamic range is reduced compared to a normal-hearing listener’s dynamic range, and more so at the high frequencies than at the low.

This phenomenon, that the loudness perception seems to grow faster (from threshold to discomfort level) for the hearing-impaired person than for a normal-hearing person, is called *recruitment of loudness*, and it can be explained by a loss of the compressive nonlinearity produced by healthy outer hair cells. Without the active process supplied by the outer hair cells, the hearing-impaired person will have difficulties detecting soft input
signals. If the inner hair cells are intact, the perceived loudness at high input levels can be the same as the loudness perceived by a normal-hearing person. Loudness recruitment has implications for the design of hearing aids and hearing aid fitting methods. In order to make low-level sounds audible without making high-level sounds intolerably loud, a hearing aid needs to provide more gain for low-level sounds than for high-level sounds.

Frequency Selectivity

Frequency selectivity, our ability to separate components in a complex sound signal, is crucial for many auditory tasks. The filtering in the cochlea is the key to understanding this phenomenon. The activity at one point along the basilar membrane is affected by all sound components within a frequency band centred at the characteristic frequency at that point. Many results from psycho-acoustical experiments can be explained by a model where sound is analysed in a number of overlapping band-pass filters, one filter for each critical band centre frequency (Fletcher, 1940). The frequency selectivity can be described by the equivalent rectangular bandwidth (ERB) of these band-pass filters (theory summarised by e.g. Moore, 1996)

Outer hair cell loss results in auditory filters that are broader than normal. As a consequence, the frequency selectivity is reduced. This, in turn, results in increased problems with masking. Thus, a person with a cochlear hearing loss is more troubled by background noise than a normal-hearing person is. Today there is little knowledge about how to compensate for reduced frequency selectivity.

2.4 Loudness Models

Loudness cannot be directly measured with, for instance, a sound level meter, and many attempts to calculate loudness based on sound intensity, frequency, spectral content, and duration have been made. Stevens (1957) suggested that loudness is a power function of the intensity of the stimulus

\[ N = kI^{0.3} \]

where \( N \) is the loudness of the stimulus, \( I \) the intensity of the stimulus, and \( k \) a constant depending on the participant and the units used. The power function reflects the compressive characteristics of the cochlea. The relation does not hold for sounds close to absolute threshold, and alternative equations have been proposed (Humes & Jesteadt, 1991; Zwiskocki, 1965). Stevens proposed the unit sone for the measure. One sone is defined, arbitrarily, as the loudness of a 1 kHz tone at 40 dB SPL.²

²The acronym SPL means sound pressure level; dB SPL should be interpreted as “dB relative to 20μPa.”
2. Introduction

Figure 7: The main stages in Zwicker’s loudness model. A fixed filter accounts for the transmission through the outer and middle ear. The auditory filter bank transforms the spectrum to excitation patterns. Excitation $E(x)$ at place $x$ along the basilar membrane is the output from the filter at this place. Loudness density (specific loudness, $N'$) is calculated based on the excitation patterns. The $N'$-contributions are summed to form the total loudness, $N$.

A number of loudness models for determining the loudness of complex stimuli have been proposed. Zwicker’s model (Zwicker, 1958) combines the filter-bank theory (Fletcher, 1940) with the power law (Stevens, 1957). Zwicker’s loudness model is schematically represented in Figure 7. A fixed filter accounts for the transmission through the outer and middle ear. In the cochlea, the auditory filter bank transforms the spectrum to excitation patterns. Loudness density (specific loudness), the loudness contributions from each auditory filter band, is then calculated using a modified version of Steven’s law, and summed to give the total loudness. The model has been modified and extended to include hearing impairment (e.g., Florentine & Zwicker, 1979; Launer, 1995, and Moore & Glasberg, 1997). Loudness models for time-varying sounds have also been derived for normal and impaired hearing (Zwicker, 1977; Chalupper & Fastl, 2002).

The loudness model of Moore & Glasberg (1997) is used in this study. The model is applicable for both normal and impaired hearing. The total hearing loss is assumed to consist of two parts, one due to outer hair cell damage and the other due to inner hair cell damage. Hearing loss due to outer hair cell damage is modelled by raising an internal absolute threshold. Hearing loss due to inner hair cell damage is modelled by attenuating the excitation level. Moore and Glasberg suggest that the hearing loss due to outer hair cell damage should be set to 80% of the total hearing loss for...
mild hearing losses and to 70% of the total hearing loss for moderate to severe hearing losses (up to a maximum of 55 dB below 2 kHz and 65 dB at 2 kHz and above). The model is based on long-term root-mean-square (RMS) spectra for the sound and does not include the effects of time-varying signals.

2.5 Speech Intelligibility Index

The speech intelligibility index, SII, is a method to quantify the audibility of a speech signal (French & Steinberg, 1947; Kryter, 1962; Pavlovic, 1987; Pavlovic & Studebaker, 1984). ANSI S3.5 (1997) describes a method for calculating SII based on measured or estimated speech spectrum levels, noise spectrum levels, and hearing threshold levels. Audibility of a speech signal is determined in a number of frequency bands. Background noise, the hearing threshold, and self-masking of the speech signal can limit the audible part of the speech signal. When the audibility of the speech signal is determined, the contributions from different frequency bands are weighted according to their importance in understanding speech, and the sum across the frequency bands constitutes the total SII:

\[ SII = \sum_{i=1}^{n} A_i I_i, \]

where \( A_i \) is the audible part of the speech signal in band \( i \), and \( I_i \) the band importance function for band \( i \).

In the standard from 1997, a level distortion factor (LDF) is introduced based on observations that speech recognition performance deteriorates at high presentation levels. When the overall speech sound pressure level exceeds 72 dB SPL, the contribution to SII is reduced:

\[ SII = \sum_{i=1}^{n} A_i I_i L_i, \]

where \( L_i \) is the level distortion factor in band \( i \).

Even if the hearing threshold is one of the parameters used when calculating SII, the standard method is derived for otologically normal listeners, and the effects of reduced supra-threshold performance are not included. Studies have shown that people with sensorineural hearing impairment perform worse on speech tests than predicted by the SII (e.g., Ching, Dillon, & Byrne, 1998; Hogan & Turner, 1998; Pavlovic, 1984; Rankovic, 1991). Various modifications to the method have been proposed. The audibility function may be replaced by effective audibility \( e_i(A_i) \):

\[ SII = \sum_{i=1}^{n} e_i(A_i) I_i L_i, \]
where \( c_i(A_i) \) is the effective audibility in band \( i \), a function of the audible part of the speech signal \( A_i \) in that band. In the study reported in Paper II, two ways of calculating the effective audibility are explored.

Pavlovic, Studebaker, and Sherbecoe (1986) suggest that the audibility contributions from the various frequency bands should be multiplied by a threshold-dependent desensitisation factor, \( D_i \):

\[
e_i(A_i) = D_i A_i,
\]

where \( D_i \) is the desensitisation factor in band \( i \) and depends on the hearing threshold. The desensitisation factor decreases linearly from 1 to 0 as the hearing threshold increases from 15 to 94 dB HL. This modification leads to a reduction in SII contributions from frequency regions with large hearing loss, and it improves the speech recognition prediction for hearing-impaired people (Magnusson, 1996a).

Ching, Dillon, Katsch, and Byrne (2001) suggest another, more complex, function to calculate effective audibility based on sensation levels and hearing loss. In short, the maximum value of the effective audibility in each frequency band varies as a function of hearing loss, but deviates markedly from 1 only when the hearing loss is fairly severe. The function that governs how effective audibility grows with sensation level also depends on the hearing loss and is not a linear function. Effective audibility grows fast for low sensation levels. Furthermore, instead of limiting the usable range of sensation levels to 30 dB (the maximum contribution to audibility according to the standard), the value of the effective audibility increases asymptotically towards the maximum, with contributions also for sensation levels that are greater than 30 dB.

Calculated SII values can be converted to predicted speech recognition scores using a transfer function that is specific for the speech material. With a highly redundant speech material, such as running speech, 100% correct in a speech recognition test can be achieved with a much smaller SII value than for a low-redundant speech material, such as nonsense words. Magnusson (1996b) has derived transfer functions that make it possible to calculate predicted speech recognition scores for a given SII value for two Swedish speech test materials.

2.6 Hearing Aids

A cochlear hearing loss leads to a number of perceptual consequences as described above. The hearing aids we know today are mainly designed to alleviate the problems of reduced audibility and abnormal loudness perception. Most hearing losses call for gain that varies with frequency. A hearing aid that can adjust its gain according to the sound pressure level of the incoming sound is needed in order to compensate for the reduced dynamic
range, which is a characteristic of cochlear hearing loss. As an example, in Figure 6 the hearing loss at 2 kHz is 45 dB HL. In order to make a 2-kHz tone that is just above the threshold for a normal-hearing person (0 dB HL) audible for this hearing-impaired person, the gain needs to be 45 dB. Assume that the hearing-impaired person then listens to a 2-kHz tone that is 80 dB HL. If the same amount of gain is given for that signal, the amplified sound will be intolerably loud, far above the person’s loudness discomfort level, which in this example is only slightly raised from the normal level. A nonlinear hearing aid can analyse the input signal with regard to level and adjust the gain accordingly.

Hearing aids are often classified as linear or nonlinear. The classification is not strict according to the theory of signals and systems. Hearing aids are called linear if their signal processing is linear over a wide range of input levels. However, all linear hearing aids are nonlinear for very high input levels, where the output is limited, most often due to safety aspects, but also due to technical limitations.

Nonlinear hearing aids, on the other hand, are nonlinear not only for the highest input levels, but also for input signals such as speech at normal levels. However, for very low input levels these hearing aids are often linear. The nonlinear hearing aids used today have good sound quality and very small amounts of disturbing nonlinear distortion. This is possible because the signal processing is “quasi-linear”, i.e., the hearing aid’s signal processing characteristics change relatively slowly (usually over periods longer than
5 ms) (Leijon & Nordqvist, 1999). Over a short time frame, the system’s characteristics do not change. For this reason, the nonlinearity primarily affects the envelope pattern of the input signal, and does not distort its fine structure.

The compression ratio is a measure of the inverse slope of the input-output function (Figure 8). For a linear hearing aid, the compression ratio is 1. The compression threshold indicates the input level where the hearing aid changes from linear to nonlinear signal processing. A wide dynamic range compression (WDRC) hearing aid has a low compression threshold (often below 60 dB SPL), usually combined with a rather low compression ratio (typically about 2:1). Examples of gain-frequency responses and an input-output function for a WDRC hearing aid are seen in Figure 8.

2.7 Hearing Aid Fitting

The initial step in the hearing aid adjustment process is to find the best preliminary hearing aid settings, so that the hearing-impaired person can start using the hearing aids. There are so many hearing aids and adjustment possibilities available that this initial step must be carried out in a prescriptive manner. A prescription is basically a formula by which the audiologist can calculate a reasonable gain setting based on some measures of auditory function, usually the hearing threshold.

A number of prescriptive methods for WDRC instruments have been developed. The methods differ in several ways. They use different underlying rationales, such as speech intelligibility maximisation, or loudness density normalisation or equalisation (reviewed by Byrne, 1996). The input data to the methods also vary. Some methods use only hearing threshold levels, and some are based on supra-threshold measurements such as loudness scaling. Some methods are generic and intended to be used for different types of hearing aids. Examples are NAL-NL1 (Byrne, Dillon, Ching, Katsch, & Keidser, 2001), DSL[i/o] (Cornelisse, Seewald, & Jamieson, 1995), FIG6 (Killion & Fikret-Pasa, 1993), IHAFF (Cox, 1995), CAMEQ (Moore, Alcántara, Stone, & Glasberg, 1999; Moore, Glasberg, & Stone, 1999), and CAMREST (Moore, 2000). Other methods are developed by hearing aid manufacturers for their specific devices and are usually not described in reviewed journals. Below, three rationales are reviewed.

Maximisation of Speech Intelligibility

The most basic requirement in order to achieve good speech intelligibility is that the speech sounds are audible, i.e., above the listener’s hearing threshold and not masked by background noise. This requirement leads to different gain settings in different listening environments. Both the amount
Figure 9: Speech spectra for three listening situations (panels A, B, and C) plotted with normal hearing thresholds (dashed line) and the hearing threshold from Figure 6 (solid line with \( \times \)). Speech RMS spectra (solid line) and the dynamic range of speech (dotted lines) are plotted. Listening situation A represents a distant speaker in an environment without background noise. Listening situation B represents average level speech at a normal conversation distance. Listening situation C represents loud speech (raised vocal effort) in background noise and at close distance. (Redrawn from Smeds and Leijon (2000) with permission from the authors and CA Tegnér AB.)

of gain and the shape of the gain-frequency response need to change between listening situations in order to achieve good audibility.

Consider a person with a typical cochlear hearing loss. In Figure 9, speech RMS spectra and the dynamic range of speech for three listening situations are plotted together with the hearing thresholds for normal-hearing persons and the hearing loss from Figure 6. In a listening situation with distant speech in a quiet background (panel A), high gain over a broad frequency range is needed to make the speech signal audible. More gain is needed at high frequencies than at low frequencies. In a listening situation with average level speech at a normal conversation distance (panel B), low-frequency speech components are audible without amplification, but gain is needed in the frequency region above 1 kHz. In a listening situation
with loud speech (raised vocal effort) in a noisy background (babble at a signal-to-noise ratio of 5 dB, panel C), most of the speech signal is audible without amplification and only gain at the highest frequencies can increase the amount of speech information available to the listener.

The SII measure can be used to find a gain-frequency response that maximises speech intelligibility (e.g., Köbler & Leijon, 1999; Magnusson, Karlsson, & Leijon, 2001; Rankovic, 1991). The NAL-NL1 hearing aid fitting method aims at optimising speech intelligibility, using one of the modified versions of the SII described above (Ching et al., 2001), under the constraint that total loudness should be equal to or less than normal (Byrne et al., 2001).

**Loudness Density Normalisation**

All hearing aid fitting methods attempt to give amplified speech acceptable loudness. Most prescriptive methods aim at providing the hearing aid user with the same overall loudness that a normal-hearing person would perceive without amplification for the same situation. This goal is called loudness normalisation. If the goal is to restore not only the overall loudness to normal, but also the relative loudness balance across frequency, the strategy is called loudness density normalisation (Leijon, 1991). Loudness density normalisation will amplify a broadband sound so that each auditory filter gives the same loudness contribution as would a normal ear. If loudness density normalisation is achieved, loudness normalisation is accomplished. The reverse is not true; normal overall loudness can be achieved with a variety of gain-frequency responses.

The rationale behind loudness density normalisation is that restoring the relative loudness contributions to normal across frequency will present sound at an acceptable loudness and with good sound quality. However, loudness density normalisation does not guarantee maximal speech intelligibility. A method like CAMREST (Moore, 2000) is based on loudness density normalisation. Many other prescriptions are based on normalising loudness contributions across frequency (e.g., Allen, Hall, & Jeng, 1990; Beck et al., 1994; Kiessling, Schubert, & Archut, 1996; Killion & Fikret-Pasa, 1993; Ricketts, 1996).

**Loudness Density Equalisation**

Another rationale that also builds on loudness density is loudness density equalisation. This rationale takes the spectral characteristics of speech into account, and the goal is to amplify the speech spectrum to constant loudness density across frequency. Thus, all auditory filters will give the same contribution to the total loudness. With this approach, speech components
are not presented with their normal loudness relations. The low-level, high-frequency speech signals are amplified to a higher than normal loudness, whereas high-level, low-frequency speech signals are amplified to less than normal overall loudness.

The rationale behind this strategy is to make a large portion of the speech signal audible, regardless of the volume control setting. Often the rationale is accompanied by a constraint which states that the overall loudness should be normal. The NAL-R method (Byrne & Dillon, 1986) and CAMEQ (Moore, Alcántara et al., 1999; Moore, Glasberg et al., 1999) both use loudness density equalisation as the underlying rationale.

Threshold and Supra-Threshold Based Prescriptions

Prescriptive methods calculate hearing aid gain based on some measure of hearing abilities. Threshold-based prescriptive methods are the most common. Gain calculations are based only on the hearing-impaired person’s absolute hearing thresholds. Hearing models and statistical data are then used to predict the hearing-impaired person’s performance at, the more relevant, supra-threshold levels. For example, loudness models for normal and impaired hearing can be used to calculate the required gain according to a loudness density normalisation strategy.

An alternative to threshold-based prescriptive methods is to measure some supra-threshold abilities. The hearing threshold is usually supplemented by a measure of perceived loudness. This could be the loudness discomfort level, which, together with the hearing threshold, can be used to calculate the hearing-impaired person’s auditory dynamic range. Sometimes a more thorough investigation of the auditory dynamic range is advocated (e.g., Allen et al., 1990; Beck et al., 1994; Kessling et al., 1996; Ricketts, 1996).

It has also been suggested that other measures of the signal-processing ability of the cochlea could be useful for fitting hearing aids. Lindblad, Olofsson, and Hagerman (1993) and Olsen, Lindblad, Olofsson, and Hagerman (1999) describe a hearing aid fitting method that uses psycho-acoustical modulation transfer functions to explore the supra-threshold properties of the damaged cochlea. They state that a reasonable goal for an amplification scheme is to amplify the speech spectrum to the level where the signal-analysing properties of the ear are the best. However, the results from this research are inconclusive.

Comparison Between Prescriptive Methods

Comparisons between prescriptive methods for linear hearing aids often show large differences when the prescriptions are compared theoretically, but the resulting differences in aided performance are usually small (e.g.,
2. **Introduction**

Byrne, 1987). There are two principal explanations for this. First, linear hearing aids have a volume control. If the comparison is made in a field trial, differences in absolute gain are not taken into account, because the volume control setting is unknown. Therefore, some of the fitting methods for linear hearing aids only prescribe the gain-frequency shape, not the absolute gain. Second, due to technical deficits, the hearing aids used in previous evaluation studies usually could not provide the prescribed high-frequency gain. This means that theoretical differences in both absolute gain and in gain-frequency shape were reduced in practice.

At the start of the current thesis work, there was a lack of studies comparing prescriptive methods for WDRC hearing aids. Theoretical comparisons between general fitting methods for nonlinear hearing aids had revealed large differences in prescribed gain, both in gain-frequency response shape and in overall gain (e.g., Ricketts, 1996; Stelmachowicz et al., 1998). Both differences are important. For a linear hearing aid with a volume control it can be argued that the overall gain setting is not that important, because it can be changed by the user. For WDRC hearing aids, which often lack volume controls, it is important to study also the differences in prescribed overall gain levels, not only differences in gain-frequency response shape.
Chapter 3

Aims and Outline of the Current Work

The most common type of hearing loss has cochlear origin. Cochlear hearing loss is associated with a reduced auditory dynamic range. The hearing loss is best alleviated by the use of nonlinear, wide dynamic range compression (WDRC) amplification. At the start of this thesis work there was a lack of studies comparing prescriptive methods for WDRC hearing aids.

The general aim of the current study was to investigate prescriptive methods for WDRC hearing instruments. The goal was to draw general, not only method-specific, conclusions about the preferences of hearing aid users. Therefore, the prescriptions were evaluated using well-established models for loudness and speech intelligibility.

It was felt that prescriptive methods play a larger role when fitting hearing aids to first-time hearing aid users than to experienced hearing aid users. First-time hearing aid users would typically have a less clear idea of what they want in terms of amplification, and of what the hearing aids can offer. As a result of this choice, the work focuses on hearing losses typical for this group of clients. The hearing losses are described as mild to moderate and include hearing thresholds better than 60 dB HL except at the highest frequencies, 4-8 kHz, where the hearing loss can be more severe. This is not a serious limitation to the scope of the work, because about 75% of the clients seen in a typical Swedish hearing clinic have hearing losses in that range (Holst, 2002). In one of the studies included in the thesis (reported in Paper IV and Paper V), experienced hearing aid users with mild to moderate hearing losses also participate for comparison.
3. AIMS AND OUTLINE OF THE CURRENT WORK

There are four main questions for the thesis:
1. How much do existing prescriptive methods for WDRC hearing aids differ in prescribed gain?
2. How do generic prescriptive methods compare with regard to measured and calculated loudness and speech intelligibility?
3. Do first-time hearing aid users prefer overall loudness to be restored to normal?
4. What overall loudness do normal-hearing and hearing-impaired listeners, both with and without hearing aid experience, prefer?

The four questions are treated in the following four chapters of the thesis. The results from the studies are then discussed and compared in the subsequent Discussion chapter and answers to the four main questions are presented in the Conclusions chapter.
Chapter 4

Prescriptive Methods—Same or Different? (Paper I)

Theoretical comparisons between general fitting methods for nonlinear hearing aids have revealed large differences in prescribed gain. For WDRC hearing aids, which often lack volume controls, it is important to study differences in overall gain, and existing methods show large differences in both gain-frequency response shape and in overall gain (Ricketts, 1996; Stelmachowicz et al., 1998). Are these differences preserved when the methods are implemented in hearing aids?

The aim of this first study (Paper I) was to compare a number of software-implemented prescriptions, both generic and proprietary methods. It was hypothesised that the large differences seen when fitting methods are compared theoretically would be reduced when the prescriptions were implemented in hearing aids. There were three main questions:

1. How do prescriptive methods, both generic and hearing aid specific, compare in prescribed gain when they are implemented in hearing aids?
2. Do the prescriptive methods lead to differences in calculated loudness?
3. Do the prescriptive methods lead to differences in calculated $SII$?

4.1 Methods

Six modern WDRC hearing aids were selected and programmed according to the manufacturers’ suggested initial fittings for a hypothetical audiogram, representing a mild to moderate hearing loss (20 dB HL at 250 Hz, and gently sloping down to 70 dB HL at 8 kHz). The prescribed gain was compared using coupler measurements with a speech-like signal, the ICRA noise (Dreschler, Verschuure, Ludvigsen, & Westermann, 2001). Three simulated
listening situations were used: soft speech in a quiet background (55 dB SPL), average level speech in a quiet background (65 dB SPL), and loud speech (74 dB SPL) in a background of speech babble (71 dB SPL). Coupler gain results were transformed to estimated insertion gain, and loudness and SII were calculated.

### 4.2 Results

Surprisingly, the differences among the hearing aid implemented gain-frequency responses were as large as the differences found at theoretical comparisons (Figure 10). These differences in gain led to large differences in calculated loudness, but only to minor differences in calculated SII.
Chapter 5

Laboratory Comparison between Prescriptive Methods (Paper II)

After finding that the differences between prescriptive methods were so large, the resulting questions was: Which prescription do hearing-impaired listeners prefer? The prescribed gain should be correct, on average, for a listener with the audiogram used.

The results in Paper I indicated that the prescriptions led to large differences in calculated loudness, but only to small differences in calculated speech intelligibility. Do the results hold in an experiment using hearing-impaired participants? The aim of this second study (Paper II) was to compare generic prescriptive methods for WDRC hearing aids in terms of loudness, speech intelligibility, and preference. There were three main research questions:

1. Do prescriptions for WDRC hearing aids differ in calculated and perceived loudness, judged by hearing-impaired listeners?
2. Do prescriptions differ in calculated and measured speech intelligibility, tested in speech recognition tests with hearing-impaired listeners?
3. How is preference, judged by hearing-impaired listeners, related to measured loudness and speech intelligibility?

5.1 Methods

Three prescriptive methods were compared in a laboratory study where 20 hearing-impaired listeners with mild to moderate hearing loss and without hearing aid experience participated. Loudness was calculated using the loudness model of Moore and Glasberg (1997), rank ordered between prescriptions using paired comparisons, and rated using category ratings. Speech recognition results were predicted using three versions of the SII
5. **Paper II**

(according to the standard (ANSI-S3.5, 1997) and using two desensitisation factors (Ching et al., 2001; Pavlovic et al., 1986)) and measured using a Swedish phonemically balanced speech material. Preference was explored using paired comparisons.

5.2 **Results**

In Figure 11, the results from paired comparisons of loudness are shown in the top panels (left panel for a situation with speech at 55 dB SPL and right panel for a situation with speech at 75 dB SPL and babble noise at 71 dB SPL). The middle panels show the calculated loudness. When the top and middle panels are compared, it can be seen that the calculated rank order between the methods agrees with the measured. The bottom panels show the results of the category ratings of loudness. The participants judged if presented sentences were “Too soft”, “OK”, or “Too loud”, and the shades in the stacked columns represent the three answers. In addition to the results for the hearing-impaired participants, results for a group of normal-hearing listeners are included.

In the speech recognition test, the only statistically significant difference between the methods was seen for the 55-dB situation where Method A, which prescribed considerably less gain than the other methods, led to lower scores than the rest of the methods. The predicted speech recognition scores showed that the calculations according to the standard and according to Ching et al. generally over-predicted the results, whereas the calculations according to Pavlovic et al. tended to under-predict the measured scores for the 76-dB situation, but gave close to measured scores for the 55-dB situation.

For the listening situation with low-level speech, the participants preferred the method that prescribed the highest gain. For the listening situation with high-level speech in background noise, the participants preferred methods that prescribed low gain.

Method C was based on CAMEQ (Moore, Alcántara et al., 1999; Moore, Glasberg et al., 1999), and participants seemed to do well with the prescription in the tests. The rated loudness for the hearing-impaired listeners using CAMEQ was close to the ratings for the normal-hearing listeners, and the prescription led to the highest speech recognition scores for both situations.
Figure 11: Various loudness measures for a 55-dB situation (left panels) and a 76-dB situation (right panels). The top panels show the results from the paired comparisons of loudness. The number of times one method was judged as louder than the others is presented. The middle panels show the results from loudness calculations using the loudness model of Moore and Glasberg (1997). The overall loudness, in sone, has been recalculated to give the loudness level, in phon, relative to normal hearing listeners, i.e., 0 phon re. normal indicate that the loudness level is normal according to the loudness model used. Medians, inter-quartile ranges, 10th, and 90th percentiles are presented. The bottom panels show the results from the category ratings of loudness. The number of times the various methods have been judged to be “Too soft”, “OK”, or “Too loud” are presented. Here, data for 23 normal-hearing (NH) listeners are included as a comparison.
Chapter 6

Do First-Time Hearing Aid Users Prefer Normal Overall Loudness? (Paper III)

The results presented in Paper II indicated that the large differences in calculated loudness, which were seen when comparing generic prescriptions in both Paper I and Paper II, were clearly distinguishable to the hearing-impaired participants. The method that led to high loudness, relative to normal, was preferred for low-level speech, whereas methods that led to lower loudness, relative to normal, were preferred for high-level speech. These results might be influenced by the laboratory condition, where there were no negative consequences of high gain for low-level sounds because testing was done using headphones in a sound-treated booth.

In this third study (Paper III) two prescriptive methods for WDRC hearing aids were compared both in the field and in well-controlled laboratory tests. The study focuses on differences in overall loudness. Most prescriptive methods for WDRC hearing aids are based on the assumption that a hearing-impaired listener should perceive amplified sounds at the same overall loudness as would a normal-hearing listener without amplification. However, some previous research on linear amplification has indicated that participants prefer less overall gain than prescribed by the most commonly used prescriptive method for linear hearing aids, NAL-RP (Humes, Wilson, Barlow, & Garner, 2002; Leijon, Lindkvist, Ringdahl, & Israelsson, 1990), a method that gives close to normal overall loudness for a mid-level input and mild to moderate hearing losses.

The aim of this study was to compare two prescriptive methods, which differed in the overall loudness they provided the hearing aid user. There were two main research questions:
1. Do first-time hearing aid users, with typical hearing losses for this group of clients, prefer a method that restores overall loudness to normal or a method that gives less than normal overall loudness?

2. How do the methods compare in terms of measured speech recognition?

6.1 Methods

One method, called NormLoudn, was based on CAMEQ, a generic method that prescribes gain with the aim to equalise loudness density and restore overall loudness to normal according to the loudness model of Moore and Glasberg (Moore & Glasberg 1997; Moore, Alcántara et al., 1999; Moore, Glasberg et al., 1999). NormLoudn was the same method as Method C in the previous study. Another method, called LessLoudn, was based on a hearing-aid–specific prescription that aims at loudness density normalisation. Despite its goal, the method led to less than normal overall calculated loudness according to the loudness model of Moore and Glasberg. Twenty-one first-time hearing aid users with typical hearing losses for this group of clients participated in a crossover blinded field study where the two fitting methods were compared using a multi-programmable hearing aid. Preference in the field was evaluated in an interview, with a questionnaire, and using a diary. The field test was accompanied by laboratory tests, which included paired comparison judgments of preference and loudness as well as a speech recognition test. Loudness calculations were used when interpreting the results, and a theoretical comparison with other prescriptive methods for WDRC hearing aids was made.

6.2 Results

After necessary adjustments, the measured gain for the two methods was surprisingly similar in gain-frequency response shape considering the deviating rationales of the two methods. However, NormLoudn prescribed higher overall gain than LessLoudn (Figure 12). Generally, NormLoudn fittings led to calculated overall loudness that was close to normal, whereas LessLoudn fittings, in median, led to 3–7 phon less than normal calculated overall loudness according to the loudness model used. At the interview performed after the field test, 19 out of the 21 participants stated that they preferred LessLoudn. Also the questionnaire and the diary showed a clear preference for LessLoudn in all types of listening situations. Paired comparisons of preference in the laboratory supported the findings in the field. LessLoudn was preferred to NormLoudn in all tested situations, except for soft speech in very soft noise where there was no significant preference for
Results

Figure 12: Measured insertion gain, presented as the median across 42 ears, for the 21 participants. A speech-weighted noise signal was used for the measurements. The top panel shows measurements for 55 dB SPL input, the bottom panel measurements for 80 dB SPL. Inter-quartile values were in the 5–10 dB range for frequencies above 500 Hz.

either method. Speech recognition scores were similar for the two prescriptions. The difference in calculated loudness was clearly distinguishable to the participants and seemed to govern their preferences.

A comparison between the measured gain for NormLoudn and the gain prescribed by CAMEQ, NAL-NL1, and DSL[i/o], suggests that all three prescriptive procedures, and DSL[i/o] in particular, would probably overestimate the required gain for participants without hearing aid experience and with mild to moderate hearing loss.
Chapter 7

Preferred Overall Loudness
(Papers IV and V)

The results from the third study showed that first-time hearing aid users did not prefer overall loudness to be restored to normal. Two existing prescriptive methods were compared, but no quantification of the preferred deviation from normal overall loudness was made.

The aim of this fourth study (reported in Paper IV and Paper V) was to determine the preferred overall loudness for hearing-impaired listeners. Perhaps the previously found results were biased by the fact that the participants were first-time hearing aid users. Hearing-impaired listeners both with and without hearing aid experience were included in this study. Perhaps not even normal-hearing listeners prefer normal overall loudness if given a choice. To explore also this possibility, normal-hearing participants were included in the study. There were three main questions:

1. What overall loudness do hearing-impaired listeners prefer?
2. Is there a difference in overall loudness preference for inexperienced and experienced hearing aid users?
3. What overall loudness do normal-hearing listeners prefer?

7.1 Methods

Fifteen normal-hearing (NH) and 24 hearing-impaired (HI) people participated in a laboratory study (Paper IV). The hearing-impaired participants had mild to moderate, symmetrical cochlear hearing loss. Half of the hearing-impaired participants had previous hearing aid experience, whereas half of them had none. In the laboratory, the participants watched and listened to video sequences, representing various types of listening situations, both speech and non-speech situations, ranging in overall sound-field levels from
46 to 86 dB SPL. For the hearing-impaired listeners, a slow-acting AGC hearing aid was simulated using a digital equaliser. The participants’ first task was to rate how loud and how interesting they found the listening situations. The participants then adjusted a volume control to preferred loudness. Finally they rated loudness and interest again, now using their preferred volume control settings.

A subset of the participants (8 normal-hearing and 15 hearing-impaired participants) from the laboratory study participated in a field trial (Paper V), where the task was to wear a binaurally fitted digital research hearing aid for a week and adjust a volume control to preferred loudness in everyday listening situations. The hearing aid logged the participants’ preferred loudness and the acoustic characteristics of the environments in which those preferences were determined.

In order to tie the two experiments together, the participants from the field test also partook in a final experiment where the laboratory test was repeated, now using the research hearing aid.

7.2 Results

The hearing-impaired participants preferred considerably less than normal overall calculated loudness, and the data from the laboratory test and the field test were in good agreement. In the laboratory, the preferred loudness levels varied from about −5 phon re. normal\(^1\) for lower input levels to −15 phon re. normal for higher input levels. In the field, the preferred loudness levels did not vary with input level and were about −14 phon re. normal (Figure 13). Neither in the laboratory tests, nor in the field test were there any differences between the groups of inexperienced and experienced hearing aid users.

The normal-hearing participants preferred closer to normal overall loudness (Figure 13). The results from the laboratory test and the field test did not agree as well as for the hearing-impaired participants. In the laboratory the participants generally preferred less than normal overall loudness, whereas in the field they preferred slightly higher than normal overall loudness.

When rated loudness from the laboratory experiment was compared for normal-hearing and hearing-impaired participants (Figure 14), it was seen that the hearing-impaired participants rated loudness higher than the normal-hearing participants, especially at high input levels. Loudness calculations for the laboratory situations revealed that the prescriptive method

\(^1\) The unit “phon re. normal” refers to a calculated loudness level difference. In this case, loudness levels at preferred volume control settings were calculated and the loudness level for a normal-hearing person in the same situation, without any adjustments, was subtracted.
Figure 13: Comparison between the loudness levels (relative to normal) preferred in the two laboratory tests and the field test for the input range 50–79 dB SPL and for the people who participated in all experiments (8 normal-hearing and 15 hearing-impaired participants). The upper panel shows the results for the normal-hearing participants and the lower panel the pooled results for the two groups of hearing-impaired participants. The laboratory situations were grouped according to their original levels (prior to amplification for the hearing-impaired participants) into the three 10-dB intervals used. Average loudness levels for each of the 23 participants in the field test were calculated for each 10-dB interval, and percentiles were calculated across participants. Median, inter-quartile, maximum, and minimum values are shown.
Figure 14: Scaling of loudness for normal-hearing (NH) and hearing-impaired listeners without (HI Inexp) and with (HI Exp) hearing aid experience. Medians, inter-quartile, maximum, and minimum values across the participants are shown. The original C-weighted sound pressure levels (prior to amplification with NAL-NL1 for the hearing-impaired participants) for each of the situations are presented.

used (NAL-NL1) actually did not lead to normal, but less than normal overall calculated loudness. Despite the fact that the hearing-impaired participants were provided with less than normal overall loudness, they rated loudness higher than the normal-hearing participants.
Chapter 8

Discussion

8.1 Paper I

The results presented in Paper I showed that the prescriptive methods for WDRC hearing aids, implemented in hearing aid manufacturers’ fitting software, varied substantially in prescribed gain for the typical audiogram used. The most modern hearing aids at the time of the study (1998) were chosen. Are the results still relevant? In a recent study, Keidser, Brew, and Peck (2003) compared generic and proprietary prescriptions for a number of hypothetical audiograms. They did not measure the prescribed gain, but derived the prescriptions from the fitting software. Their results were in agreement with those in Paper I. The generic method DSL[i/o] prescribed the highest gain values among all prescriptions. For all audiograms, except for a steeply sloping loss, the other generic method included, NAL-NL1, gave the second highest gain values.

To be able to compare the results in Paper I with the results from Keidser et al., NAL-NL1 gain was calculated for a 65-dB SPL broadband input signal (using stand-alone software v. 1.28) for the audiogram used in Paper I. When the prescribed NAL-NL1 gain is compared to the measured gain for the six hearing-aid-implemented prescriptions from Paper I (Figure 15), one can see that NAL-NL1 lies in the middle of the range of gain-frequency responses in the low frequencies, but at the bottom of the range at higher frequencies. This is in contrast to the results by Keidser et al. If the results from the two studies are comparable, it suggests that the variation in prescribed gain is still large, but the proprietary prescriptions now prescribe less gain than they did 1998.

The study reported in Paper IV and Paper V showed that NAL-NL1 prescribed too much gain for both inexperienced and experienced hearing aid users with mild to moderate hearing loss. As a consequence, most of the methods explored in Paper I would prescribe too high gain. It seems as
8. Discussion

Figure 15: Measured gain for a speech-like signal at 65 dB SPL for six software-implemented prescriptions (thin lines, from Paper I). As a reference, NAL-NL1 prescribed gain for the same audiogram and listening situation is shown (bold line).

If the manufacturers’ change towards lower prescribed gain is in line with hearing-impaired listeners’ preference.

8.2 Paper II

The study presented in Paper II compared three generic prescriptive methods for WDRC hearing aids. The methods differed substantially in calculated and rated loudness, whereas the differences in speech intelligibility were small. Method B, based on DSL[i/o], prescribed higher gain than the rest of the methods, probably too much gain for high-level speech in background noise. Method C, based on CAMEQ, seemed to do well in the study. The method provided the highest speech recognition scores for both listening situations, it was preferred in the high-level situation, and it gave closest to normal overall loudness. The method was a natural candidate when comparing prescriptions in the field (Paper III).

8.3 Paper III

The results in Paper III indicate that hearing-impaired listeners preferred a method that did not restore overall loudness to normal according to the loudness model of Moore and Glasberg. The preferred prescription, Less-Loudn, was based on the proprietary fitting method (Logic) for the hearing aid chosen for the study (Danalogic 163D). In order to relate the results from Paper I and Paper III, additional coupler measurements were made using the Danologic hearing aid and the audiogram from Paper I. For the
65-dB SPL input signal, the measurements showed that Logic was among the methods that prescribed the least gain for the audiogram (Figure 16).

Based on the measurements presented in Paper I, it was only possible to conclude that the variation in measured gain was so large that it was justified to ask if the prescriptions could all be “correct” for the average hearing aid user with the particular audiogram. When the results from Paper III are included, it can, again, be concluded that it probably was the prescriptions in the lower range of the set of gain-frequency responses that were more “correct” than the prescriptions in the higher range. CAMEQ, which did well in the comparison in Paper II, provided too high gain for use in real-life situations.

![Graph](image)

Figure 16: Measured gain for a speech-like signal at 65 dB SPL for six software-implemented prescriptions (thin lines, from Paper I). As a comparison, measurements using the preferred prescription from Paper III is shown (bold line).

### 8.4 Papers IV and V

The fourth study, reported in Paper IV and Paper V, showed that the hearing-impaired participants preferred considerably less than normal overall loudness, as calculated by the loudness model of Moore and Glasberg (1997). Preferences were determined both in the laboratory and in the field. Both the laboratory test and the field test had limitations. The main limitation in the laboratory was the fact that changing the volume control only changed the sound from the loudspeakers and very low volume control settings did not create a feeling of being “blocked”, or create a risk of missing soft sounds. The main limitation in the field was that the hearing aid was used with unfitted earmoulds for all participants, regardless of their low-frequency hearing. This resulted in problems with occlusion. What gives
the study its strength is the fact that both tests led to similar results for the hearing-impaired listeners.

The results for the normal-hearing participants were inconclusive. They generally preferred closer to normal overall calculated loudness than the hearing-impaired listeners, but the results from the laboratory and the field differed. We cannot conclude that normal-hearing listeners prefer overall loudness that differs from normal.

Loudness calculations based on the original presentations in the laboratory revealed that the hearing-impaired listeners were provided with less than normal overall calculated loudness. Despite this, they rated loudness higher than the normal-hearing listeners, especially for high input levels. This indicated that there is a problem with the loudness model used when comparing the loudness calculated for normal and impaired hearing.

The results in Paper III, Paper IV, and Paper V, did not agree with results obtained by Marriage, Moore, and Alcántara (2004). They found that first-time hearing aid users needed to have the gain reduced by –3.5 dB (on average) from prescriptions that aim at normal overall loudness (CAMEQ and CAMREST (Moore, 2000; Moore, Alcántara et al., 1999; Moore, Glasberg et al., 1999)), whereas a group of experienced hearing-aid users needed only small deviations (–1 dB). In the study reported in Paper IV and Paper V preferred loudness was determined, as opposed to measuring the minimal adjustments necessary for acceptable loudness as in the Marriage et al. study. Other potential explanations to the differences between the studies are discussed below.

8.5 Hearing Aid Experience

The thesis has focused on first-time hearing aid users. In a typical Swedish hearing aid clinic, about 60% of the hearing aid fittings performed in a year are for first-time hearing aid users (Holst, 2002), and it was argued that prescriptive methods play a more important role for first-time hearing aid users.

In the study presented in Paper IV and V, half of the hearing-impaired participants had previous hearing aid experience, whereas half of them had not. Not in any of the experiments were there any statistically significant differences between the two groups in terms of rated loudness or in terms of loudness preference.

It generally seems difficult to find scientific support for the sometimes-quoted clinical experience that first-time hearing aid users will gradually prefer higher listening levels. There are, however, studies indicating that experienced hearing aid users tolerate or prefer higher listening levels, or higher gain, than first-time hearing aid users, at least for high input levels (Cox & Alexander, 1994; Marriage et al., 2004; Olsen, Rasmussen, Nielsen,
We know that people can get used to high loudness. One consequence of using linear hearing aids is that if gain is selected so as to make average-level speech comfortable, high-level speech will be too loud. With linear amplification, hearing aid wearers had to get used to this (or be very quick with the volume control). The participants with hearing aid experience in the studies by Cox & Alexander (1994) and Olsen et al. (Olsen, personal communication 2000) were all used to linear amplification. In the study by Marriages et al. (2004), 9 participants were used to linear amplification and 11 participants were used to nonlinear amplification. Perhaps many experienced hearing aid users today are biased by previous experience with too much amplification in linear hearing aids. The question still remains if first-time hearing aid users will gradually prefer more gain as they acquire more hearing aid experience.

Humes, Barlow, Garner, and Wilson (2000) and Humes, Wilson, Barlow, and Garner (2002) measured preferred hearing aid gain settings for a group of 134 participants (55 participants in the 2000 study), using linear amplification. Among the participants, 56 were experienced hearing aid users, and 78 were first-time hearing aid users. Participants preferred, on average, 6–9 dB less gain than prescribed by NAL-RP (or 3–6 dB less for a binaurally corrected NAL-RP). User-preferred gain was measured 2 weeks, 1 month, 6 months, and 1 year after the initial fitting session, and the preferences were stable over time. When the original data (Humes, personal communication 2003) are analysed for the two groups of participants separately, the picture is the same. The group of inexperienced participants preferred 6–9 dB less gain than prescribed by NAL-RP, and they did not, on average, increase their preferred gain over the first year of hearing aid experience. The gain settings preferred 2 weeks after the fitting were the same as the settings preferred one year after the fitting.

Horwitz and Turner (1997) presented similar results for a smaller group of 13 first-time hearing aid users, 8 provided with nonlinear and 5 with linear amplification. Hearing aid benefit increased over time, but this was not due to increased listening levels. The participants were followed for 18 weeks, and during that period the participants decreased the gain used by, on average, 2 dB relative to the gain chosen at the fitting.

Arlinger, Lyregaard, Billermark, and Öberg (2000) found no systematic preference among first-time hearing-aid users between a manufacturer’s initial prescription and a setting with reduced gain.

The only reason to try to convince first-time hearing aid users to get used to more amplification than they initially prefer would be if there was evidence that higher gain would be beneficial in the long run. However, this still needs to be proved. In the current study, high gain was only advantageous in terms of speech recognition for low-level speech in quiet. For high-level speech in noise, the situation in which most hearing-impaired listeners experience problems, high gain was not in any way advantageous.
One problem with comparing results obtained with hearing-impaired listeners with and without hearing aid experience, is the difficulty of matching the groups with regard to hearing loss. The difference in hearing loss between the inexperienced and experienced hearing aid users in the study by Marriage et al. was larger than in Paper IV and Paper V. The larger the difference in hearing loss between the two groups, the larger the risk of attributing a difference in results to hearing aid experience, when it could really be attributed to differences in hearing loss.

Shanks, Wilson, Larson, and Williams (2002) showed that listeners with mild to moderate hearing loss produced lower mean speech recognition scores (for speech in babble with a constant signal-to-noise ratio) when presentation levels increased from 52 to 62 dB SPL, and from 62 to 74 dB SPL. The probable explanation of the result is that speech was audible already at low presentation levels and higher presentation levels only led to more of the babble becoming audible. People with more severe losses, on the other hand, produced higher speech recognition scores when presentation levels increased from 52 to 74 dB SPL. These results suggest that people with more severe hearing loss might be willing to tolerate higher loudness than people with less severe hearing losses in order to increase speech understanding. In the study reported in Paper IV and Paper V, where the difference in hearing loss between the two groups was smaller than in the study by Marriage et al., the results for the two groups of hearing-impaired participants did not differ significantly.

8.6 Prescriptions for Narrowband and Broadband Sounds

Another difference between the studies reported in Papers III–V and the study of Marriage et al. (2004) lies in exactly how CAMEQ was fitted. The software for CAMEQ, CanFit, provides gain targets both for a broadband speech signal and for pure tones. Based on information about the hearing aid's compression characteristics, the prescription for the broadband signal has been recalculated to gain targets for pure tones. For a hearing aid like Datalogic 163D, with 14 highly overlapping compression channels, the consequences of filter shape and overlap are difficult to predict. The calculation can, for instance, depend on the degree of hearing loss. In the study reported in Paper III, the prescription for a broadband signal was used and verification measurements were made using speech-shaped noise, whereas Marriage et al. used the prescription for narrowband stimuli and verified the prescription using pure tones.

In order to investigate if this difference in targets could have influenced the results from the two studies, hearing aid measurements were made in an anechoic chamber using pure tones and the ICRA noise for a male talker.
Figure 17: The software estimated difference between broadband and narrowband targets at 65 dB SPL minus measured difference using broadband (ICRA noise, male talker, Dreschler et al., 2001) and narrowband (pure tones) signals at 65 dB SPL. Positive values indicate that the software estimated difference between broadband and narrowband stimuli was overestimated.

(Dreschler et al., 2001). The spectral shape for this noise is very similar to the speech-shaped noise proposed as the measurement signal in CamFit. Five audiograms were used: one mild to moderate, gently sloping hearing loss (the median audiogram from Paper III), two flat hearing losses at 40 and 60 dB HL respectively, one steeply sloping hearing loss, and one hearing loss with a reverse slope (better hearing at high frequencies). A Danalogic 163D hearing aid (used both in the study reported in Paper III and in the study by Marriage et al.) was programmed according to the pure-tone targets derived from CamFit for each audiogram, and measurements were made with both pure tones and the speech shaped noise for two input levels, 65 dB SPL and 80 dB SPL, where CAMEQ prescriptions are available. The measured difference in gain for the broadband signal and the narrowband signal was compared to the calculated difference from the CamFit software. The software-estimated difference was overestimated by about 4 dB at 1 kHz for the measurement signal at 65 dB SPL for all audiograms except for the steeply sloping audiogram (Figure 17). For the steeply sloping audiogram, the estimated difference was overestimated by 6 dB at 250 Hz and by 4 dB at 2 kHz. For the measurement signal at 80 dB SPL, on the other hand, the software estimated difference was underestimated by about 4 dB at 4 kHz for all audiograms.

The overestimated difference between the broadband and narrowband signals in the software for the 65-dB SPL measurement signal is consistent with the differences in results from the two studies. In the study reported
in Paper III, the broadband targets were used. In the study by Marriage et al. the narrowband targets were used. These recalculated narrowband targets actually produced a lower hearing aid gain setting than the broadband targets. Thus, the participants in the study of Marriage et al. were fitted with lower gain to start with, which might explain why they did not need as large gain reductions as the participants in the study reported in Paper III. The underestimated difference for the 80-dB SPL signal is more difficult to interpret, because the target at that input level might be changed by the hearing aid’s output-limiting system.

When a prescription is derived from broadband signals, as for CAMEQ, the recalculation to gain targets for narrowband stimuli will always be an approximation. It seems safest to use the broadband prescription, and a measurement signal with a spectral shape as closely matching the spectral shape used when deriving the prescription.
Chapter 9

Conclusions

Based on the studies included in the thesis, the following four main conclusions can be drawn:

1. Prescriptive methods for WDRC hearing aids differed substantially in prescribed gain, 15–20 dB across the whole frequency range and across input levels from 55 to 75 dB SPL, for an audiogram representative for first-time hearing aid users.

2. The differences in prescribed gain led to large differences in measured and calculated loudness, but only to minor differences in measured and calculated speech intelligibility. In the laboratory, and evaluated by hearing-impaired listeners without hearing aid experience, methods that prescribed high gain were preferred for low-level speech, whereas methods that prescribed low gain were preferred for high-level speech.

3. In the field, first-time hearing aid users clearly preferred a method that led to less-than-normal overall loudness, as calculated using the loudness model of Moore and Glasberg (1997), to a method that led to normal overall loudness, and these results were confirmed in laboratory tests.

4. Hearing-impaired listeners preferred considerably less-than-normal overall calculated loudness, as calculated using the loudness model of Moore and Glasberg (1997). The results were confirmed in both field and laboratory tests. There were no statistically significant differences between inexperienced and experienced hearing aid users in any of the experiments performed. The normal-hearing listeners preferred close-to-normal overall loudness. A problem with the loudness model was encountered. The prescriptive method used in the study, NAL-NL1, led to less-than-normal overall calculated loudness. Despite this, the hearing-impaired listeners rated loudness higher than the normal-hearing listeners.
9. Conclusions

The main result of the current study—that hearing-impaired listeners preferred considerably less than normal overall loudness—could be a result of the focus on people with mild to moderate hearing loss. Clients with this audiometric configuration constitute a majority of the clients who are provided with hearing aids in a typical Swedish hearing clinic. However, many evaluations of hearing aids and prescriptive methods have been made with people with more severe hearing loss.

Perhaps such a simple measure as to provide first-time hearing aid users with less gain than prescribed by the generic fitting methods we know today, will increase the number of satisfied hearing aid users.
Chapter 10

About the Papers

Authors: K. Smeds and A. Leijon.

Prescriptive methods for nonlinear hearing aids implemented in hearing aid manufacturers’ computer-based fitting software were compared for a speech-like measurement signal, and the results were interpreted using calculated loudness and speech intelligibility index.

Contribution of first author: The first author carried out the measurements and calculations (using existing software) and wrote the paper.

Paper II: Comparison of prescriptions for nonlinear (WDRC) hearing instruments—loudness, speech intelligibility and preference.
Authors: K. Smeds and E. Agelfors.

Three modified prescriptive methods for nonlinear hearing aids were compared both theoretically, using loudness and SII calculations, and in a laboratory study with 20 hearing-impaired participants.

Contribution of first author: The first author made the theoretical work, designed the experiment, carried out approximately one-third of the laboratory tests, analysed the data, and wrote the paper.
Paper III: Is normal or less than normal overall loudness preferred by first-time hearing aid users?

Author: K. Smeds.

Two prescriptive methods for nonlinear hearing aids, one leading to normal overall loudness and the other to less than normal overall loudness, were compared in a field test and accompanying laboratory tests.

Paper IV: Preferred overall loudness. I: Sound field presentation in the laboratory

Paper V: Preferred overall loudness. II: Listening through hearing aids in the field and in the laboratory

Authors: K. Smeds, G. Keidser, J. Zakis, H. Dillon, A. Leijon, F. Grant, E. Convery, and C. Brew.

Normal-hearing and hearing-impaired listeners participated in laboratory and field tests to determine their preferred overall loudness.

Contribution of first author: The first author collaborated with Keidser, Dillon, and Leijon in the overall design of the study. The experimental setup for the laboratory study was designed together with Keidser, who also participated in the design of the field test. In the field test, a research hearing aid, designed by Zakis, was used. Testing in the laboratory and field was carried out by the first author in collaboration with Grant, Convery, and Brew. The first author analysed the data and wrote the papers.
Chapter 11

References


11. References


11. References


11. References


