Review

Some characteristics of amplified music through hearing aids

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Abstract

Hearing aids are a relatively non-invasive means of reducing the negative effects of hearing loss on an individual who does not require a cochlear implant. Music amplified through hearing aids has some interesting characteristics but high fidelity is not typically one of them. This poses a serious problem for the investigator who wants to perform research on music with hearing impaired individuals who wear hearing aids. If the signal at the tympanic membrane is somewhat distorted then this has consequences for the assessment of music processing when examining both the peripheral and the central auditory system. In this review article on the subject of hearing aids and music, some of the acoustical differences between speech and music will be described. Following this, a discussion about what hearing aids do well and also less well for music as an input will be presented. Finally, some recommendations are made about what can be done for hearing-impaired individuals who wear hearing aids to listen to music.

This article is part of a Special Issue entitled <A window into the hearing brain>.

Abbreviations: A/DAnalog-to-Digital; AGCAutomatic Gain Control; ANSIAmerican National Standards Institute; dBDecibel; MPOMaximum Power Output; msecmillisecond; OSPL-90Output Sound Pressure Level with a 90 dB SPL input; SPLSound Pressure Level; WDRCWide dynamic range compression

1 Introduction

Sensorineural hearing loss has many negative effects on auditory perception. The result is that hearing-impaired individuals face numerous challenges which include decreased audibility, decreased dynamic range, decreased frequency resolution and decreased temporal resolution (Moore, 1996, 2007). Additionally, they often experience negative social effects such as increased isolation and withdrawal from social situations (Dalton et al., 2003; Strawbridge et al., 2000; Weinstein and Ventry, 1982). Amplification with hearing aids can address many of these concerns and has been shown to reduce the negative effects of hearing loss for those individuals who do not require a cochlear implant (NCOA, 1999; Chisolm et al., 2007; Kochkin, 2011).

Sensorineural hearing loss is most often described in terms of the effects it has on the perception of speech since difficulties with communication are a key reason why many hearing-impaired individuals, or their families, seek amplification or other rehabilitative interventions (Kochkin, 2012; Laplante-Lévesque et al., 2011, 2012). Hearing aids are therefore designed to amplify speech signals well, and this is of primary importance for the manufacturers of these devices. Amplification schemes for hearing aids are derived in terms of both audibility and comfort for speech (Cox and Moore, 1988; Keidser et al., 2011; Moore et al., 2010; Scollie et al., 2005). However, does this focus on speech really reflect the way that all hearing aid users live? There are many hearing aid users and potential users who require their hearing aids to amplify music well regardless of genre (Killion, 2009; Revit, 2009; Rutledge, 2009; Uys and van DijK, 2011; Uys et al., 2012), be they musicians or even enthusiastic concert goers.

Successful listening to music by a person with hearing loss involves many factors, including the nature of the input signal, the hearing aid processing, the signal at the output of the hearing aid, the auditory system (both peripheral and central) and the person's personal attributes such as musical training and experience. The focus of this review is on the music input signal and the hearing aid processing.

When exploring the subject of hearing aids and music it is important to first look at some of the acoustical differences between speech and music. After this, a discussion about what hearing aids do well and also less well for music as an input is necessary. Finally, some improvements and suggestions are made regarding adjusting hearing aids so that they can perform better for the hearing aid wearer when listening to or performing music.

2 Acoustic properties of music versus that of speech

2.1 Sound levels of music

Many hard-of-hearing consumers of hearing aids are requesting, and in some cases, demanding, improved fidelity of amplified music (Chasin, 2003; Revit, 2009). These requests may come from those people who like to listen to music on occasion, or those who actually play music. There may even be an audiological requirement to amplify the softer elements of music, and attenuate the higher levels of music, but in a pattern that amplifies music in a manner which the hard-of-hearing person remembers as being of high fidelity.

Whatever the audiological requirements, the hearing aid fitting goal remains the same: a musical signal that is both audible but not too intense, and one that has sufficient fidelity. This goal is not unlike that for speech, however, with music there are some additional aspects that need to be addressed. These additional requirements are based on the spectral nature of music and how it differs from that of speech.

Table 1 is adapted from Chasin (2006a) and is based on the spectral assessment of the musical instruments from over 1000 musicians. In all cases but one, the level measurements were made from a distance of 3 m ~~meters~~ on the horizontal plane. The one exceptional case is an additional set of measurements made at the left ear meatal opening for the violin players. In all cases, the average level of the musical instrument is far in excess of that which would be produced during normal conversational speech.

Table 1 Average sound levels of a number of musical instruments measured from 3 m ~~meters~~. Also given is the sound level for the violin measured near the left ear of the players. Adapted from Chasin (2006a). Used with permission.

Musical instrument	dBA ranges measured from 3 m ~~meters~~
Cello	80–104
Clarinet	68–82
Flute	92–105
Trombone	90–106
Violin	80–90
Violin (near left ear)	85–105
Trumpet	88–108

There are two issues that arise when considering the greater spectral levels of music versus those of speech. One is whether the music can produce hearing loss in the same vein as industrial noise can result in hearing loss. Speech ~~–~~ even shouted speech ~~–~~ does not achieve a level that can be damaging to one's own hearing. The same cannot be said of music and many studies have demonstrated the potential for permanent hearing loss from long term exposure (see for example, Axelsson and Lindgren, 1981; Behar et al., 2006; Camp and Horstman, 1991; MacDonald et al., 2008; Phillips and Mace, 2008; Poissant et al., 2012; Royster et al., 1991; Schmidt, 2011). These examples are from classical non-amplified music and the deleterious situation can be further enhanced with amplified music (see for example, Axelsson and Lindgren, 1978; Clark, 1991; Flugrath, 1969; Hart et al., 1987). The situation may be further complicated if the music has significant mid-and-high frequency sound energy. Furthermore, in cases such as playing a violin one ear may be exposed to a different level than the other because of head shadow effects. This may lead to an asymmetrical sensorineural hearing loss with the left ear being worse than the right ear (Schmidt, 2011) and this would be a complicating factor for providing amplification to these individuals via hearing aids.

The second issue related to the higher sound levels of music than speech is the capability of the hearing aid to transduce the higher input levels without significant distortion. As will be discussed in subsequent sections, this has direct ramifications for current hearing aid technology.

2.2 Spectral shape

There are differences in the spectral shape between speech and music that need to be accounted for when comparing these two signals. The spectral properties of speech are typically defined according to the long-term average speech spectrum (LTASS) (Dunn and White, 1940). The LTASS is defined in terms of a sample of natural running speech which contains all of the natural pauses between syllables and sentences. The measurement period of the LTASS is generally accepted as being more than a minute, as defined by Dunn and White (1940) or, more precisely, 64 ~~second~~s by Byrne et al. (1994). The LTASS is well defined and is relatively consistent between languages (Byrne et al., 1994). This is understandable because speakers of all languages have similar vocal tract lengths, similar oral and nasal cavity volumes, and similar mechanical and surface features. There is no inherent reason why a speaker from Japan should have a LTASS that is significantly different than that of a speaker languages have similar vocal tract lengths, similar oral and nasal cavity volumes, and similar mechanical and surface features. There is no inherent reason why a speaker from Japan should have a LTASS that is significantly different than that of a speaker from Peru. There are differing linguistically distinctive (phonemic) characteristics of various languages and these would direct a change in the amplification parameters of speakers of different languages (Chasin, 2012), however the input to a hearing aid microphone would be similar from language to language.

The spectral shape of the LTASS has a peak in the 500 Hz region and falls off at about 5–6 dB/octave above that. Low frequency sonorants (vowels, nasals, and liquids) have greater sound pressure levels than the higher frequency obstruents (affricates, fricatives, and stop consonants). Again this is true of all languages of the world. In contrast, the spectral shape of music can be similar to the LTASS or it can bear no resemblance. Non-vocal music is not constrained by the mechanical and physical characteristics of the human vocal tract. The shape of a music spectrum can be low-frequency emphasis, high-frequency emphasis, and anything in between. While vocal music has much of its spectral content in the lower and mid-frequencies, percussion instruments would generate primarily mid- to high-frequency energy. There is no single long-term average music spectrum that can be used for hearing aid fitting ~~formulae~~formulas. Because some sources of music have significant mid to high frequency content, the sound levels that reach each of the musicians' ears can be quite different.

2.3 Crest factor

Another difference between speech and music as an input to a hearing aid is the crest factor of the input signal. The crest factor is the difference between the average or RMS (Root Mean Square) amplitude of the signal and the instantaneous peak of the signal. The crest factor that is commonly used in the hearing aid industry is defined by the American National Standards Institute (ANSI) for testing hearing aids with a broad-band signal (ANSI S3.42, 1992). The studies on the levels of speech, from where the crest factor was derived, used analyses windows of 120 msec (Cox et al., 1988) and 125 msec (Dunn and White, 1940). This makes sense in that the time constants (temporal limitations) of our auditory systems are assumed to be on the order of 125 msec so shorter temporal analysis windows would not make sense.

Although highly modulated compared to steady state noise, human vocalizations are significantly damped within the vocal tract and this is independent of frequency. There are substantial constrictions in the oral and nasal cavities, a narrow opening in the velo-pharyngeal port (connecting the oral and nasal cavities), a significant amount of tissue in the turbinates of the nasal cavity, and soft buccal walls and lingual structures (Johnson, 2003). In short, the human vocal tract is highly damped such that the difference between the RMS of a generated signal and its instantaneous peak is relatively small. In contrast, the output waveform of music is “peakier” relative to speech because of the lower level of damping inherent in most musical instruments, therefore, the crest factor for music can typically be on the order of 16–18 dB whereas that for speech is assumed to be only about 12 dB (Cox et al., 1988).

The typical values of the crest factor for speech (12 dB) and music (16–18 dB), when added to the RMS values of speech and also to music (see Table 1), suggest that the input to the hearing aid can be quite high, especially for music. The LTASS levels plus a crest factor of 12 dB does not stress the various hearing aid components. Even the peaks of shouted speech as an input will be well within the optimal operating range of currently available digital hearing aids. The same cannot be said of many sources of music where the average levels plus the crest factor can exceed the input capability of most hearing aids. Table 2 shows some measured crest factors of some speech samples and some music samples.

Table 2 Shows some measured crest factors of some speech samples and some music samples. Adobe Audition 1.0 (San Jose, California) was used to calculate the difference between the average value of the signal and the instantaneous peak.

	Speech #1	Speech #2	Speech #3	Music #1	Music #2	Music #3
Crest factors (dB)	11.3	12.2	12.2	16.45	18.1	16.95

This problem can be further exacerbated if the crest factor was even greater than 16–18 dB. In Table 2 the crest factor was calculated using a 125 msec analysis window. Hearing aid circuits are designed to amplify speech with technical features that are designed to fall within the temporal limits of the auditory characteristics of the cochlea. The 125 msec analysis window is consistent with this line of reasoning. However when it comes to the input of a hearing aid, no such limitations exist, very brief musical signals can be received on the order of 50 msec in duration.

Using shorter temporal analysis windows than 125 msec for music may provide a more accurate estimate of the input to a hearing aid. Recall that the crest factor is the difference between the instantaneous peak and the average or RMS peak. Shorter temporal analyses than 125 msec will result in higher instantaneous peaks with a resulting higher crest factor. A complex signal that is analyzed with a 125 msec window may indeed have a 16 dB crest factor, but the same sample if analyzed with a 50 msec window may result in a 20 dB crest factor. This is an “input issue” to the hearing aids and has nothing to do with the temporal characteristics of our auditory system.

Table 3 demonstrates that the crest factor for the same music (#1) sample is a function of the time analysis window. Using the traditional 125 msec analysis window the crest factor is on the order of 16 dB but it is 21 dB if a shorter window is used. The shortened window will better estimate the instantaneous peak intensity component to the crest factor. Adding the value of the crest factor, with a 50 msec analysis window, to the values of music from Table 1 further demonstrates that the inputs of music to a typical digital hearing aid may result in levels beyond the optimal operating range of the device. The danger is that these high levels may in fact introduce unwanted distortion into the amplification chain which may be perceptible by the hearing-impaired individual wearing the device.

Table 3 For the same music sample (Music #1 in Table 2), the difference between the RMS of the signal and its instantaneous peak is given. For shorter analysis windows the instantaneous peak is higher than for longer windows of analysis with a resulting higher crest factor. Adobe Audition 1.0 (San Jose California) was used to calculate the difference between the average value of the signal and the instantaneous peak.

Analysis window (msec)	500	400	300	200	125	100	50	25
Crest factor (dB)	16.45	16.43	16.45	16.44	16.45	18.22	21.68	21.68

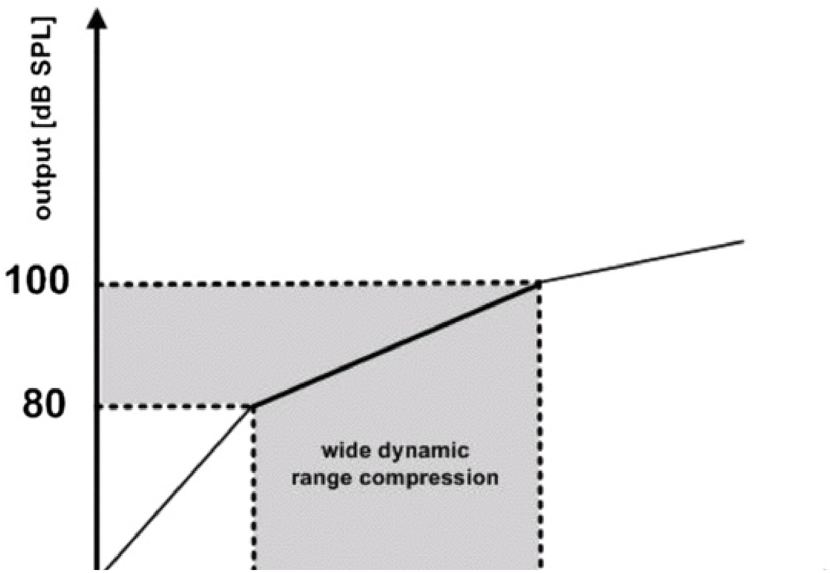
3 How well can hearing aids work with music or speech as an input?

If hearing aids are designed primarily for speech inputs, then how well do they function for music? Hearing aids must first be seen as amplifiers, their primary purpose is to apply gain to an input to establish a desired output. This can be frequency-specific and level-specific to accommodate the needs of the hearing aid wearer to obtain comfort and audibility in a wide range of listening environments.

The amplification that is needed is typically prescribed by the fitting rationale that is selected. From the 1940's through the 1980's the fitting methods for hearing aids were linear (Venema, 2006), so the same gain was applied to the input regardless of the input level. It was therefore necessary that the hearing aid had a volume control in order for the user to adjust the level depending upon the environment that he or she might find themselves in. The problem with this is that the hearing aid wearer has to potentially adjust the hearing aid almost constantly in a complex environment such as when talking with a colleague on a busy street. Making almost constant volume control adjustments are not convenient and are especially unrealistic for a young child, for example (Kuk, 2002). Could the volume control actions be automated or the need to use it be reduced? The solution was to apply compression, commonly known as an Automatic Gain Control (AGC) as a means to adjust the volume automatically. With compression the amount of amplification applied to an input decreases as the sound level increases (Dillon, 2012; Hudgins et al., 1948; Steinberg and Gardner, 1937). As AGC systems became available, first in the analog domain, then in the digital domain, it was possible to selectively apply gain depending upon the input, so the fitting methods were now revised to be non-linear.

The most common implementation of compression technology in the digital domain is called wide dynamic range compression (WDRC) which allows the hearing aid to fit the prescribed non-linear targets developed in terms of audibility and comfort (Keidser et al., 2011; Scollie et al., 2005; Moore et al., 2010). WDRC can apply more amplification to weaker signals than to stronger signals so that the large dynamic range at the input can be reduced into a smaller dynamic range at the output (Moore, 1996). WDRC is far more complex than the simple AGC functions described earlier. The advantages of digital implementations of WDRC are that it is possible to have precise control over many compression parameters such as the design and use of filter characteristics (Kates, 2005; Levitt, 2007; Schaub, 2008).

A simplified view of WDRC is shown in Fig. 1. There are two threshold kneepoints, which are points at which the slope of the input–output function changes; signifying where the compression function changes. The first kneepoint (1) shows the change from linear amplification to compression, while the second kneepoint (2) shows the application of more compression as the aid approaches its output limits. In between these two kneepoints an input of 40 dB is compressed into a dynamic range of 20 dB, resulting in a compression ratio 2:1 (also read as “2 to 1”).



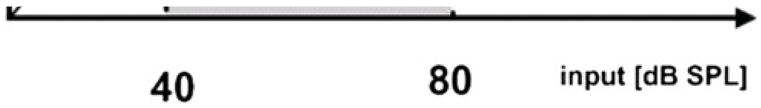


Fig. 1 Input/output function in dB SPL illustrating the concept of wide dynamic range compression within the shaded ~~grey~~ gray area. As the input signal is increased (x-axis), there is an increase in the output (y-axis). An input of 40 dB is compressed into a dynamic range of 20 dB. 40/20 = compression ratio 2:1. The kneepoints 1, and 2 indicate the changes to the amplification function.

The application of compression can limit overly intense outputs so that the hearing aid wearer is not disturbed by amplified signals that are above his or her tolerance for loud sounds. Compression in this case will also prevent the hearing aid amplifier from generating harmonic distortion due to peak clipping. The implementation of compression across the various manufacturers of hearing aids can differ quite significantly and so clear conclusions about the settings that are best for music are quite difficult to make at the current time (Chasin and Russo, 2004; Davies-Venn et al., 2007; Moore, 2008, 2012b). More research is definitely required to determine if the compression characteristics in hearing aids for music should be similar to speech.

In order for the hearing aid to be comfortably worn by any individual two common complaints need to be addressed: the occlusion effect and acoustic feedback. Both of these complaints can be reduced via the use of acoustic feedback cancellation systems. The occlusion effect is generated by body (bone) conducted sound. Typically the subjects' own voice or the sounds of chewing are judged to be unnaturally loud and disturbing when a hearing aid is blocking or even partially blocking the external ear canal (Dillon, 2012). To counteract occlusion an easy method is to provide acoustic venting in the fitting which is accomplished by the use of a small sound channel for air ventilation (Lybarger, 1980; Carle et al., 2002; Kiessling et al., 2005). When the hearing aid is sufficiently vented, this decreases the transduction of self-generated low frequency energy by allowing it to exit while at the same time the vents allows unobstructed sound in to the ear, especially for the lower frequencies. The free flow of low frequency information can improve the perceived sound quality for the hearing aid wearer (Cox and Alexander, 1983; Kuk, 1991) and this is important for listening to music. There is a drawback to venting, however: as the size of the vent increases, there is a greater chance of acoustic feedback.

Acoustic feedback results when the amplified signal from the receiver (speaker) is picked up the microphone and then is re-amplified, adding to the initial signal if it is in phase (Kates, 2008). The aid is in a state of oscillation and an acoustic “howl” or “whistle” is produced that is very disruptive to the hearing aid user and anyone else who is in the immediate proximity. The use of feedback cancellation systems dramatically reduces the incidences of acoustic feedback so the hearing aid fitting can have a large vent diameter while still providing the appropriate amount of more gain. These systems mostly involve the use of phase cancellation systems where by the feedback signal is detected and classified as acoustic feedback and then the hearing aid produces a similar signal that is out of phase. The subsequent addition of these two signals together results in a 0 sum, and therefore the feedback signal is ~~cancelled~~canceled. Feedback cancellation systems need to be sensitive enough to detect sub-oscillatory acoustic feedback which is not strong enough to produce an audible whistle but may be perceived as an echo or another distortion. Sub-oscillatory feedback may affect the ability of the hearing aid to provide appropriate high frequency gain (Cox, 1979; Levitt, 2007). The implementation and effectiveness of feedback cancellation systems can vary greatly between hearing aids as has been shown by Ricketts et al. (2008) and Spriet et al. (2010). These findings indicate that broad generalizations about the effectiveness of feedback cancellation systems cannot be made. In some cases, feedback cancellations systems can try to remove prolonged periodic signals similar to that produced by musical instruments (Dillon, 2012). They can also, in rare circumstances, produce other audible artifacts (Kates, 2008; Dillon, 2012). In these cases, if the perception of music is affected then the feedback cancellation systems may need to be disabled with the result that the hearing instrument gain will need to be reduced. We will return to this issue within the clinical recommendations at the end of this article. In summary, the use of feedback cancellation systems enable the aid to provide sufficient gain while also providing venting that reduces the occlusion effect and creates a natural path for unamplified sound to travel. These factors are all important for the hearing-impaired individual to successfully use the hearing aids for both speech and music.

As a by-product of the circuitry within the hearing aid there is an inherent low-level noise that can be audible when the hearing aid user is in a very quiet environment, or has good low-frequency hearing (Venema, 2006). Changes in this circuit noise can be seen as a function of stimulus level and can be attributed to either the microphone, the signal level or the application of adaptive features (Lewis et al., 2010). It can be reduced or even eliminated by the use of expansion (the opposite of compression) circuitry employed within the hearing aid that can apply less gain to soft sounds and therefore reduces the circuit noise.

In addition to suppressing internal noise, expansion can also reduce soft environmental sounds such as that produced by ventilation systems. Expansion thresholds and time constants need to be applied carefully to ensure that the perception of soft speech cues is not affected (Plyler et al., 2005a, 2005b, 2006). Live music, even played quietly (pianissimo), should not be affected by the expansion system. However, recorded music may be affected but the level would need to be much less than 50 dB SPL to have any effect.

One concern about sound quality especially with regards to music is the effect of throughput delay. This delay refers to the sum of delays inherent in the signal path of a digital hearing aid and usually is less than 10 ms (Dillon et al., 2003). Hearing aids apply a lot of processing to the auditory signal and so this naturally takes some time. When the hearing aid fitting is vented, then unamplified sound can pass through the vent and subsequently arrive at the tympanic membrane before the amplified signal. The result is that there are two signal paths that may interact, the natural path and the amplified path which may be out of phase due to the throughput delay (Dillon, 2012). Investigations examining delays in the range of 1~~to~~–10 ms in simulation have demonstrated negative effects on sound quality as judged by normal and hearing-impaired hearing listeners (Stone and Moore, 1999, 2002; 2003, 2005; 2008). These negative effects delays in the range of 1 ~~to~~ – >10 ms in simulation have demonstrated negative effects on sound quality as judged by normal and hearing-impaired hearing listeners (Stone and Moore, 1999, 2002, 2003, 2005, 2008). These negative effects were not evident when tested with real hearing aids worn by hearing-impaired listeners (Groth and Sondergaard, 2004). Zakis et al. (2012) examined the effect of throughput delay on the sound quality of music in real hearing aids with trained musicians. Differences in sound quality could be described by the musicians for each delay condition; however, no significant difference was found between preferences assigned to each delay condition compared to the no-delay condition.

4 What hearing aids cannot do for music as an input

In this discussion of hearing aid technology and music we have explored what hearing aids can do for music and speech as an input source. We will now move on to discuss some limitations, and potential solutions for the use of digital hearing aids with music.

When the hearing loss affects the outer hair cells, the tuning for the inner hair cells becomes broader (Halprin, 2002; Moore, 2007). This broadening of the tuning curves results in a reduction of the frequency resolution within the cochlea. Hearing aids, after all, just provide amplification and they cannot currently directly compensate for this widening of the auditory filters. Frequency lowering systems, where information in a region of most cochlear damage is moved into a region where the cochlea can adequately resolve it, could be employed to address this issue and other conditions that involve the loss of inner hair cells (Dillon, 2012). However, the effects of different frequency lowering implementations on the perception of music need to be further researched~~.~~ (Uys et al., 2012; Parsa et al., 2013).

Another area where hearing aids have difficulties is that the hearing aid environmental detection systems cannot distinguish accurately between signals with the same spectral content, so the ability to detect and react reliably to music of many different genres is not currently possible. Some progress has been shown in the detection of single instruments and singing, for example, however more work needs to be done in order to reliably detect more popular styles of music (Büchler et al., 2005). In contrast, hearing aids can detect speech very reliably due to the well-defined acoustic properties that speech possesses (Schaub, 2008). Finally most hearing aids with 16 bit digital architecture, which accounts for most hearing aids currently available today, cannot transduce signals that are in excess of 95 dB SPL without encountering compression and distortion before the analog-to-digital (A/D) converter. It is therefore important to now look at the conversion of a signal from the analog domain to the digital domain.

4.1 Analog-to-digital conversion

So far in our discussion of hearing aid technology we have not seen any “show stopping” limitations when it comes to music as an input source. There is however one area of great concern that is hardware based and is not a software issue – this is in the conversion from the analog domain into the digital domain. Most digital hearing aids compress the peaks of an input when they reach 95 dB SPL before the A/D conversion. This is based on the 16 bit A/D conversion architecture that is employed by most of the hearing aids currently in use (Agnew, 2000, 2002; Hamacher et al., 2005; Edwards, 2007). Applying compression at 95 dB SPL and above before the A/D converter in most cases is more than adequate even for loud speech (Olsen, 1998). However, depending on how the compression is implemented, the resulting amplification of music can potentially sound compressed, unnatural, and even slightly distorted. To find out whether the hearing aid can in fact handle the intense peaks of live music there is a simple test that can be made within a typical commercial hearing aid test box that is commonly found in a clinician's practice (Chasin, 2006b). The hearing aid in question needs to be programmed with only 5–8 dB of gain, but with the maximum possible output setting. An intense input on the order of 100–110 dB SPL needs to be used. If a particular hearing aid can process the more intense elements of music, then the intense input + low gain (5–8 dB) should not cause any measured distortion in excess of 10%. Since the maximum output setting is programmed to be very high, the input + gain << than the maximum output so any measured distortion is in the front end A/D converter and is not a saturation effect.

There is nothing that can be done via hearing aid fitting software to correct or even reduce the effects of a low input compression threshold on the musical signal before the A/D converter. To truly accommodate the amplitude peaks of live music a digital hearing aid with an A/D conversion system of at least 20 bit word lengths must be used to obtain a potential dynamic range of 120 dB. Hearing aid integrated circuits need to be efficient with regards to battery drain and so ensuring that the digital word length is appropriate for speech while minimizing battery drain is very important. Battery efficiency is one of the main reasons why 20 or even 24 bit A/D conversion is not widely seen in hearing aids that are currently available today (Kates, 2008). The question still remains though, as to what can be done to improve current digital hearing aid systems that use 16 bit A/D converters?

4.2 Overcoming the current limitations of dynamic range

To be able to handle the intense peaks of music with 16 bit digital architecture the answer lies in using a 16 bit A/D converter but shifting the maximum input level to where the hearing aid works more linearly, so that the Automatic Gain Control input compression (AGC_{input}) is not applied until the level from the microphone exceeds approximately 110 dB SPL (Chasin, 2003). Modern hearing aids have a feature (called AGC_{input}) where the gain can be altered automatically based on the detected level of a~~n~~ (input) signal. This basic idea of raising the compression knee-point was implemented, in a commercially available hearing aid in 2010 (Hockley et al., 2010; Hockley et al., 2012). The outcome is that the majority of the peaks in amplitude of live music are not compressed before the A/D converter.

In Fig. 2 the behavior of the AGC_{input} is shown with an input/output diagram. In the reference situation (~~grey~~ gray line) the threshold of the AGC_{input} becomes active at 95 dB SPL whereas the black line shows the level-shifted condition that has its threshold at 110 dB SPL. This measurement was made with a 1000 Hz pure tone sinusoidal sweep signal.

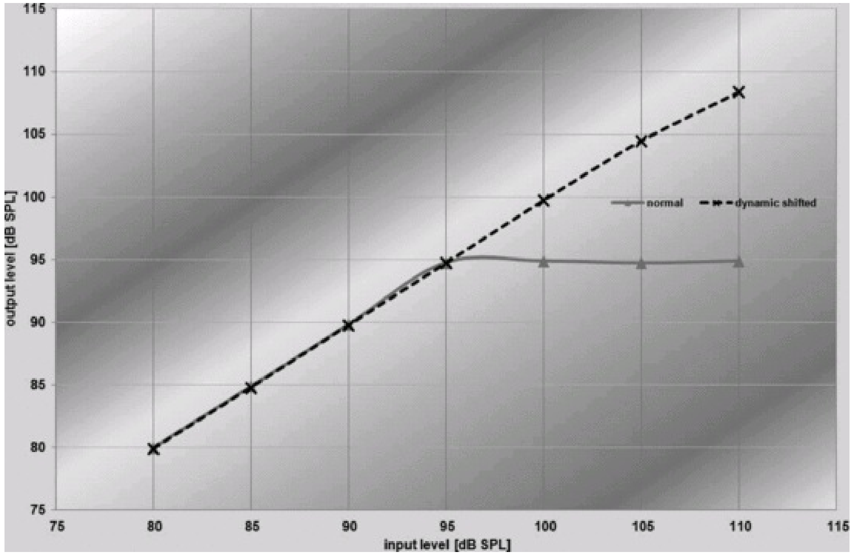


Fig. 2 Input/output function with the level-shifting processing on (dynamic shifted) and off (normal) for a 1 kHz sinusoidal signal.

A more ecologically valid stimulus is a piece of classical music, in this case an excerpt from *Eine kleine Nachtmusik* (Serenade No. 13 for strings in G major), K. 525 by Mozart (CD EMI records) played at 110 dB SPL (peak). The first waveform in Fig. 3 shows the music with the threshold for compression before the A/D converter set to 95 dB SPL. It is easy to see from this waveform that the resulting output (in ~~grey~~gray) is compressed in comparison to the original recording (in black). In contrast in Fig. 4 the compression input is level shifted to 110 dB and this compressed effect is not seen. It is clear to see that the dynamic range of the music is preserved. The natural dynamic characteristics of the music will be converted into the digital domain.

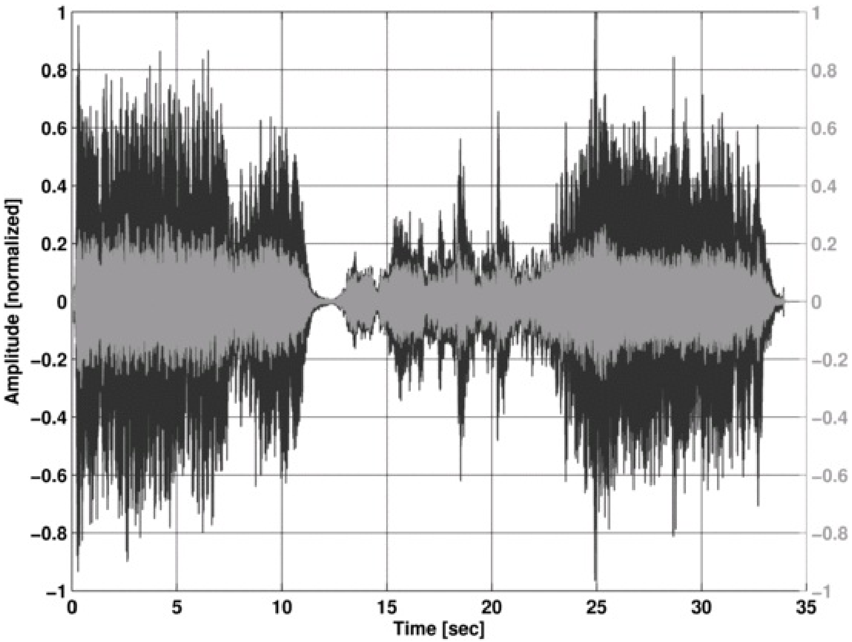


Fig. 3 Recording with the reference processing; input level 110 dB SPL. The black waveform is the original input file.

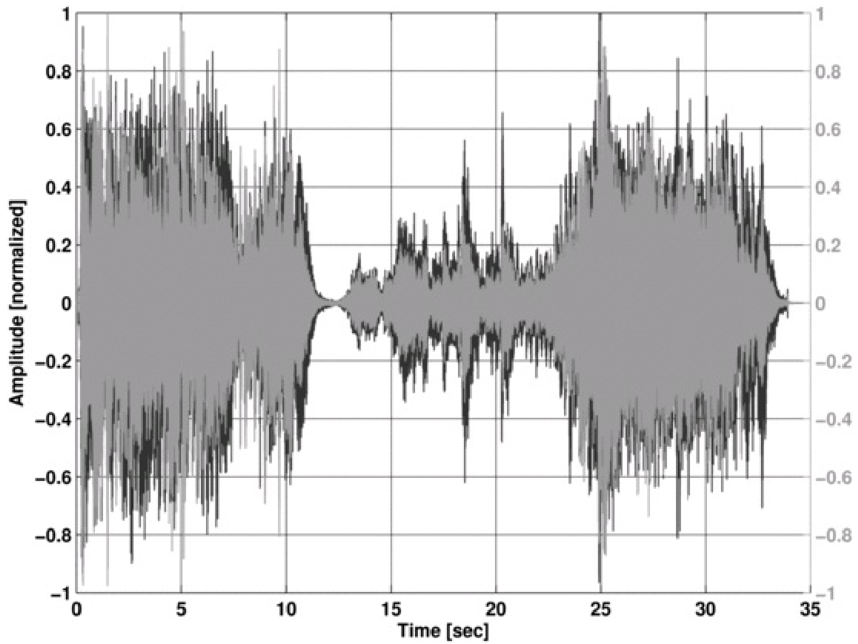


Fig. 4 Recording with the level shift; input level 110 dB SPL. The black waveform is the original input file.

It is very important to emphasize that the level shift modification is occurring in the front end of the hearing aid, before the A/D converter and thus before any amplification is applied. The peaks of music are not increasing the output levels and therefore the maximum power output (MPO) is not changed. The MPO is always set based on the real or calculated uncomfortable loudness levels (UCL) within the hearing aid to avoid any potential issues with the hearing-impaired individual's residual hearing.

The question naturally arises regarding how this modification to the hearing aid is received by hearing impaired individuals who listen to, or perform music. Hockley et al. (2010) conducted a study which looked at the ratings of sound quality attributes by 9 professional musicians (8 males and 1 female). Four of these musicians were woodwind players (clarinet, saxophone, and flute); 3 played jazz, while the other was a classical musician. Three of the musicians were classical violinists who also played the viola. The final two musicians were both rock (electric) guitarists. The attribute scales used by Hockley et al. (2010) were based on the work of Gabrielsson et al. (1974), Gabrielsson and Sjögren (1979) and Gabrielsson et al. (1991) and Cox and Alexander (1983). The scales consisted of qualitative descriptions of sound quality. Each subject gave a numerical rating towards the attribute that best suited what he/she experienced. Fullness is an example of an attribute that was used, where the perceptual dimension is from “full” to “thin”. Another example of an attribute that was measured is for naturalness, where the perceptual dimension is from “true to the source” to “artificial”. The subjects were asked to compare, within the same hearing aids, a program that applied the level shift with a standard program that did not.

Overall, for the judgment of fullness, a program with the level shift was judged to be significantly fuller than for the standard program without the level-shift. Overall fidelity for the level shifted program was judged to be significantly better than for the standard program. In this small investigation it was concluded that the level shift contributed to a better rating of sound quality for these musicians.

5 Some technologies and practical strategies to handle the more intense inputs of music

There are a number of creative strategies and electro-acoustic techniques that are being used in the hearing aid industry to resolve this “high level input- front end” problem for music (Chasin, 2010). The following is a list of seven of the more commonly used approaches that have been useful. The first four are strategies that can be shared with clients who already have hearing aids that are optimal for speech but which require additional fidelity for listening or playing music. The next three are technical innovations that are currently available in hearing aids or can be implemented within a clinical setting. While each of these strategies and technical innovations have been used successfully in the clinic with hard of hearing clients, more research is required to further verify the integrity and the scope of function of these approaches.

5.1 Four strategies that can be used with the hearing-impaired persons current hearing aids

5.1.1 Strategy #1: turn down the input and turn up the hearing aid volume (if necessary)

5.1.1 Strategy #1: turn down the input and turn up the hearing aid volume (if necessary)

If the excessive level of the input to the hearing aid does cause distortion of the A/D converter then turn down the input if at all possible. If traveling in a car, turn down the level of the sound system and, if necessary, turn up the level of the hearing aid to compensate. The output limiting of the device will ensure that the output will be the same, but the input would have been reduced to a level that is well within the operating range of the front end of the hearing aids.

5.1.2 Strategy #2: removal of the hearing aid for music

While this strategy may be at first seem to be counterintuitive given the discussion so far, it is definitely an option. Given the higher level inputs of music, the required gain may be close to 0 dB for a desired output. Table 4 shows some data derived for a range of severities of hearing losses at 1000 Hz and the required gains for speech and for music. Even for an 85 dB HL sensorineural hearing loss at 1000 Hz, while a person may require 45 dB of gain for certain speech sounds, they may only require several decibels of amplification for many types of music. The best strategy for many hard of hearing people may be to simply remove their hearing aids when listening or playing music.

Table 4 Calculated amounts of gain required for a given hearing loss at 1000 Hz (column 1) based on FIG6 (Gitles and Niquette, 1995). For average levels of music (95 dB A) inputs, virtually no amplification may be required even for very significant hearing losses. Used with permission. www.hearinghealthmatters.org/hearthemusic blog, Chasin, 2011). Downloaded Sept. 21, 2012.

dB HL at 1000 Hz	65 dB input	80 dB input	95 dB input
15	0	0	0
25	2	1	0
35	8	4	0
45	14	7	0
55	20	10	1
65	28	15	2
75	36	20	3
85	44	24	4

5.1.3 Strategy #3: use a tape to cover the hearing aid microphones

This is the lowest technology level and is perhaps the easiest to implement clinically. Like the use of a less sensitive microphone, 6 dB/octave for example (Schmidt, 2012), using a temporary microphone covering such as Scotch tape shifts the microphone's ability to transduce sound downwards by about 10 dB for three layers of Scotch tape (3M Corporation, Minneapolis, MN). The A/D converter is therefore presented with a signal that is 10 dB less intense and can often be within its optimal operating range. There needs to be some trial-and-error with this approach so therefore the hearing aid user may need to be instructed to play with one, two, or three pieces of tape over both hearing aid microphones. The exact number does depend on the gauge and the brand of the tape. Attenuations of 10 dB which are relatively flat across the frequency range have been measured by the first author using this clinical “low tech” approach.

5.1.4 Strategy #4: change the musical instrument

Changing the musical instrument is a common strategy used by many musicians. The musician could change to an instrument that has more of its energy in an audiometric region of better hearing. Many violin players have switched to the viola which is a fifth lower in frequency. For many this is a simple approach that has extended a musicians' enjoyment of their music for many years.

5.2 Three examples of current hearing aid technology that can be used to amplify music

The next three sections are about hearing aid technologies that have been shown to be quite useful. This review is not meant to be technically complete. This is not an exhaustive list, but includes technologies that have, to date, been found to be clinically useful. As stated earlier, although these techniques have been shown to be useful in a clinical situation, more research is required to verify the function and scope of these approaches.

5.2.1 Technology #1 – analog K-AMP technology

Although no longer widely commercially available, the analog K-AMP was first manufactured in 1988 (Killion, 1988, 1990; 1993), and it has been the mainstay for musicians since its inception. It was designed with the capability of being able to transduce very intense inputs with virtually no distortion. And because it is analog, there is no A/D converter to be overdriven. Some hearing aid manufacturers have taken a valuable lesson from the K-AMP hearing aid and are in the process of creating a “hybrid” device that has an analog input stage that compresses the input levels and brings them into the optimal operating range of the A/D converter. There is an equivalent, but digital, expansion phase after the A/D converter. This technology can be thought “hybrid” device that has an analog input stage that compresses the input levels and brings them into the optimal operating range of the A/D converter. There is an equivalent, but digital, expansion phase after the A/D converter. This technology can be thought of as ducking under a low hanging bridge and then standing up again. There are some “third party” manufacturers of this type of circuitry that seek to “auto range” the inputs in order to ensure that the signal does not overdrive the A/D converter. These third party manufacturers have been supplying the hearing aid industry with this technology for more than a decade.

5.2.2 Technology #2: changing where the dynamic range of the hearing aid operates

Another technological approach that was discussed in detail earlier in this review is to change where the dynamic range of the hearing aid operates. This is an approach that is based on the actual definition of dynamic range. The theoretical dynamic range of current 16 bit hearing aids is 96 dB (and not 96 dB SPL). It is a range between the least intense signal and the most intense signal and is 96 dB (without any scale). In this approach, when the circuit is implemented, by the selection of a music program, it transduces all inputs between 15 dB SPL and 111 dB SPL – still a 96 dB dynamic range – but it has been shifted up by 15 dB. Levels of 111 dB SPL can be transduced distortion free and are much better for the listening to, or the playing of, live music.

5.2.3 Technology #3: the use of a different microphone

The use of −6 dB/octave microphone instead of a broadband microphone has been shown to be quite beneficial with many forms of music (Schmidt, 2012). As the name suggests, the hearing aid microphone has been made less sensitive to the more intense lower frequency components of music: specifically, 6 dB less sensitive at 500 Hz and 12 dB less sensitive at 250 Hz. This approach will not change the fidelity of the higher frequency elements of music, but since most of the intense components of music are below 1000 Hz, this “fools” the A/D converter into thinking that the input is well within its operating range. A draw-back of using a −6 dB/octave microphone is that it does increase the internal noise floor of the hearing aid. However, expansion can be used successfully in its maximum setting to offset this change in noise floor as was discussed in our earlier discussion of this circuitry.

It is important to state that none of these strategies or approaches are software adjustments. Software changes occur after the A/D converter and once an intense signal is distorted by a poorly configured front end, no amount of software manipulation will ameliorate the situation. Fitting software modifications are simply not the approach that should be taken when dealing with the more intense components of music.

6 General recommendations for an “optimal hearing aid for music”

Assuming that the clinician has been able to select or configure a hearing aid to receive the more intense components of music with minimal distortion, what are some of the optimal software and electro-acoustic settings that can be made for music? There are four general clinical recommendations.

6.1 Recommendation #1 – similar WDRC parameters for speech and for music

This is an area where more research is required as was discussed earlier but it is quite possible that no changes would need to be implemented for a “music program” which are different to a “speech in quiet program”. This was perhaps the case in the past because some hearing aid circuits used a peak detector versus an average or RMS detector. Because of the differing crest factors of music and speech this detection point for activation should have been different for the two stimuli, but with the current trend to design primarily with an RMS or average level detector, any differences are obviated. The use of the WDRC circuitry is primarily an attempt to re-establish normal loudness growth due to outer hair cell damage, and indeed that is what it does. Then the use of this circuit primarily addresses damage to the auditory system rather than the nature of the input stimuli per se~~.~~ (Davies-Venn et al., 2007). Chasin and Russo (2004) suggested that “WDRC… may be better for music…. That hypothesis was supported by the present data.” (p. 696).

6.2 Recommendation #2: gain settings within a dedicated music program

The “music program” should be set with about 6 dB lower OSPL90 and 6 dB lower gain than the client's “speech in noise” program. The OSPL90 is a measure of the most intense output that is possible given a 90 dB SPL input signal to a hearing aid. This “−6 dB rule” is based on the fact that many forms of music have a crest factor that is about 6 dB greater than that of speech, however as seen in Table 2, this can vary significantly. For example, if the crest factor of music is 18 dB and that of speech is 12 dB, then the peaks of music are 6 dB more intense (18 ~~dB~~–12 dB) than those of speech for a given presentation level. Therefore, in order to prevent the peaks of music from causing discomfort, the OSPL90 and, assuming similar WDRC parameters for speech and music, the gain should be 6 dB less intense than for the “speech in quiet” program setting.

6.3 Recommendation #3: bandwidth for a music program

Examining the work of Moore et al. (2011), Moore (2012b) as well as the work of Ricketts et al. (2008) several general recommendations can be made. If the hearing loss is mild, and at most up to a moderate level, then a broader bandwidth for music is better. If however the hearing loss is greater than a moderate level, then less may be more - a narrower bandwidth (which can avoid dead regions in the cochlea) may provide a more pleasant sound than a wider bandwidth that extends into the high frequency region. The same can be said about the configuration of the audiogram- a person with a relatively flat audiometric configuration should have the widest bandwidth possible. In contrast, if the audiogram has a precipitous high frequency loss configuration then again, less may be more and a narrower frequency response would be ideal.

6.4 Recommendation #4: turn off the feedback cancellation and noise reduction systems

Noise reduction systems and feedback cancellation systems are designed to remove acoustic information that is judged to be undesirable, or even uncomfortable, for the hearing aid user. For this reason a general recommendation can be made to disable the feedback cancellation systems and noise reduction systems, to the extent that they can be disabled, so they do not affect the musical signal. This is not a well-researched area but clinical experience by the first author suggests that some feedback cancellation systems can "turn off" the hearing aid while listening to or playing music. In some instances, the pure tone nature of harmonics in music can be confused with the pure tone like nature of a feedback signal. This is especially true in the higher frequencies. To help resolve this in cases where the feedback cancellation system cannot be disabled, some manufacturers have limited the feedback circuit to the higher frequency region and this is a reasonable solution to an otherwise problematical situation. The use of a feedback cancellation system with a dedicated music program should definitely be applied on a case to case basis depending upon the needs of the hearing-impaired individual. The benefits that the feedback cancellation system provide, such as provision of a large vent for the natural entry of sound into the ear canal, may override any artifacts generated by the feedback cancellation system.

7 Conclusions and recommendations for further research

Table 5 shows the one major problem concerning the capability of modern digital hearing aids to handle the more intense elements of music. The limitation is the A/D converter. Expending effort on designing better microphones, amplifiers, and D/A converters would not appreciably improve the fidelity of amplified music. The same is true about software programming manipulations unless the limitations of the A/D converter stage of amplification is first addressed.

Table 5 Summary table showing the one electro-acoustic element in modern digital hearing aids where music as an input is typically beyond the optimal operating range. Hearing aid microphones can handle very intense inputs and this has been the case since 1988. Amplifiers, D/A converters and output stages are also well designed to handle the characteristics of both speech and music.

	Speech	Music
Microphone	Ok	Ok
A/D converter	Ok	Problem
Amplifier	Ok	Ok
D/A converter	Ok	Ok
Output stage/receiver	Ok	Ok

Most of the strategies and technologies that have been discussed are related to the finding that most currently available digital hearing aids cannot handle the more intense inputs of music within their optimal operating range. A study of crest factors that are relevant to the input of a hearing aid, rather than the output to our auditory systems, may have far reaching implications for music listening. Like most areas within the field of audiology the realm of music as an input to hearing aids and the technologies that are available is a rapidly changing one. New technologies are on the horizon and many similar ones may be implemented by various manufacturers under a score of different names. Further research should also be conducted to create questionnaires and other tools to establish the needs and requirements of hearing-impaired individuals when to listening to music through their hearing aids (Rutledge, 2009; Uys and van DijK, 2011). Finally, to best of the authors' knowledge there is no work published to date that directly investigates the central auditory system with music as an input via hearing aids. We sincerely hope that this research will also be pursued in the future.

Uncited reference

Moore, 2012a.

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References

Agnew J., Digital hearing aid terminology made simple: a handy glossary, *The Hearing Journal* **53** (3), 2000, 37–43.

Agnew J., Amplifiers and circuit algorithms for contemporary hearing aids, (Chapter 3)In: Valente M., (Ed), *Hearing Aids: Standards. Options and Limitations,* second ed., 2002, Thieme; New York NY.

American National Standards Institute, American National Standard: Testing Hearing Aids with a Broad-band Noise Signal, ANSI S3.42–19921992, American National Standards Institute; New York.

Axelsson A. and Lindgren F., Hearing in pop musicians, *Acta Otolaryngologica* **85**, 1978, 225–231.

Axelsson A. and Lindgren F., Hearing in classical musicians, *Acta Otolaryngologica* 1981, 3–74, (Suppl. 377).

Behar A., Wong W. and Kunov H., Risk of hearing loss in orchestra musicians: review of the literature, *Medical Problems of Performing Artists* **21** (4), 2006, 164–168.

Büchler M., Allegro S., Launer S. and Dillier N., Sound classification in hearing aids inspired by auditory scene analysis, *EURASIP Journal on Applied Signal Processing* **18**, 2005, 2991–3002.

Byrne D., Dillon H., Tran K., et al., An international comparison of long-term average speech spectra, *Journal of the Acoustical Society of America* **96** (4), 1994, 2108–2120.

Camp J.E. and Horstman S.W., Musician sound exposure during performance of Wagner's Ring Cycle, *Medical Problems of Performing Artists* **7** (2), 1991, 37–39.

Carle R., Laugesen S. and Nielsen C., Observations on the relations among occlusion effect, compliance and vent size, *Journal of the American Academy of Audiology* **13**, 2002, 25–37.

Chasin M., Music and hearing aids, *The Hearing Journal* **56** (7), 2003, 36–41.

Chasin M., Hearing aids for Musicians, *Hearing Review* **13** (3), 2006a, 11–16.

Chasin M., Can your hearing aid handle loud music? A quick test will tell you, *The Hearing Journal* **63** (12), 2006b, 22–24.

Chasin M., Six ways to improve listening to music through hearing aids, *The Hearing Journal* **63** (9), 2010, 27–30.

Chasin M., Ajustements spécifiques de gain des appareils auditifs pour les finales de phrases de certaines langues autres que l'anglais, [Sentence final hearing aid gain requirements of some non-English languages], *Canadian Journal of Speech-language Pathology and Audiology* **36** (3), 2012, 196–203.

Chasin M. and Russo F.A., Hearing aids and music, *Trends in Amplification* **8** (2), 2004, 35–47.

Chisolm T.H., Johnson C.E., Danhauer J.L., Portz L.J.P., Abrams H.B., Lesner S., McCarthy P.A. and Newman C.W., A systematic review of health-related quality of life and hearing aids: final report of the American Academy of Audiology task force on the health-related quality of life benefits of amplification in adults, *Journal of the American Academy of Audiology* **18** (2), 2007, 151–183.

Clark W.W., Noise exposure from leisure activities: a review, *The Journal of the Acoustical Society of America* **90** (1), 1991, 175–181.

Cox R.M., Acoustic aspects of hearing aid-ear canal coupling systems, *Monographs in Contemporary Audiology* **1** (3), 1979, 1–44.

Cox R.M. and Alexander G.C., Acoustic versus electronic modifications of hearing aid low-frequency output, *Ear and Hearing* **4** (4), 1983, 190–196.

Cox R.M. and Moore J.N., Composite speech spectrum for hearing aid gain prescriptions, *Journal of Speech and Hearing Research* **31**, 1988, 102–107.

Cox R.M., Mateisch J.S. and Moore J.N., Distribution of short-term RMS levels in conversational speech, *Journal of the Acoustical Society of America* **84**, 1988, 1100–1104.

Dalton D.S., Cruickshanks K.J., Klein B.E., Klein R., Wiley T.L. and Nondahl D.M., The impact of hearing loss on quality of life in older adults, *Gerontologist* **43** (5), 2003, 661–668.

Davies-Venn E., Souza P. and Fabry D., Speech and music quality ratings for linear and nonlinear hearing aid circuitry, *Journal of the American Academy of Audiology* **18** (8), 2007, 688–699.

Dillon H., Hearing Aids, second ed., 2012, Boomerang Press; Turramurra, Australia.

Dillon H., Keidser G., O'Brien A. and Silberstein H., Sound quality comparisons of advanced hearing aids, *Hearing Journal* **56** (4), 2003, 30–40.

Dunn H.K. and White S.D., Statistical measurements on conversational speech, *The Journal of the Acoustical Society of America* **11**, 1940, 278–288.

Edwards B., The future of hearing aid technology, *Trends in Amplification* **11** (1), 2007, 31–45.

Flugrath J.M., Modern-day rock-and-roll music and damage risk criteria, *The Journal of the Acoustical Society of America* **45** (3), 1969, 704–711.

Gabrielsson A. and Sjögren H., Perceived sound quality of hearing aids, *Scandinavian Audiology* **8**, 1979, 159–169.

Gabrielsson A., Rosenberg U. and Sjögren H., Judgements and dimension analyses of perceived sound quality of sound-reproducing systems, *The Journal of the Acoustical Society of America* **55** (4), 1974, 854–861.

Gabrielsson A., Lindström B. and Ove T., Loudspeaker frequency response and perceived sound quality, *The Journal of the Acoustical Society of America* **90** (2), 1991, 707–719.

Gitles T.C. and Niquette P.T., FIG6 in ten, *The Hearing Review* **2** (10), 1995, 28–30.

Groth J. and Sondergaard M.B., Disturbance caused by varying propagation delay in non-occluding hearing aid fittings, *International Journal of Audiology* **43** (10), 2004, 594–599.

Halprin C., The tuning curve in clinical audiology, *American Journal of Audiology* **11**, 2002, 54–56.

Hamacher V., Chalupper J., Eggers J., Fischer E., Kornagel U., Puder H. and Rass U., Signal processing in high-end hearing aids: state of the art, challenges, and future trends, *Journal on Applied Signal Processing* **18**, 2005, 2915–2929.

Hart C.W., Geltman C.L., Schupbach J. and Santucci M., The musician and occupational sound hazards, *Medical Problems of Performing Artists* **2** (3), 1987, 22–25.

Hockley N.S., Bahlmann F. and Chasin M., Programming hearing instruments to make live music more enjoyable, *The Hearing Journal* **63** (9), 2010, 30–38.

Hockley N.S., Bahlmann F. and Fulton B., Analog to digital conversion to accommodate the dynamics of live music in hearing Instruments, *Trends in Amplification* **16** (3), 2012, 146–158.

Hudgins C.V., Marquis R.J., Nicholds R.H., Peterson G.E. and Ross D.A., The comparative performance of an experimental hearing aid and two commercial instruments, *The Journal of the Acoustical Society of America* **20** (3), 1948, 241–258.

Johnson K., Acoustic and Auditory Phonetics, second ed., 2003, Blackwell; Oxford, UK.

Kates J.M., Principles of digital dynamic-range compression, *Trends in Amplification* **9** (2), 2005, 45–76.

Kates J.M., Digital Hearing Aids, 2008, Plural Publishing; San Diego CA.

Keidser G., Dillon H., Flax M., Ching T. and Brewer S., The NAL-NL2 prescription procedure, *Audiology Research* **1** (e24), 2011.

Kiessling J., Brenner B., Thunberg Jerspersen C. and Drylund Jensen Ole, Occlusion effect of earmolds with different venting systems, *Journal of the American Academy of Audiology* **16**, 2005, 237–249.

Killion M.C., Technological report: an "acoustically invisible" hearing aid, *Hearing Instruments* **39** (10), 1988 www.etymotic.com.

Killion M.C., A high fidelity hearing aid, *Hearing Instruments* **41** (8), 1990 www.etymotic.com.

Killion M.C., The K-AMP hearing aid: an attempt to present high fidelity for the hearing-impaired, In: Beilin Joel and Jensen Gert R., (Eds.), *Recent Developments in Hearing Instrument Technology: 15th Danavox Symposium,* 1993, Stougaard Jensen; Copenhagen, 167–229.

Killion M.C., What special hearing aid properties do performing musicians require?, *The Hearing Review* **16** (2), 2009, 20–31.

Kochkin S., MarkeTrak VIII: patients report improved quality of life with hearing aid usage, *Hearing Journal* **64** (6), 2011, 25–32.

Kochkin S., MarkeTrak VIII: the key influencing factors in hearing aid purchase intent, *Hearing Review* **19** (03), 2012, 12–25.

Kuk F.K., Perceptual consequence of vents in hearing aids, *The British Journal of Audiology* **25**, 1991, 163–169.

Kuk F., Considerations in modern multichannel nonlinear hearing aids, (Chapter 5)In: Valente M., (Ed), *Hearing Aids: Standards, Options and Limitations,* second ed., 2002, Thieme; New York NY, 178–213.

Laplante-Lévesque A., Hickson L. and Worrall L., Predictors of rehabilitation intervention decisions in adults with acquired hearing impairment, *Journal of Speech, Language, and Hearing Research* **54**, 2011, 1385–1399.

Laplante-Lévesque A., Hickson L. and Worrall L., What makes adults with hearing impairment take up hearing aids or communication programs and achieve successful outcomes?, *Ear and Hearing* **33**, 2012, 79–93.

Levitt H., A historical perspective on digital hearing aids: how digital technology has changed modern hearing aids, *Trends in Amplification* **11** (1), 2007, 7–24.

Lewis J.D., Goodman S.S. and Bentler R.A., Measurement of hearing aid internal noise, *The Journal of the Acoustical Society of America* **127** (4), 2010, 2521–2528.

Lybarger S.F., Earmold venting as an acoustic control factor, (Chapter 10)In: Studebaker Gerald A. and Hochberg Irving, (Eds.), *The Acoustical Factors Affecting Hearing Aid Performance,* 1980, University Park Press; Baltimore MD, 197–217.

MacDonald E.N., Behar A., Wong W. and Kunov H., Noise exposure of opera musicians, *Canadian Acoustics* **36** (4), 2008, 11–16.

Moore B.C.J., Perceptual consequences of cochlear hearing loss and their implications for the design of hearing aids, *Ear and Hearing* **17** (2), 1996, 133–161.

Moore B.C.J., Cochlear Hearing Loss: Physiological, Psychological and Technical Issues, second ed., 2007, John Wiley and Sons Ltd; Chichester UK.

Moore B.C.J., The choice of compression speed in hearing aids: theoretical and practical considerations and the role of individual differences, *Trends in Amplification* **12** (2), 2008, 103–112.

Moore B.C.J., An Introduction to the Psychology of Hearing, sixth ed., 2012a, Emerald Group; Bingley UK.

Moore B.C.J., Effects of bandwidth, compression speed, and gain at high frequencies on preferences for amplified music, *Trends in Amplification* **16** (3), 2012b, 159–172.

Moore B.C.J., Glasberg B.R. and Stone M.A., Development of a new method for deriving initial fittings for hearing aids with multi-channel compression: CAMEQ2-HF, *International Journal of Audiology* **49** (3), 2010, 216–227.

Moore B.C.J., Fullgrabe C. and Stone M.A., Determination of preferred parameters for multiband compression using individually fitted simulated hearing aids and paired comparisons, *Ear and Hearing* **32** (5), 2011, 556–568.

NCOA: National Council on the Aging, The Consequences of Untreated Hearing Loss in Older Persons, 1999, Washington D.C..

Olsen W.A., Average speech levels and spectra in various speaking/listening conditions: a summary of the Pearson, Bennett, and Fidell (1977) report, *American Journal of Audiology* **7**, 1998, 1–5.

Parsa V., Scollie S., Glista D. and Seelisch A., Non-linear frequency compression: effects on sound quality ratings of speech and music, *Trends in Amplification* **17** (1), 2013, 54–68.

Phillips S.L. and Mace S., Sound level measurements in music practice rooms, *Music Performance Research* **2**, 2008, 36–47.

Plyler P.N., Hill A.B. and Trine T.D., The effects of expansion on the objective and subjective performance of hearing instrument users, *Journal of the American Academy of Audiology* **16** (2), 2005a, 101–113.

Plyler P.N., Hill A.B. and Trine T.D., The effects of expansion time constants on the objective performance of hearing instrument users, *Journal of the American Academy of Audiology* **16** (8), 2005b, 614–621.

Plyler P.N., Trine T.D. and Hill A.B., The subjective evaluation of the expansion time constant in single-channel wide dynamic range compression hearing instruments, *International Journal of Audiology* **45** (6), 2006, 331–336.

Poissant S.F., Freyman R.L., MacDonald A.J. and Nunes H.A., Characteristics of noise exposure during solitary trumpet playing: immediate impact on distortion-product otoacoustic emissions and long-term implications for hearing, *Ear and Hearing* **33** (4), 2012, 543–553.

Revit L.J., What's so special about music?, *Hearing Review* **16** (2), 2009, 12–19.

Ricketts T.A., Dittberner A.B. and Johnson E.E., High frequency amplification and sound quality in listeners with normal through moderate hearing loss, *Journal of Speech-language-hearing Research* **51**, 2008, 160–172.

Ricketts T.A., Johnson E. and Federman J., Individual differences within and across feedback suppression hearing aids, *Journal of the American Academy of Audiology* **19** (10), 2008, 748–757.

Royster J.D., Royster L.H. and Killion M.C., Sound exposures and hearing thresholds of symphony orchestra musicians, *The Journal of the Acoustical Society of America* **89** (6), 1991, 2793–2803.

Rutledge K.L., A Music Listening Questionnaire for Hearing Aid Users, Unpublished Master Thesis2009, The University of Canterbury; Christchurch, New Zealand.

Schaub A., Digital Hearing Aids, 2008, Thieme; New York.

Schmidt J.H., Sound exposure of symphony orchestra musicians, *Annals of Occupational Hygiene* **55** (8), 2011, 893–905, DOI: 10.1093/annhyg/mer055, Downloaded Sept 21, 2012.

Schmidt M., Musicians and hearing aid design – is your hearing instrument being overworked?, *Trends in Amplification* **16** (3), 2012, 140–145.

Scollie S., Seewald R., Cornelisse L., Moodie S., Bagatto M., Launagaray D., Beaulac S. and Pumford M., The desired sensation level multistage input/output algorithm, *Trends in Amplification* **9** (4), 2005, 159–197.

Spriet A., Moonen M. and Wouters J., Evaluation of feedback reduction techniques in hearing aids based in physical performance measures, *Journal of the Acoustical Society of America* **128** (3), 2010, 1245–1261.

Steinberg J.C. and Gardner M.B., The dependence of hearing impairment on sound intensity, *The Journal of the Acoustical Society of America* **9**, 1937, 11–23.

Stone M.A. and Moore B.C.J., Tolerable hearing aid delays: I. Estimation of limits imposed by the auditory pathway alone using simulated hearing losses, *Ear and Hearing* **20** (3), 1999, 182–192.

Stone M.A. and Moore B.C.J., Tolerable hearing aid delays: II. Estimation of limits imposed during speech production, *Ear and Hearing* **23** (4), 2002, 325–338.

Stone M.A. and Moore B.C.J., Tolerable hearing aid delays: III. Effects on speech production and perception of across-frequency variation in delay, *Ear and Hearing* **24** (2), 2003, 175–183.

Stone M.A. and Moore B.C.J., Tolerable hearing aid delays: IV. Effects on subjective disturbance during speech production by hearing-impaired subjects, *Ear and Hearing* **26** (2), 2005, 225–235.

Stone M.A., Moore B.C.J., Meisenbacker K. and Derleth R.P., Tolerable hearing aid delays. V. Estimation of limits for open canal fittings, *Ear and Hearing* **29** (4), 2008, 601–617.

Strawbridge W.J., Wallhagen M.I., Shema S.J. and Kaplan G.A., Negative consequences of hearing impairment in old age: a longitudinal analysis, *Gerontologist* **40** (3), 2000, 320–326.

Uys M. and van DijK C., Development of a music perception test for adult hearing-aid users, *South African Journal of Communication Disorders* **58**, 2011, 19–47.

Uys M., Pottas L., Vinck B. and van DijK C., The influence of non-linear frequency compression on the perception of music by adults with a moderate to severe hearing loss: subjective impressions, *South African Journal of Communication Disorders* **59**, 2012, 53–67.

Venema T.H., Compression for Clinicians, second ed., 2006, Delmar Cengage Learning; Clifton Park NY.

Weinstein B.E. and Ventry I.M., Hearing impairment and social isolation in the elderly, *Journal of Speech and Hearing Research* **25** (4), 1982, 593–599.

Zakis J.A., Fulton B. and Steele B., Preferred delay and phase-frequency response of open-canal hearing aids with music at low insertion gain, *International Journal of Audiology* **51**, 2012, 906–991.

Queries and Answers

Query: Please check the address for the corresponding author that has been added here, and correct if necessary.

Answer: The address is correct.

Query: Please check the hierarchy of the section headings.

Answer: The hierarchy of the section headings is fine

Query: Kindly check the table caption 2 and amend if necessary.

Answer: Replace the table caption for tables 2 and 3 with the following

Table 2 The measured crest factors for some segments of speech and music. For each segment, Adobe Audition 1.0 (San Jose, California) was used to calculate the difference between the RMS value of the signal and the instantaneous peak.

Table 3 For the Music #1 segment in Table 2, the difference between the RMS of the signal and its instantaneous peak is given. For shorter analysis windows the instantaneous peak is higher than for longer windows of analysis with a resulting higher crest factor.

Query: Kindly note that reference "Chasin, 2011" is unlisted.

Answer: I am unsure about how a blog should be referenced. It should be included in the reference list

Query: Uncited references: This section comprises references that occur in the reference list but not in the body of the text. Please cite each reference in the text or, alternatively, delete it. Any reference not dealt with will be retained in this section.

Answer: Please cite this reference after Moore, 2007 in line 500

Query: Please confirm that given names and surnames have been identified correctly.

Answer: These names are correct.

The email address should not be capitalized.

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