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An Acoustical Assessment of the Music Memory in Commercially Available Hearing Aids

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Abstract

Hearing aids amplify speech-input signals using nonlinear amplification (i.e., wide dynamic range compression). When WDRC is used to process a music-input signal, listener's report a negative aided listening experience. To circumvent this negative experience, hearing aids allow for music-input stimuli to be processed using a modified frequency-gain response, known as a memory or program. The music memory, in general, processes the input signal using a linear-like frequency-gain response, elevated output, or both. Increasing gain and output, we conjecture, has the potential to place the wearer at-risk (i.e., ≥85 Leq dBA) for hearing-aid-induced hearing loss (HAIHL). We assessed the potential of this risk in two experiments. In Experiment I, 2-cc coupler gain was determined in three commercially available receiver-in-the ear/receiver-in-the-canal (RITE/RIC) hearing aids. Coupler gain responses were determined for a composite signal presented at 65 and 100 dB SPL for the WDRC memory and music memory, and for different degrees of occlusion. Results from this experiment were reported qualitatively. In Experiment II, the same three devices were fit on an acoustical manikin. Recordings of 10 musical passages were obtained for the same two memories, adjusted for the degree of occlusion at three presentation levels (i.e., 85-, 94-, and 103-dB SPL). Analyses of the recordings revealed that two devices programed in the music mode exceeded the at-risk threshold at presentation levels of 94- and 103-dB SPL. In addition, the same two devices programmed in WDRC exceeded the at-risk threshold at a presentation level of 103-dB SPL. Implications and future directions are discussed.

Keywords Hearing aid; Music memory; Hearing loss; Accountability

Introduction

Hearing aids are engineered with the goal of responding optimally to the characteristics of speech as an input signal, not music. There are important consequences emanating from this engineering requirement, grounded primarily on the acoustic differences between these input signals. The dynamic range of speech spans a range of 30-35 dB, stemming from amplitude differences between the softest (i.e., /ə/) and the loudest (i.e., /ɔ/) sounds of speech [1]. Music, conversely, spans a 100 dB, ranging between 20 and 120 dB SPL [2]. In addition, the long-term average spectrum (LTAS) of continuous discourse is 65 dB SPL, having a maximum short-term spectra (i.e., crest factor) of +12 dB [3]. Thus, speech, at maximum vocal effort, rarely exceeds 85 dB SPL. The same is not true for music. The LTAS for music often exceeds 85 dB SPL, with a crest factor ranging between 18 and 20 dB [4]. Lastly, speech perception is dependent on the listener's ability to comprehend high-frequency cues, typically 1000 Hz and greater [3]. For music perception, the most salient cues are in the low frequencies, many of which extend below 100 Hz [5].

Survey data reports that three-quarters of impaired listeners use their hearing aid to listen to music [6,7]. To accommodate different listening situations, hearing aids are engineered with multiple memories, or programs. Memories allow listeners the opportunity to change the frequency-gain response, based on the listening environment and perceptual needs of the wearer (e.g., listening in

noise, listening in a reverberant room). Listeners select a memory by either manually toggling between memories or the device automatically (i.e., adaptively) changes between memories using a proprietary sound classification algorithm.

In the memory designated for a speech-input signal, a multichannel wide-dynamic range compression (WDRC) circuit processes the input signal using fast, automatic gain adjustments [8]. The outcome of this scheme places more low-input speech cues within the listener's audible range through increased gain, while providing less gain to high-input sounds to ensure listening comfort. The amplified output results in a flattened temporal (i.e., time-amplitude) envelope. The outcome of these adjusted amplitudes is a smeared and distorted signal heard by the listener [9].

When music is the input signal, the speech-centric processing (i.e., WDRC) in a hearing aid negatively affects the aided listening experience [10-12]. Thus, a second (i.e., music) memory is stored within the hearing aid, whereby a music-input signal is processed using a linear or slow-acting compression amplification scheme. Both schemes preserve the temporal envelope better than the fast-acting compression scheme [13]. Preserving the temporal envelope for music also requires extending the device's output level to between 105 dB SPL and 115 dB SPL [4]. In theory, settings in the music memory, which provides a robust amplitude envelope, could result in hearing-aid-induced hearing loss (HAIHL). The purpose of this undertaking is to assess the potential of this risk.

The risk of HAIHL, one could argue, is minimized through the programming capabilities available in the manufacturer's software. This statement posits that the clinician programming the device has knowledge of the electroacoustic settings among the different memories available. The fact is, manufacturers are not required to provide electroacoustic specifications of their program settings [14]. As an example, when the music memory is stored on a given device, the clinician is unaware of the electroacoustic increase to the hearing aid's output level, change from WDRC to linear amplification scheme, or both.

In this paper, we undertook two experiments to assess the acoustical response of speech- and music-centric schemes in three commercially available receiver in-the-canal/receiver in-the-ear (RIC/RITE) devices. We included this style of hearing aid because it represented the largest proportion of devices dispensed in the United States [15]. In Experiment I, we determined the 2-cc coupler gain response of each device's WDRC and music amplification schemes. Coupler gain was obtained using a composite signal presented at levels of 65 and 100 dB SPL. Experiment II took place in a music studio housed within the College of Music at the University of North Texas (UNT). Here, we obtained hearing-aid recordings of musical stimuli for the same three devices fit on an acoustical manikin. Recordings were obtained at three levels (i.e., 85-, 94-, and 103-dB SPL) for both amplification schemes and for different degrees of occlusion. We then analyzed the recordings to assess those levels the WDRC and music amplification schemes placed aided listeners at risk for HAIHL. Results from Experiment I are described qualitatively, while results from Experiment II are analyzed quantitatively. Recommendations from the findings are also discussed.

Experiment I

The purpose of Experiment I was to determine the 2-cc coupler gain response of each device's WDRC and music memories. Each memory's frequency-gain responses were also adjusted for occlusion, a software-based adjustment that increases low-frequency gain when occluded and decreases low-frequency gain when unoccluded.

Methods

Hearing aids

Three monaural hearing aids were selected from devices consigned to the University of North Texas Speech and Hearing Center: GN Resound Live 9 TS Open, Oticon Epoq XW, and Starkey Wi Series i110.

The GN Resound Live 9 TS Open RITE device is a four memory, high-end digital device engineered to process incoming sounds using a 17-band warp. We configured this device with a low-power receiver option and an omnidirectional microphone pattern. Memory 1 was used for both the WDRC and music conditions. Specifically, we programmed both environmental settings manually into memory 1, as opposed to an environmental setting in memory 1 (e.g., WDRC) and the other environmental setting in memory 2 (i.e., music). This manual programming methodology prevented the hearing aid from switching automatically between memories during the experiments.

Using the Aventa 2.9 fitting software, target gain was determined using the NAL-NL1 prescriptive formula [16]. The target gain was derived from the hypothetical audiogram shown in Figure 1. We

accounted for electroacoustic differences in occlusion during the recording process through adjustments available in the fitting software, reducing low-frequency gain in the unoccluded condition and increasing low-frequency gain in the occluded condition. All remaining features (e.g., noise reduction, feedback control) were disabled during the recording process.
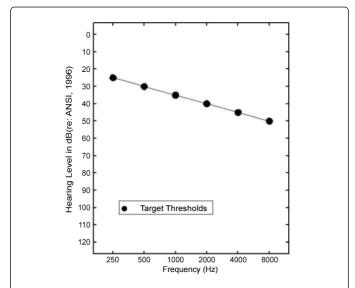

Figure 1: Target thresholds used to program the three commercially available devices used in this study.

The Oticon Epoq XW RITE is a four memory, high-end digital device that processes incoming sounds in 10 fitting bands. The device's microphone sensitivity was programmed to omnidirectional, and the electroacoustic characteristics were based on the energetic identity. Memory 1 was programmed to the WDRC setting and memory 2 was programmed to the music setting. The remaining two memories were disabled. Memories were not linked together during the recording process. This prevented the hearing aid from switching automatically between memories during the recording process. Similar to the other devices, we accounted for the degree of occlusion using the manufacturer's software.

Target gain was determined using the Voice Aligned Compression (VAC) prescriptive approach [17], a compression algorithm based on the work of Buus and Florentine [18]. VAC is based on the following objectives: (1) reduce gain more rapidly for soft-input levels (i.e., below 30 dB SPL) to reduce low-intensity background noise, (2) apply curvilinear compression between 30 and 45 dB SPL to heighten the listener's awareness of the environment, (3) reduce compression at moderate-to-high-level intensities to improve speech perception, and (4) allow no amplification (0-dB gain) at high-input levels. The target gain derived from the VAC algorithm was predicated on the same hypothetical audiogram (Figure 1), which was entered directly into the Genie 2011.1 fitting software. All additional technical features, including noise reduction and feedback cancellation, were disabled during this study.

The third device used in this study was the Starkey Wi Series i110 RIC 312. This device is a four memory, high-end digital device that processes incoming sounds in 16 channels. The device was configured with a 40-dB receiver option and programmed with an omnidirectional microphone pattern. We entered the thresholds from

Figure 1 into the manufacturer's NOAH-compatible Inspire 6.1.200 fitting software. The software then applied the thresholds, yielding NAL-NL1-derived target gain [16]. The WDRC setting was programmed into memory 1 and the music-classical setting was programmed into memory 2. The remaining two memories were disabled. To prevent the hearing aid from switching automatically between memories during the recording process, memories 1 and 2 were not linked together. Similar to the other two devices, electroacoustic characteristics also accounted for degree of occlusion (i.e., unoccluded, occluded). All other technical features were disabled during this study.

Procedures

Measurements of 2-cc coupler gain were made in a Fonix 7000 hearing aid analyzer. We introduced a composite test signal as the input signal, presented at 65- and 100-dB SPL. During testing, the hearing aid's position remained constant within the test box. Fun-Tack® was used to adhere the receiver of each device to an HA-1 coupler, with the receiver positioned flush and centered with respect to the aperture of the coupler. All programming changes to a given device were initiated through each manufacturer's NOAH-compatible fitting software and transmitted via NOAH-link.

Results

Figures 2-4 and Figures 5-7 display the 2-cc coupler gain response for each of the three hearing aids. At an input level of 65 dB SPL, both the GNResound and Starkey devices showed marked increases in gain between the WDRC memory and music memory. For the GNResound device, Figure 2 shows an increase in unoccluded gain ranging between 5 and 10 dB between the WDRC memory, denoted by the solid black line, and music memory, indicated by the dashed black line. When the hearing aid was programmed to an occluded setting, results yielded less variable gain between memories at frequencies less than 1000 Hz. In addition, the occluded setting revealed a 3 to 5 dB increase in gain for the music memory between 1000 and 4000 Hz compared to the unoccluded condition.

As seen in Figure 4, the Starkey device demonstrated the largest coupler gain difference between programs, with the music memory (dashed black line) providing roughly 15 dB more unoccluded amplification than the WDRC memory (solid black line). This finding was seen mainly between the frequencies of 500 and 3000 Hz. In the occluded condition, the music memory continued to provide more amplification between 3 and 10 dB between 100 and 3000 Hz than the WDRC memory. The Oticon device, conversely, provided essentially the same coupler gain, regardless of memory and type of occlusion (Figure 3).

At an input level of 100 dB SPL, both the WDRC and music memories in the GNResound device provided essentially no or negative unoccluded gain (Figure 5). For the occluded conditions, gain increased by a maximum of 5 dB. The Oticon device, as shown in Figure 6, provided substantial negative gain for both memories and under both occlusion conditions. Conversely, the Starkey device provided the greatest difference in gain at this input level. Specifically, coupler results yielded an increased gain—between 10 and 15 dB—when the device was programmed to the music memory (Figure 7) compared to the WDRC memory. Further, the Starkey device provided between 5 and 10 dB of positive amplification between 800 and 2000 Hz in the unoccluded condition. Similar positive

amplification was seen again—this time, between 2000 and 6000 Hz—in the occluded condition. Clearly, the findings from this electroacoustic testing indicate that some devices provide an increase amount of amplification in the music memory compared to the WDRC memory.

The findings from our electroacoustic testing are conservative, mainly because we used a HA-1 coupler instead of the not-yet-standardized Open-Fit coupler. Specifically, the HA-1 coupler is standardized to a volume of 2-cc. This volume represents the distance between the receiver of a custom device and the tympanic membrane. For open-fit devices, on the other hand, the volume between the receiver and the tympanic membrane is smaller. Had we used the Open-Fit coupler with its smaller volume, our coupler gain results would have yielded larger amounts of gain. This unaccounted coupler gain further supports our hypothesis that the music memory in commercially available hearing aids poses a risk of accelerated hearing loss to the wearer.
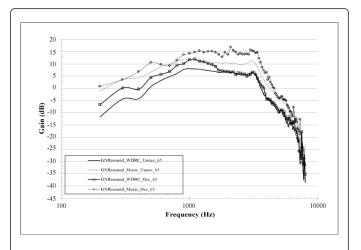

Figure 2: 2-cc coupler gain measures obtained at a 65 dB SPL for the GN Resound Live 9 TS receiver in-the-ear (RITE) device programmed for the memory settings used in this study. (WDRC=Wide Dynamic Range Compression; Unocc=Unoccluded setting; Occ=Occluded setting).
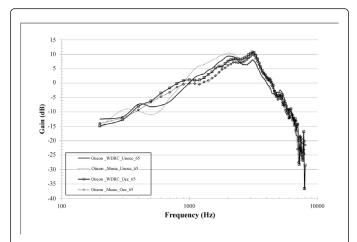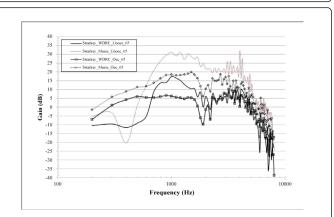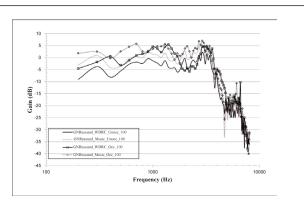
Figure 3: 2-cc coupler gain measures obtained at a 65 dB SPL for the Oticon Epoq XW RITE device programmed for the memory settings used in this study. (WDRC=Wide Dynamic Range Compression; Unocc=Unoccluded setting; Occ=Occluded setting).

Figure 5: 2-cc coupler gain measures obtained at a 100 dB SPL for the GN Resound Live 9 TS receiver in-the-ear (RITE) device programmed for the memory settings used in this study. (WDRC=Wide Dynamic Range Compression; Unocc=Unoccluded setting; Occ=Occluded setting).
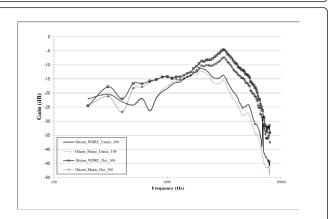

Figure 4: 2-cc coupler gain measures obtained at a 65 dB SPL for the Starkey Wi Series i110 receiver in-the-canal (RIC) 312 device programmed for the memory settings used in this study. (WDRC=Wide Dynamic Range Compression; Unocc=Unoccluded setting; Occ=Occluded setting).
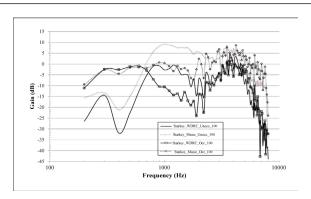

Figure 6: 2-cc coupler gain measures obtained at a 100 dB SPL for the Oticon Epoq XW receiver in-the-ear (RITE) device programmed for the memory settings used in this study. (WDRC=Wide Dynamic Range Compression; Unocc=Unoccluded setting; Occ=Occluded setting).

Figure 7: 2-cc coupler gain measures obtained at a 100 dB SPL for the Starkey Wi Series i110 receiver in-the-canal (RIC) 312 device programmed for the memory settings used in this study. (WDRC=Wide Dynamic Range Compression; Unocc=Unoccluded setting; Occ=Occluded setting).

Experiment II

The purpose of Experiment II was to compare hearing-aid recordings of musical stimuli processed in the WDRC memory and music memory obtained on an acoustical manikin.

Methods

Hearing aids

We used the same devices and settings from Experiment I in Experiment II. To ensure reliability, we compared 2-cc coupler gain measures during Experiment II to those reported in Experiment I for a 65-dB input level. Results revealed reliable coupler gain measurement.

Musical stimuli

The experimental stimuli consisted of 10 musical segments selected randomly from a library of digital recordings of live student performances at the UNT College of Music. Each segment spanned between 20 and 25 seconds in duration and represented a cross-section of musical genre, ensemble size, and performance venue (Figure 8). In addition, a five-second silent interval was inserted between each musical segment on the wav file. This period of silence allowed sufficient time for the hearing aid compression parameters to return to their linear state.

Procedures

Recordings of hearing-aid processed stimuli took place in a music studio housed within the UNT College of Music. The studio dimensions were 5.9 m (l) x 4.2 m (w) x 3.0 m (h). The studio had a carpeted floor, concrete-block walls with acoustic damping material covering one third of each wall area, and a suspended ceiling covered by standard 2 feet x 2 feet acoustical panels. The room includeda table and chairs, and digital and analog audio equipment. The mean reverberation time (RT) of this studio was 1.246 seconds averaged across the frequencies of 500, 1000, 2000, and 4000 Hz.
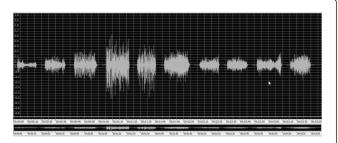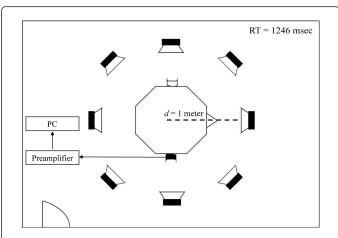

Figure 8: Waveform depicting the 10 short musical segments recorded through each of the hearing aids used in this study. See text for additional details.

We recorded hearing-aid processed musical stimuli without and with amplification. In the aided condition, each device was placed in the right ear of a Knowles Electronics Manikin for Acoustic Research (KEMAR) [19]. The musical stimuli were presented to KEMAR using the loudspeaker layout shown in Figure 9 for all test conditions. We positioned loudspeakers (Tannoy 800A) at an equal distance of 1 meter, indicated by *d*, and at a height of 1.5 meters relative to the center-head position of KEMAR (Figure 9).

The music stimuli were presented simultaneously (i.e., correlated) from eight loudspeakers positioned in 45-degree increments relative to 0-degrees azimuth, and at three respective levels: 85, 94, and 103 dB SPL. The presentation levels were determined by presenting pink noise from the loudspeaker array and measuring the sound pressure level (+ 1 dB) at 1000 Hz using a sound level meter (Larson-Davis 800B) positioned in the center and directly above KEMAR. During the recording process, the amplified output from KEMAR (Etymotic Research microphone preamplifier model ER-11) was routed to a personal computer (PC) and saved as a monaural wav file (mono, 16-bit resolution, 44100 Hz sampling rate). During the aided recordings, the ER-11 preamplifier was enabled to exclude external ear effects [20]. A single recording of the music stimuli was obtained for each of the three hearing aids (GNResound, Oticon, Starkey). Recordings were acquired across amplification scheme (WDRC, music) and degree of occlusion (unoccluded, occluded), at three presentation level (85, 94, 103 dB SPL).

Figure 9: Loudspeaker arrangement used to present the musical passage for recording via a KEMAR manikin. See text for explanation of parameter *d*.

Acoustical analyses

The time average level, or equivalent continuous noise level (Leq), of each aided condition was determined using a custom-written algorithm in Matlab. Specifically, the algorithm calculated the A-weighted sound pressure level (dBA) in octave bands—using a Kaiser-window filter for anti-aliasing—for a given input wav file. The equation used to calculate dBA was:

$$L = 10 \ log \sum_{i=1}^{n} 10^{Li+ \ Ki/10}$$

where L represented the combined level in dB SPL, n the number of bands being combined, i the ith band, Li the octave band level, and Ki the A-weighted correction to simulate human auditory sensitivity for the octave frequencies ranging between 31.5 and 8000 Hz, per ANSI S1.4-1983 (R2006) [21].

Acoustical analyses were performed for the entire music passage (i.e., all 10 musical segments), the softest musical segment (i.e., first waveform in Figure 8), and the loudest musical segment (i.e., fourth waveform in Figure 8). We included the softest and loudest musical passages because of the level-dependent nature of most devices to input signals at different levels. To analyze the softest and loudest segments of each wav file, we digitally excised the softest and loudest passages from their respective wav files. Excising was performed by placing the marker 100 milliseconds prior to the onset and 100 milliseconds following the offset of each musical segment using digital editing software (Goldwave, version 5.58).

Statistical analysis

Each device provided 36 observations (2 [memories] x 2 [domes] x 3 [presentation levels] x 3 [passage segments]) for statistical analysis. However, because we recorded only a single wav file at each presentation level for each independent variable, we are unable to statistical analyze differences using a random-effects model. We were, however, able to calculate the mean and standard deviation for a given treatment variable (e.g., hearing-aid processed stimuli of all passages processed with WDRC) and compare it to another treatment variable (e.g., hearing-aid processed stimuli of all passages processed with music). Statistical comparisons across means were performed using a paired Student's t-tests at a significance level of 0.05. We employed a Bonferroni-correction factor to reduce thepotential of increasing the type I error rate when multiple t-test comparisons were performed.

Results

Tables 1-3 depict the results from our acoustic analyses for each device. A summary of the findings follows.

Stimulus	85 dB level		94 dB level		103 dB level		Mean (SD)		
	WDRC	Music	WDRC	Music	WDRC	Music	WDRC	Music	Both
All Passages									
Unoccluded	77.2	79.9	81.2	84.1	85.3	89.3	81.2 (4.1)	84.4 (4.7)	82.8 (4.3)
Occluded	74.3	77.4	76.7	79.2	75.6	77.1	75.5 (1.2)	77.9 (1.1)	76.7 (1.7)
Softest Passage									
Unoccluded	79.3	80.6	83.9	86.1	88.3	92.7	83.3 (4.5)	86.5 (6.1)	85.2 (5.0)
Occluded	76.8	78.9	79.6	85.8	79.9	82.7	78.7 (1.7)	82.5 (3.5)	80.6 (3.2)
Loudest Passage									
Unoccluded	80.4	81.6	85.4	87.1	89.6	91.3	85.1 (4.6)	86 (4.9)	85.6 (4.3)
Occluded	81.7	81.0	83.6	85.1	87.6	85.5	84.3 (3.2)	84.5 (3.2)	84.4 (2.8)
Mean	78.3	79.9	81.7	84.6	84.4	86.4			
(SD)	(2.7)	(1.5)	(3.2)	(2.8)	(5.5)	(5.8)			

Table 1: Equivalent continuous levels, or dBA Leq, for musical stimuli processed through the GNResound hearing aid fit on KEMAR.

Stimulus	85 dB level		94 dB level		103 dB level		Mean (SD)		
	WDRC	Music	WDRC	Music	WDRC	Music	WDRC	Music	Both
All Passages									
Unoccluded	72.1	70.5	75.6	75.0	78.7	79.7	75.5 (3.3)	75.1 (4.6)	75.3 (3.6)
Occluded	68.0	66.0	72.9	70.9	73.7	74.0	71.5 (3.1)	70.3 (4.0)	70.9 (3.3)
Softest Passage									
Unoccluded	73.2	71.6	77.4	75.9	81.3	79.9	77.3 (4.1)	75.8 (4.2)	76.6 (3.8)
Occluded	70.4	69.6	76.0	74.2	79.4	77.2	75.3 (4.6)	73.7 (3.8)	74.5 (3.9)
Loudest Passage									
Unoccluded	75.6	74.0	80.9	79.0	86.7	85.6	81.1 (5.6)	79.5 (5.8)	80.3 (5.2)
Occluded	74.9	73.1	80.1	77.5	84.8	83.3	79.9 (5.0)	78.0 (5.1)	79.0 (4.6)
Mean	72.4	70.8	77.2	75.4	80.8	80.0			
(SD)	(2.9)	(2.9)	(3.0)	(2.8)	(4.6)	(4.2)			

Table 2: Equivalent continuous levels, or dBA Leq, for musical stimuli processed through the Oticon hearing aid fit on KEMAR.

Stimulus	85 dB level		94 dB level		103 dB level		Mean (SD)		
	WDRC	Music	WDRC	Music	WDRC	Music	WDRC	Music	Both
All Passages									
Unoccluded	69.0	75.5	73.3	78.0	80.2	82.9	74.2 (5.7)	78.8 (3.8)	76.5 (5.0)
Occluded	68.7	76.7	72.7	79.7	79.7	86.1	73.7 (5.6)	80.8 (4.8)	77.3 (6.1)
Softest Passage									
Unoccluded	74.8	79.1	80.9	86.8	85.4	89.1	80.4 (5.3)	85.0 (5.2)	82.7 (5.3)
Occluded	74.7	79.7	78.7	85.6	86.6	90.3	80 (6.1)	85.2 (5.3)	82.6 (5.8)
Loudest Passage									
Unoccluded	76.3	79.5	82.7	86.2	89.8	91.5	82.9 (6.8)	85.7 (6.0)	84.3 (5.9)
Occluded	74.3	77.0	83.9	86.3	88.1	91.8	82.1 (7.1)	85.0 (7.5)	83.6 (6.7)
Mean	73.0	77.9	78.7	83.8	85.0	88.6			
(SD)	(3.3)	(1.8)	(4.8)	(3.9)	(4.2)	(3.5)			

Table 3: Equivalent continuous levels, or dBA Leq, for musical stimuli processed through the Starkey hearing aid fit on KEMAR.

Assessing HAIHL

The primary rationale for undertaking this study was to determine whether the music-centric memory in a commercially available hearing aid could place the aided listener at-risk for HAIHL. An at-risk condition was met when the dB Leq(A) level exceeded 85 dB Leq (A). This level is the recommended exposure limit (REL) according to the National Institute for Occupational Safety and Health (NIOSH) [22].

Across all 36 conditions per hearing aid, the GNResound and Starkey devices exceeded the REL in 38.9% (i.e., 14 out of 36) and 36.1% (i.e., 13 out of 36), respectively, of the conditions tested in this study. The Oticon device, on the other hand, exceeded the REL in 5.5% (i.e., 2 out of 36) of the conditions tested in this study. In the following subsections, we examine the independent variables of memory, type of dome, influence of presentation level, and passage segment as a function of each device to determine the factor(s) placing the end user at-risk for potential accelerated hearing loss.

WDRC memory vs. music memory

The columns in Tables 1-3 show the mean dB Leq(A) for the WDRC memory and music memory as a function of presentation level for each device, independent of type of dome and passage segment.

For the GNResound and Starkey devices, the music memory yielded a larger dB Leq(A) than the WDRC memory, regardless of presentation level. Table 1 indicates that the GNResound device exceeded the REL only in the music memory for the input level of 103 dB SPL. At this presentation level, the music memory was found to be provide borderline statistically (Bonferroni-corrected t test; p<.05/3=0.017) greater dB Leq(A) values (t[5]=-4.02; p<.017) compared to the WDRC memory.

For the Starkey device, Table 2 reveals that both the WDRC memory and music memory at input levels of 103 dB SPL exceeded the REL. From a statistical standpoint, the music memory engineered for this manufacturer provided statistically (p<.017) greater dB Leq(A) values than the WDRC memory at all three input levels. For all three input levels, the WDRC memory and music memory designed in the Oticon device did not exceed the REL (Table 3).

Post-hoc analysis

The primary advantage of open-fit devices (i.e., RIC/RITE) is reduction of the occlusion effect [23]. The occlusion effect for a hearing aid wearer is reduced markedly by providing a large vent that allows low-frequency sounds, typically those below 1000 Hz, to escape from the ear canal. Given the modular nature of RIC/RITE fittings—where custom ear molds are replaced with premolded tips, or domes, of varying sizes and with various venting options—each manufacturer's software compensates for low-frequency gain based on the degree of occlusion. The concern with fitting of an unoccluded dome, as it pertains to at-risk environments, is that ambient sounds are able to enter the ear canal and stimulate the cochlea directly. The occluded dome, we hypothesize, could serve to reduce the amount of sound pressure reaching the cochlea.

A statistical analysis was performed to determine whether the degree of occlusion reduced the potential risk for HAIHL. Results revealed a statistical significant main effect ($F_{1,72}$=10.71, p<.01) for the degree of occlusion. Overall, we found that the unoccluded condition allows an additional Leq(A) of 2.26 dB to enter the ear canal compared to the occluded condition. When the mean attenuation provided by the occluded dome is applied to the NIOSH maximum time of exposure formula, a four-hour exposure at 88 dB Leq(A), as an example, is now increased by a coefficient of 0.59, or by 2 hours and 37 minutes (i.e., (4 hours at 88 dB Leq(A)*0.59).

Maximum time of exposure for an 8-hour exposure, using the NIOSH (1998) standard, was calculated as:

$$t = \frac{480}{2^{(L-85)/3}}$$

Where t represents the maximum exposure duration, in minutes, L represents the exposure level, in dBA, 3 represents the exchange rate, in dB, and 85 denotes the recommended exposure level.

This finding indicates that the occluded dome acts as a pseudo hearing protector, allowing for unamplified exposure to this environment (i.e., 88 dB SPL) for up to 6 hours and 37 minutes. The coefficient of 0.59 can be applied to all NIOSH exposure levels.

Results also yielded a significant interaction effect ($F_{2,72}$=3.051, p=.05) between manufacturer and degree of occlusion. Specifically, the unoccluded condition allowed an additional 4.16 and 2.59 dB Leq(A) to enter the ear canal compared to the occluded condition for the GNResound and Oticon devices, respectively. Results revealed

essentially no difference (i.e., 0.02 Leq(A)) between the occluded and unoccluded conditions in the Starkey device. This latter finding suggests that the Starkey occluded domes did not reduce the SPL levels entering the ear canal compared to the unoccluded domes used in this study.

Discussion and Summary

The purpose of this undertaking was to assess the acoustical response of speech- and music-centric amplification schemes obtained in three commercially available RIC/RITE devices. In Experiment I, we obtained the 2-cc coupler gain response of each device's WDRC memory and music memory. Overall, electroacoustic findings revealed that some devices provide an increase amount of amplification compared to WDRC when the music memory was enabled at input levels of 65 and 100 dB SPL. In Experiment II, we investigated the equivalent continuous sound level to musical stimuli between three commercially available devices. All three devices processed the same input stimuli in two memories (i.e., WDRC, music), presented at three presentation levels (i.e., 85-, 94-, and 103-dB SPL), and in with two types of dome configurations (i.e., unoccluded, occluded). Overall, results revealed that two devices exceeded the at-risk threshold (> 85 dB Leq(A)) at presentation levels of 94-and 103-dB SPL for all stimuli processed through the music memory. In addition, the same two devices exceeded the at-risk threshold at the presentation level of 103-dB SPL for all stimuli processed through the WDRC memory. When we assessed the softest and loudest individual passages, the percentage of devices and conditions that exceeded 85 dB Leq(A) increased as presentation level increased. All three devices exceeded the NIOSH REL for the loudest passage in both the WDRC memory and music memory.

While our results are limited to the measurements obtained on devices used in this study, the fact remains that commercially available hearing aids have the potential to place the wearer at-risk for HAIHL. This potential risk increases markedly as the level of the input signal is increased, regardless of whether the device is processing the input signal in the WDRC memory or music memory. Given these findings, there is an ethical obligation (1) to reassess how manufacturers disclose hearing aid electroacoustic parameters to clinicians, (2) to consider disclose of HAIHL risk to patients by manufacturers and clinicians, and (3) to reinforce that clinicians perform verification measures using real-ear equipment—and not aided speech-recognition testing—as part of the hearing aid fitting process. The supporting rationale for each ethical obligation is discussed below in detail.

Disclosure of electroacoustic specifications by manufacturers

ANSI S3.22-2009 is the present-day standard used by manufacturers to define the electroacoustic performance of a given device. The primary purpose of the ANSI S3.22-2009 standard is (1) to allow for universal terms and performance parameters to be defined among manufacturers and (2) to ensure that a given device meets its engineering specifications. Clinicians also use this standard to compare a given hearing aid's electroacoustic performance to the manufacturer's specification data, as a means to ensure adequate function prior to the hearing aid fitting.

The output sound pressure level for 90-dB input signal (OSPL90) is one electroacoustic parameter tested in the ANSI S3.22-2009 hearing aid standard. Specifically, this measurement estimates the maximum output of a given device for a 90-dB SPL input signal when the gain

control set to full-on. In most cases, the hearing aid should reach its maximum output under these conditions.

Most hearing aids are also designed with a peak input-limiting level, which is located just after the microphone. This peak input limiter is designed to prevent signals from exceeding 85 dB SPL from being transduced by the device, based on the LTAS (i.e., conversational speech plus crest factor).

According to Chasin and Russo [4], hearing aid manufacturers are not required to report the peak input-limiting level on ANSI hearing aid specification sheets. We believe it would be helpful to clinicians if manufacturers provided the peak-input value on the ANSI specification sheet for each device, and for each memory available in a given device.

In addition to the recommended revision to the ANSI S3.22 standard on measuring hearing aid output [24], much of the HAIHL risk assessed in this study can be mitigated if manufacturers provided a volume control on their devices. Because the majority of the devices dispensed today are designed with WDRC circuitry, volume controls are typically not available, despite the desirability of this feature by the end user [25-27]. The absence of a volume control on a today's WDRC device is predicated on the input-compression parameters that provide fixed-frequency, level-dependent gain.

Disclosure of HAIHL risk to end user

The fact that hearing aids place the wearer at-risk for hearing loss requires, in our opinion, a disclosure. A precedence on how providing disclosure can be found in related industries. In Birdsongand Waggoner v. Apple, Inc. [28], for instance, a class-action lawsuit was filed against Apple, Inc., the manufacturer of personal music players. The class-action lawsuit purported that this technology endangered hearing loss. In addition, the lawsuit claimed that Apple, Inc. did little to educate consumers about the potential of hearing damage from using the technology. The court case, heard in the United States Court of Appeals for the Ninth Circuit, was dismissed when the judge ruled that hearing loss from iPod use was a direct result of the user's judgment and the level at which the volume was placed. Despite the dismissal, this class-action lawsuit resulted in a disclosure of at-risk hearing loss in the manual that accompanies most personal music players. A similar disclosure should be provided to impaired listeners who adopt hearing aids. Further, this disclosure should be accompanied by NIOSH noise exposure levels and maximum allowable duration.

Most hearing aids dispensed do not provide the end user with a volume control. In such cases, at the least, the hearing aid's output must be limited, regardless of the input stimulus. Such precedence is available for personal music player technology in Europe. Specifically, the European Union introduced legislation that limits the output of personal music player to 80 dBA for an 8-hour working day Scientific Committee on Emerging and Newly Identified Health Risks (SCENIHR) [29]. As an option, data from our study suggest that the algorithm used by Oticon may provide a blueprint for other manufacturers in the industry.

Role of the clinician in reducing risk

Despite the limitations of present-day product disclosure by manufacturers and disclosure requirements, clinicians providing hearing aid technology must play an active role to ensure the welfare of their patients. Specifically, it is imperative that clinicians verify the amount of gain and output provided by the music memory at the tympanic membrane using a real-ear system, not aided sound-field testing. While aided sound-field testing is considered a real-ear measure, this procedure fails to provide verification of gain at moderate- and high-input levels in nonlinear devices [30], such as those tested in this study. In addition to real-ear measurements, clinicians can assess potential risk of HAIHL using a hearing aid test box, as we did in this study.

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Conflict of Interest Statement

There are no financial relationships to disclose.

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