Hearing protection: Surpassing the limits to attenuation imposed by the bone-conduction pathways[a]

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With louder and louder weapon systems being developed and military personnel being exposed to steady noise levels approaching and sometimes exceeding 150 dB, a growing interest in greater amounts of hearing protection is evident. When the need for communications is included in the equation, the situation is even more extreme. New initiatives are underway to design improved hearing protection, including active noise reduction (ANR) earplugs and perhaps even active cancellation of head-borne vibration. With that in mind it may be useful to explore the limits to attenuation, and whether they can be approached with existing technology. Data on the noise reduction achievable with high-attenuation foam earplugs, as a function of insertion depth, will be reported. Previous studies will be reviewed that provide indications of the bone-conduction (BC) limits to attenuation that, in terms of mean values, range from 40 to 60 dB across the frequencies from 125 Hz to 8 kHz. Additionally, new research on the effects of a flight helmet on the BC limits, as well as the potential attenuation from deeply inserted passive foam earplugs, worn with passive earmuffs, or with active-noise reduction (ANR) earmuffs, will be examined. The data demonstrate that gains in attenuation exceeding 10 dB above the head-not-covered limits can be achieved if the head is effectively shielded from acoustical stimulation.
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I. INTRODUCTION

When personnel are exposed to very high levels of noise, such as generated by current and new military weapons systems that approach and even exceed a continuous A-weighted sound pressure level (SPL) of 150 dB, the need for maximum hearing protection is obvious. But, maximum hearing protection may still not be enough, and, by the way, what is the maximum hearing protection that can be provided? Surprisingly, few authors have explored the limits to attenuation since von Békésy (1960) and also von Gierke and Warren (1953) addressed the question in the early 1950s. In this report we review the available data, update them as needed, and explore how the application of active noise reduction (ANR), or the use of an enclosure to shield the head, can affect those limits. The purpose is to provide hearing conservationists and hearing protection developers a reliable benchmark that defines the maximum levels of protection achievable for humans exposed to noise.

As Zwislocki (1957) observed in his landmark paper, direct measurement of such values was long overdue, and in that paper he provided values that have indeed withstood the test of time. In this current work, his data will be compared to a handful of estimates that others, including the first author, have published in the past half-century.

The concept of bone conduction (BC) limits implies that sound is transmitted via bony structures in the head that bypass the normal air-conduction mechanism of transmission through the earcanal. In effect these are flanking sound pathways that circumvent the noise-blocking features of the hearing protection device (HPD) that is covering or occluding the earcanal. The primary BC pathways as described in the literature are (a) vibration of the earcanal walls, (b) energy transmitted due to excitation of ossicular motion, and (c) direct mechanical excitation of the cochlea (Tonndorf, 1972; Khanna *et al.*, 1976). Additional discussions are contained in Berger and Kerivan (1983), Berger (1985), and Ravicz and Melcher (2001). In general terms, BC refers to any pathway other than that of conventional air conduction. For example, sound passing through the open mouth and the soft tissues of the eustachian tube excites no bones except the ossicles, but it is still included under the rubric of bone conduction. However, Zwislocki noted that it might be more appropriately called body conduction, though for purposes of adherence to common convention he chose to title his paper ‘‘bone conduction.’’ We will use that more common term herein as well.

The BC thresholds in a sound field can be measured similarly to the minimum audible field, except that sound must be prevented from being transmitted via the conventional air-conduction pathways that begin at the earcanal. This can be accomplished by sufficiently occluding the earcanal, or by canceling sound that is present at the eardrum.

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The latter approach, employed by Schroeter and Els (1980), is uncommon and has its potential shortcomings since canceling sound at the eardrum, which is directly connected to the ossicles, might affect ossicular motion, and such motion is one of the BC pathways [i.e., pathway (b) as discussed above].

When one uses the method of sound blockage at the ear, the question will naturally arise whether sufficient sound has been excluded. Whatever technique is employed, artifacts are always possible. For example, Zwislocki used resonator earplugs whose tips were metal rods coated with latex and wax, that were inserted deeply into the bony meatus. These devices substantially eliminated vibration of the canal walls, which is the dominant path in the occluded ear below 2 kHz [pathway (a) as described above]. One might argue that with actual usable HPDs such reduction of this pathway is not feasible. This suggests that the low levels of BC that Zwislocki reported would likely not be observed in practice. Alternatively, in the procedure that Berger has used with success, a very heavy lead earmuff (more than 10 times the mass of a conventional earmuff) with unusually high band force (about twice the force of a conventional earmuff) was worn in conjunction with deeply fitted foam earplugs. A concern with this approach is that the inordinately high mass and force of the combination distorts the skull in a way unlike any actual HPD, potentially creating unrealistic BC limits. Thus, within our manuscript we have compared a variety of procedures from various reports to provide a range of estimates of the limits that are likely to prevail.

In studying this paper, the reader should keep in mind the following important caveat: all of the measurements in this paper are based upon optimum fitting of HPDs in a laboratory environment. In fact, the fitting might be termed "hyper-optimum" in that in some cases the devices used are fitted uncomfortably in ways that might not be feasible in practice for the sustained periods that would be anticipated in the real world. Thus, it is questionable whether such values of protection as the BC limits reported in this paper could ever be achieved for groups of users in occupational settings regardless of the degree of motivation, training, and supervision that was employed. For additional discussion of such matters see ANSI S12.6-1997, Berger *et al.* (1998), and Berger (2000).

II. PROCEDURES

The data in this report consist of published real-ear attenuation at threshold (REAT) values from Zwislocki (1957), Nixon and von Gierke (1959), Schroeter and Els (1980), Berger (1983), Ravicz and Melcher (2001), and the current research.[1] To the authors' knowledge, these few papers, together with the early work of von Békésy (1960), represent the sum total of the available data on direct measurement of the bone-conduction limits on human subjects in a free or diffuse sound field. Others such as von Gierke and Warren (1953) reported predictions based on BC thresholds derived from direct stimulation of the forehead via "sound tubes," but did not conduct measurements with the entire body or head irradiated in a sound field. Their computations, however, provide additional support of the empirical values presented by the other authors. Békésy's work, though exceedingly clever, yielded empirical data at only a few low frequencies, and therefore are not included in the discussions that follow. However, his results, like those of von Gierke and Warren, are in accord with the findings of the five studies that are specifically reviewed. Brief descriptions of the empirical studies follow.

A. Zwislocki (1957)

These experiments consisted of the measurement of REAT in a free sound field on groups of six subjects wearing solid earplugs in combination with heavy earmuffs, three of those same subjects wearing only resonator earplugs tuned to frequencies from 300 to 600 Hz, and one subject for the frequencies below 125 Hz. The BC limits were determined by the resonator earplugs below 400 Hz, the better performing of the resonator earplugs or the dual combination from 400 to 1500 Hz, and the dual combination only above 1500 Hz.

B. Nixon and von Gierke (1959)

Measurements were conducted in a free sound field with frontally incident sounds, as well as incidence on the back of the head. Measurements were in conformance with the standard for hearing protection attenuation measurements that was in effect at the time (ANSI Z24.22-1957). The attenuation of five different commercially available earplug/earmuff combinations was measured on eight subjects. Additional measurements were conducted by covering the subjects' heads in part or in total with medical cotton wicks of 8-in. width that were wound around the head and fastened with tape until the desired thickness of 2–3 in. was achieved.

C. Schroeter and Els (1980)

REAT measurements were taken using very large custom-built (approximately 30 000 cm^3) sound attenuation enclosures that coupled to the head circumaurally with conventional earmuff cushions, and contained cancellation speakers for the low frequencies. At and above 2 kHz the attenuation of the enclosures was complemented by deeply inserted foam earplugs. The tests were conducted on ten subjects. The 1980 reference (in German) cited above is the original, but those results which are also summarized in Schroeter and Poesselt (1986), p. 512, Table II, are more easily accessed.

D. Berger (1983)

REAT values for three earplugs (one of which was worn with three different depths of insertion) and three earmuffs were evaluated both singly and in various combinations according to ANSI S3.19-1974. Thirteen subjects participated in the entire experiment, with seven common to all tests. For any one test, ten subjects were measured three times each. All of the earplugs were commercially available. The one of greatest interest for the purposes of this work was the foam earplug (E•A•R® Classic® plug with a length of approximately 19 mm) which was inserted partially (Pl; about

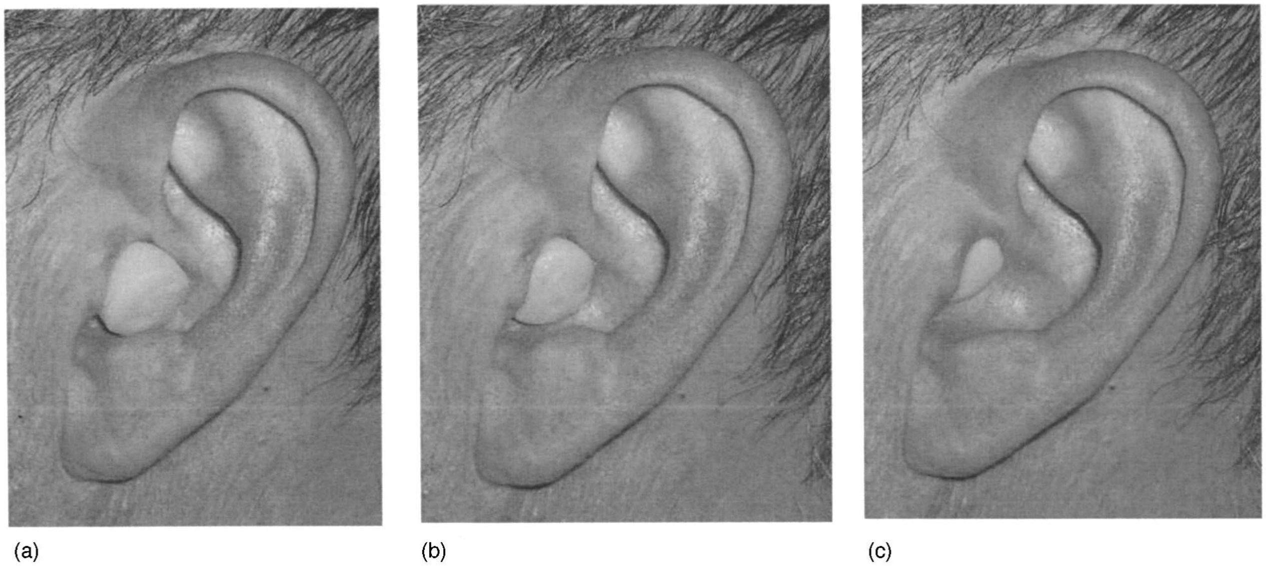

FIG. 1. (a)–(c) Partial (PI), standard (SI), and deep (DI) insertions of a Classic foam earplug in one subject's earcanal.

15%–20% of the plug in the earcanal), to a standard depth (Sl; about 50%–60% in the earcanal), or deeply (Dl; about 80%–100% in the earcanal). The DI was the maximum depth of insertion that a subject could tolerate before experiencing significant discomfort. The fit was measured by drawing black circumferential rings on the plugs at the intersection of the plug and the posterior entrance of the earcanal (i.e., the floor of the concha). All fitting was by the experimenter in conjunction with the subject. Figures 1(a)–(c) show a view of the plug inserted to the PI, SI, and DI depths.

Two of the earmuffs were commercially available and of substantially different size and mass. The third was a specially constructed damped lead earmuff with an extensional layer of a vinyl damping compound. Its total mass for both cups was 3300 g and volume (per earcup) 300 cm^3, with an inordinately high band force of 24 N. The lead earmuff was utilized for exploring BC limits. It is not a feasible product to wear outside the laboratory.

In combination with a small-volume earmuff Berger tested five different earplug conditions, each of which when worn individually provided four substantially different levels of protection. The dual combinations of the various plugs with the small earmuff also provided four substantially different values of protection for the frequencies below 1 kHz, corresponding to the increasing attenuation provided by the earplugs alone. At and above 2 kHz all combinations provided the same level of protection, which was also within 3 to 7 dB of the values reported by Zwislocki (1957) and by Nixon and von Gierke (1959) in their prior studies.

Berger's estimate of the BC limits to attenuation was taken to be the values found for the deeply inserted foam earplug worn in combination with the lead earmuff. Even though there were dramatically differing levels of attenuation for the three earmuffs in the study when used as a single hearing protector, when worn in combination with a DI foam earplug the earmuffs performed identically, i.e., results were essentially the same for all earplug-plus-earmuff combinations.

An ancillary observation that Berger made was that with one exception, the combined protection of a muff and a plug always exceed either of the individual devices at all test frequencies. The interesting exception was one premolded earplug (V-51R type) worn together with the large-volume earmuff. The combined attenuation of the plug plus muff was actually about 4 dB less at 1000 Hz than found for the earmuff alone. This could be attributed to the occlusion effect for this relatively short (and hence shallowly inserted) premolded earplug. Shallow fitting causes an amplification of the external-earcanal bone-conduction pathway (Berger and Kerivan, 1983), thus enhancing any vibrations of the earcanal walls caused by the sound field or by vibration of the earmuff itself.

E. Ravicz and Melcher (2001)

In the context of a study examining noise created by functional magnetic resonance imaging the authors explored the limits to attenuation. Their most protective condition consisted of foam plugs worn with conventional high-attenuation earmuffs, covered by a free-standing sound attenuating helmet large enough to fit loosely over the head of a subject wearing earmuffs. The helmet required a breathing tube and was sealed to the shoulders with a towel. Five subjects were tested using a REAT paradigm with unconventional stimuli consisting of tone bursts at frequencies from 500 to 2800 Hz. Microphone-in-real-ear (MIRE) data were also acquired, but those values are not reported herein.

F. Current research (this paper)

Similar procedures were employed as in Berger (1983) except that the subject count was increased to 16, with two instead of three measurements of each condition per subject. Fourteen of the subjects met the hearing sensitivity requirements of ANSI S12.6-1997, and the remaining two met the requirements at all but one frequency. With the exception of subject fitting and subject count (16 instead of 20) the pro-

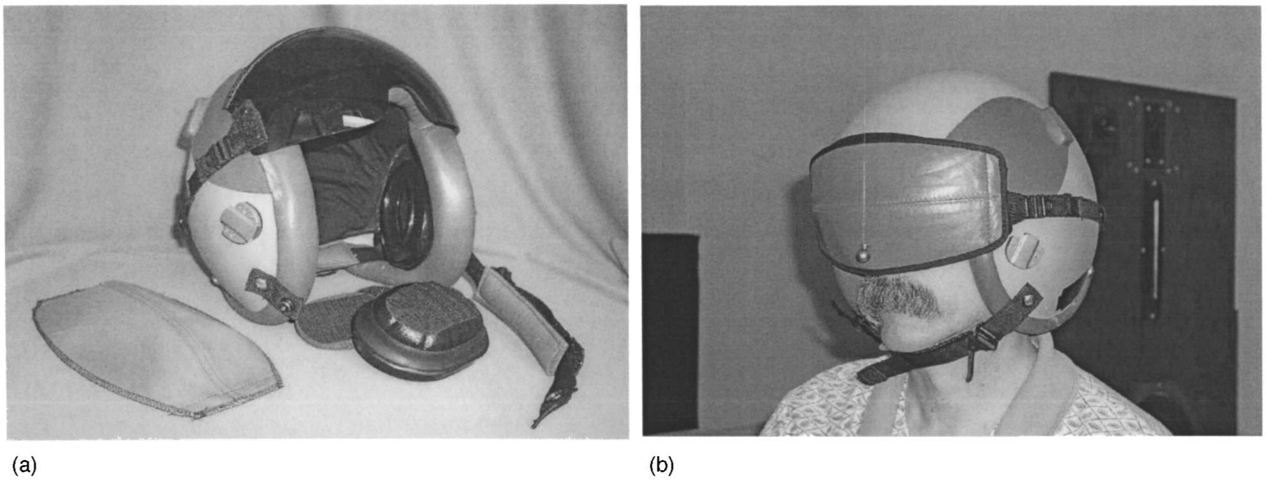

FIG. 2. (a) Gentex HGU-55/P flight helmet with one earcup removed, illustrating the hook and loop attachment material on the back of the cup and the foam/hook and loop spacers (partly beneath the cup) that are used for positioning. (b) The helmet on subject's head with visor and cover in position.

cedure conformed to Method A of ANSI S12.6-1997. All REAT measurements were conducted in the E•A•RCAL[SM] acoustical laboratory of Aearo Company. MIRE measurements on the active earmuff were made at Bose Corporation on a separate group of subjects in conformance with ANSI S12.42-1995 with two exceptions: only six subjects were tested, and measurements were made in a medium-sized audiometric booth lined with loudspeakers. Between the three fittings of the muff, subjects rotated their position in the booth to vary the incidence of the sound field. Measurements done in this way have correlated well with measurements in the Bose reverberation room.

The important new features of this series of experiments consisted of the devices. The E•A•R® Classic® Plus earplug (a longer version of the E•A•R Classic—24 mm vs 19 mm) was utilized and fitted deeply, and was worn in conjunction with one of two types of earmuffs. One was a prototype ANR headset made by Bose for U.S. military evaluation. It utilized large yet conventional circumaural earmuff cups, a conventional headband, and was an experimental adaptation of ANR technology from Bose's present commercial products for use in higher noise levels. The other was a Gentex lightweight fighter/attack aircrew helmet (HGU-55/P) with thick edge roll and internal plastic earmuff cups, as illustrated in Fig. 2(a). We utilized the medium and extra-large sizes of the helmet, and carefully inserted foam spacers between the cups and helmet shell, as is the norm for proper positioning. Larger volume cups and cushions from the Navy HGU-84/P helmet, rather than those normally fitted in the USAF HGU-55, were used, and communications were removed (i.e., receivers and cables) to maximize the attenuation achieved. The chin and nape-of-the-neck straps were adjusted for optimum fitting. All tests were conducted with the visor down in the operational position and covered by its fabric protection cover [see Fig. 2(b)]. Obviously the helmet could not be worn in such a condition, but the purpose herein was to explore the limits to protection; if need be a clear visor with higher transmission loss could always be devised. Note the foam edge roll at the rim of the helmet, and the resulting tight fit to the sides of the face and below the ears. However, protection of the lower frontal face, i.e., the mouth and jaw, was missing.

Once subjects entered the chamber they generally stayed in place for the entire set of measurements lasting approximately 90 min. Occasionally they took a brief break between the first series and the repeat series. The testing sequence was intended to limit fitting variability as much as possible to allow for the best comparison between plug-only and plug-plus-muff conditions. The testing was as follows:

(i) Flight helmet positioned, wait 2 min for acclimatization, and take occluded threshold.
(ii) Occluded threshold with ANR headset, electronics off.
(iii) Open threshold.
(iv) Occluded threshold with DI foam earplug.
(v) Occluded threshold, leaving DI foam earplug in place for this and following occluded thresholds, with the flight helmet donned as well.
(vi) Occluded threshold with the ANR headset, turned off, in place of the flight helmet.
(vii) Occluded threshold with the ANR headset left in position, but now turned on.[2]
(viii) Remove earplugs and ANR headset and repeat entire series.

Test signals were $\frac{1}{3}$-octave-bands of noise spanning the range from 125 Hz to 8 kHz at octave-band center frequencies. For one subject (the first author) test data were also acquired at 80 Hz. Although those values are not included in the subsequent analyses they indicate that the BC limits and other attenuation values at 80 Hz are essentially the same as those found at 125 Hz.

III. RESULTS

The data from the current study as well as key results from the literature summarized in the prior sections of this paper are presented in Table I.

TABLE I. Mean real-ear attenuation and standard deviation values in dB from this study, and mean values from prior published data.

Device		Frequency (Hz) 125	250	500	1000	2000	4000	8000
DI foam	Mean	39.9	44.4	47.8	43.7	37.4	44.4	47.0
	SD	5.5	4.8	3.5	4.2	2.9	3.7	4.7
ANR Muff OFF	Mean	18.1	23.7	25.9	30.1	36.1	43.3	43.3
	SD	3.5	1.8	1.6	2.6	2.9	2.8	3.1
Flight Helmet	Mean	20.6	24.1	29.5	41.7	46.5	56.6	58.6
	SD	4.2	6.0	3.5	4.3	4.4	3.2	4.9
DI foam+ANR Muff OFF	Mean	47.1	57.4	62.0	49.5	40.5	50.1	49.7
	SD	4.8	3.1	4.4	5.8	3.8	5.3	4.6
DI foam+ANR Muff ON	Mean	50.4	57.3	61.5	49.2	40.8	50.5	50.1
	SD	5.1	3.2	3.6	6.1	4.3	5.6	4.7
DI foam+Flight Helmet	Mean	42.0	50.5	60.8	53.6	48.6	60.2	61.3
	SD	5.6	4.0	4.2	6.6	4.0	5.1	6.6
Foam+small volume earmuff, S12.6-1997 Method B	Mean	26.1	27.0	34.4	38.6	40.4	50.8	47.3
	SD	6.5	6.1	6.3	5.5	5.5	5.4	4.4
Estimate of ''best possible achievable protection''	Mean	50.4	57.3	61.5	55.0	65.0	75.0	75.0
	SD	5.6	4.0	4.2	6.6	4.0	5.1	6.6
Zwislocki (1957) BC	Mean	51	60	68	60	46	54	41
Nixon and von Gierke (1959) plug+muff	Mean	30	33	38	40	42	52	41
Nixon and von Gierke (1959) plug+muff+head covered	Mean	31	35	41	45	52	68	58
Schroeter and Els (1980)	Mean	52	51	48	46	49	54	44
Berger (1983) BC	Mean	47	51	57	47	39	49	49
Ravicz and Melcher (2001) plug+muff+box	Mean			58	55	66		
Hachey and Roberts (1983) real-world plug+muff	Mean	21	24	31	35	35	43	38

A. Attenuation values for single HPDs

The attenuation values for the individual plugs and muffs of the current study are presented in Figs. 3 and 4 where they are compared to the BC limits from Berger (1983). Note that at all frequencies the attenuation of the deeply inserted Classic Plus, a longer version of the Classic foam earplug, exceeds that of the shorter version by 2 to 4 dB (Fig. 3). Statistical tests (*t*-test assuming equal variance) yielded significance at 500 Hz ($p<0.01$), and ''near significance'' at 250 and 1000 Hz ($p<0.06$ and $p<0.08$, respectively). Other differences were not statistically significant. The increased attenuation in the low frequencies may be in part due to the longer plug combined with a deeper fit, but also note that the two subject populations (10 in the prior study and 16 in the current study) are completely different. The slightly increased attenuation of the DI plug in this study, together with the fact that the plug's performance has a large impact on the attenuation of the combined devices, must be kept in mind when comparing the current BC estimates to those from Berger (1983).

Also of note in Figs. 3 and 4 is the sharp minimum of attenuation at 2 kHz in all of the head-not-covered conditions. This is a commonly observed feature of real-ear attenuation data for high-attenuation hearing protectors.

In Fig. 4 the attenuation of the ANR earmuff is compared to the small-volume and lead earmuffs from Berger (1983). Because ANR electronics generate residual low-level noise, which can mask the threshold test signals, they are not

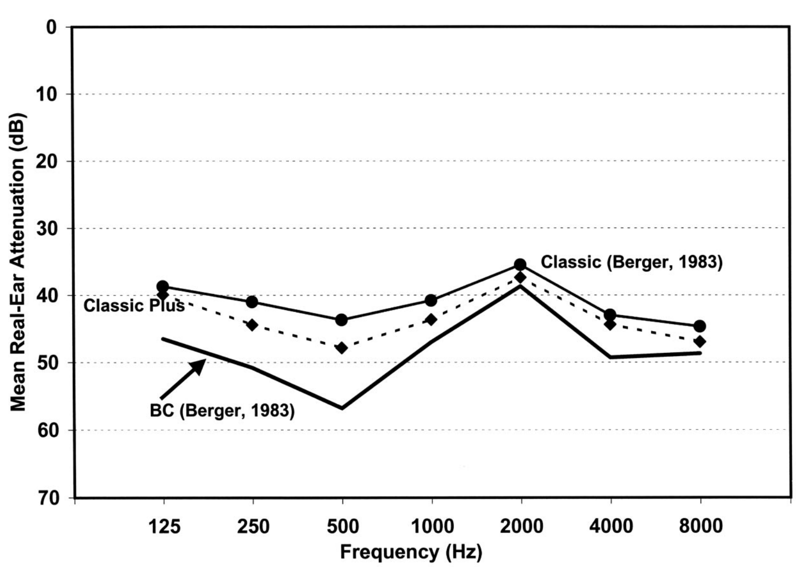

FIG. 3. Comparison of the real-ear attenuation of a deep insertion of the Classic Plus (this study) to the Classic, and to the BC limits, from Berger (1983).

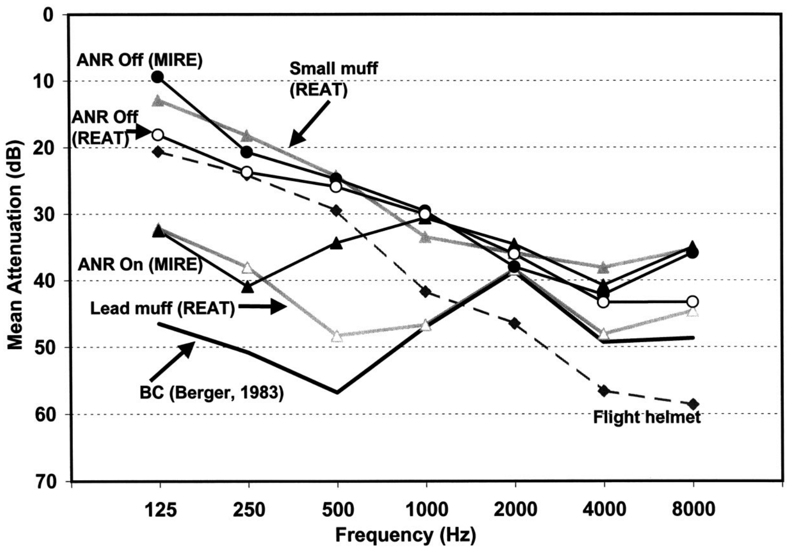

FIG. 4. Attenuation of the ANR earmuff and flight helmet, worn individually, from this study, compared to the muffs from Berger (1983). For the ANR earmuff both real-ear attenuation at threshold (REAT) and microphone-in-real-ear (MIRE) data are compared.

amenable to REAT testing. Hence for that earmuff's on-condition, MIRE data (measured at Bose's laboratory) are reported. To provide the best comparative data for the off-mode, the muff was tested with both MIRE and REAT procedures. The REAT measurement was possible for the ANR device in the off-mode since in that condition the electronics generate no residual noise. REAT measurement is also possible when an ANR device is turned on and worn over earplugs, as is reported later, since the earplugs attenuate the self-noise of the ANR device below the subject's hearing threshold.

Comparison of the dark sold lines with the filled and open circles illustrates the REAT vs. MIRE measurements for the ANR earmuff. The differences, due to low-frequency physiological-noise masking that elevates the occluded threshold (Berger and Kerivan, 1983), are of the pattern which are expected, though the magnitude of the effect, 9 dB at 125 Hz, is larger than anticipated for a muff of this size. A 5-dB REAT-MIRE difference at 125 Hz is more typical (Gauger, 2002). If it were possible to also produce REAT data for the ANR earmuff in the on position, they too would be expected to show increased low-frequency attenuation relative to MIRE values. Hence it is likely that the performance of the ANR device in the on-mode would have exceeded the REAT-measured attenuation of the lead earmuff at 125 and 250 Hz by a greater amount than shown in Fig. 4, if both were measured in the same way. In the off-mode the performance of the ANR earmuff was similar to the small-volume earmuff.

The flight helmet data are similar to earmuff attenuation values up through about 1 kHz, but quite different from the circumaural earmuffs in the high frequencies. Note that at and above 2 kHz, the flight helmet exceeds Berger's 1983 estimate of the BC limits for circumaural devices. This is not surprising since in the high frequencies the important BC pathway has been shown to be direct transmission to the cochlea bypassing the external earcanal. If the skull is shielded, less energy will be incident on the head and consequently less will be transmitted to the inner ear via this flanking pathway (Khanna *et al.*, 1976; Ravicz and Melcher, 2001).

B. Individual subject data

Before presenting the averaged data for the dual protection combinations from the current study, it will be illuminating to review the data for the 17 individual subjects in Figs. 5–7. All the subjects except for KLD were included in the subsequent analyses. KLD has had recurring ear infections since childhood and an abnormal tympanogram indicating poor tympanic mobility, but was included because his open-ear audiometric thresholds are near normal and he could be representative of a person required to wear dual protection. Also in the past, KLD has been well-fitted with foam earplugs and obtained representative attenuation data. Although his current data are aberrant, they might be experienced in the real world. Note that with respect to the foam earplug, as shown in Fig. 5, KLD experienced unusually low values of attenuation at 500, 1000, and 8000 Hz. The boxes indicate his retest data on a subsequent day. Even excluding KLD, there is a 15- to 20-dB range in data across frequencies. Thus, even though this well-fitted foam earplug delivered 25 or more decibels of attenuation at all frequencies for all subjects in this study, and more than 30 decibels at all frequencies for all *but one* subject, the actual range in attenuation values is large.

Figure 6 demonstrates the spread for the ANR earmuff, turned off, i.e., worn as a conventional earmuff. Note the reduced range in values compared to the results for the earplug. This is not surprising since earmuffs normally fit more uniformly across groups of listeners, producing less attenuation variability. The data for the flight helmet (not shown) exhibit variability greater than that of the earmuff and closer to the foam plug. Although the helmet contains circumaural cups, once the cups are ''hidden'' under the helmet, the positioning and adjustment is more problematic than with a conventional headband-mounted earmuff unit.

Finally, Fig. 7 shows the effect when the DI foam plug is combined with the flight helmet. As with the earplug alone,

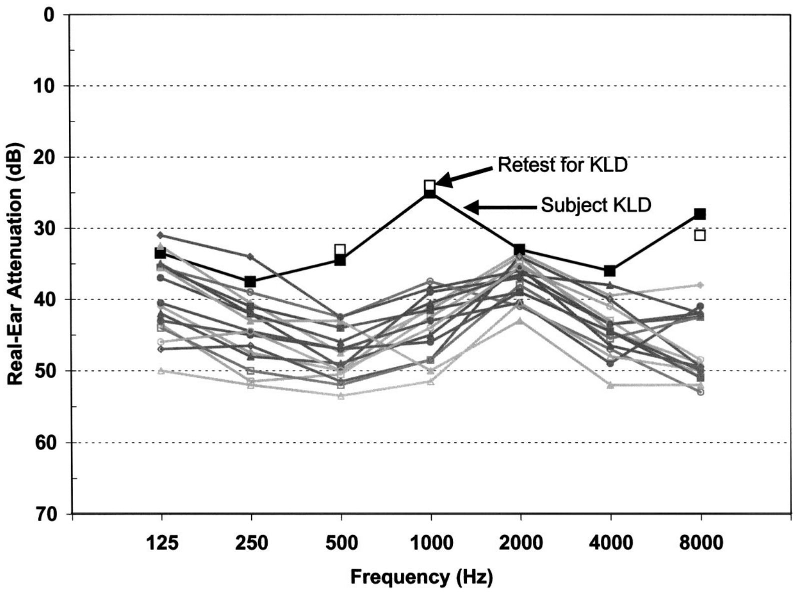

FIG. 5. Real-ear attenuation for foam plugs with a deep insertion (DI). Each line represents the average of two measurements on one subject. The ‘‘aberrant’’ subject KLD is specifically noted.

KLD is an outlier, as might be expected since the earplug strongly controls the combined attenuation of the devices. When the data for the flight helmet alone are examined, KLD is low, but only by a few decibels from 1000 Hz on upwards in frequency.

C. The value of ANR in a dual-protection scenario

Figure 8 demonstrates the performance of the DI foam earplug plus ANR earmuff, in its on- and off-modes, as compared to Berger's prior BC estimate. Turning first to the question of the value of the active noise reduction, note that the ANR-on and ANR-off values of attenuation in Fig. 8 are essentially identical at all frequencies except 125 Hz (3.3-dB difference, significant at $p<0.05$ using *t*-test for paired samples), the frequency at which ANR is typically most effective. Recall that the ANR earmuff was left in position on top of the earplug and tested turned off and then on, in immediate sequence. This eliminated fitting as a cause of potential variability between the two conditions. Although significant, the increase in protection at 125 Hz is of questionable importance due to its small magnitude and limited frequency range. The small increase provided by ANR in a dual-protection scenario with the DI plug is presumably due to the same reason that increasing muff size offers no dual-protection benefits. Transmission through a small-volume muff and DI plug, or transmission through the ANR earmuff in the off-mode when combined with the DI plug, is already equivalent to the BC limits at most frequencies, so the added attenuation of a more protective earmuff or condition is inconsequential.[3]

The current dual-protection data for the ANR earmuff demonstrate values that equal or exceed Berger's prior estimate obtained using a lead earmuff as seen in Fig. 8, with the differences achieving statistical significance at $p<0.05$ (*t*-test assuming equal variance) for the frequencies from 125 to 500 Hz. This is not surprising since the earplug is such an important component of the dual-protection system, and as shown in Fig. 3 the DI earplug fit in this study exceeds the

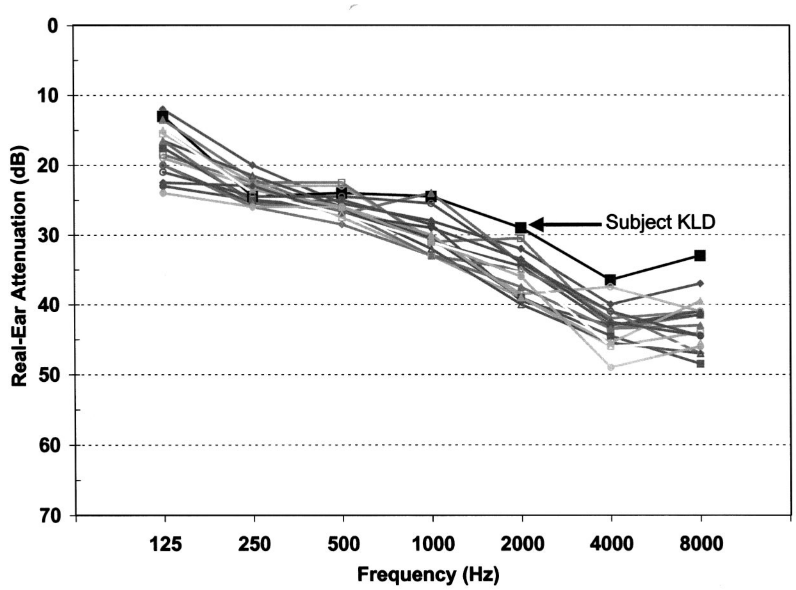

FIG. 6. Real-ear attenuation for ANR earmuff with the ANR off. Each line represents the average of two measurements on one subject. The ‘‘aberrant’’ subject KLD is specifically noted.

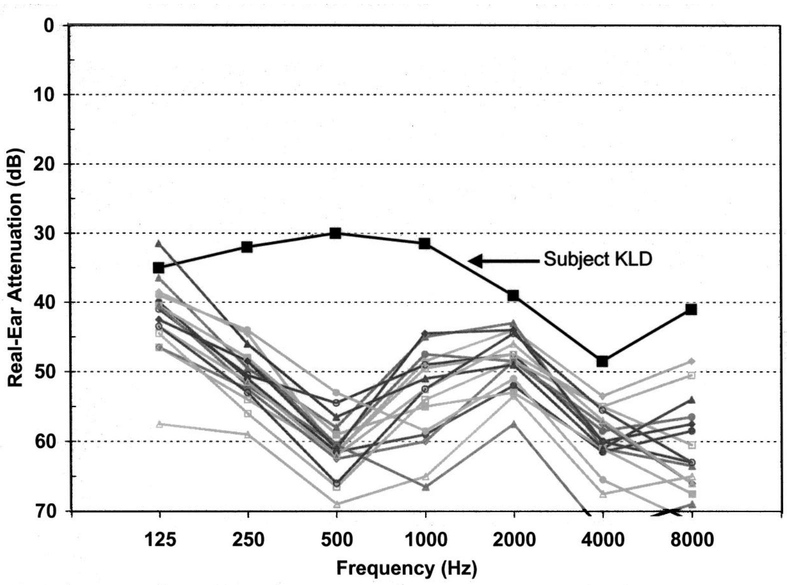

FIG. 7. Real-ear attenuation for foam plugs with deep insertion (DI) plus flight helmet. Each line represents the average of two measurements on one subject. The "aberrant" subject KLD is specifically noted.

performance of the one previously employed by Berger. Additionally, note that the difference between the studies is largest below 1 kHz. Khanna *et al.* (1976) concluded that the primary BC path at these frequencies is through the skull to the walls of the earcanal which vibrate and reradiate the sound; the deeper insertion resulting from the longer plug used in the present study may have changed the BC limit itself. Finally, the differences between the studies may also be explained by the attributes of the current subject pool, and the possibility that their inherent BC limits differ slightly from the group used nearly 20 years ago.

D. Limits to attenuation

Figure 9 presents the best estimates of the limits to attenuation from the current study, based on the DI foam plug worn with the ANR earmuff in the on-mode, and the DI foam plug worn with the flight helmet with the visor down and covered. The values are compared to the DI foam earplug worn alone and to one of the higher attenuation conventional earmuffs we have measured in our laboratory. Unless the head is shielded, switching from single to dual protection can only achieve modest gains of about 6 dB at and above 1 kHz before reaching the BC limits (compare plug+ANR muff on, to the single-HPD curves). However, at the low frequencies gains of 10 to 15 dB are realizable.

By covering the head, as accomplished in this study using the tight-fitting flight helmet with face plate (visor), additional gains above single protection of from 4 to 11 dB are possible from 1 kHz and up. However, the attenuation at 125 and 250 Hz suffers relative to the optimum low-frequency combination of a DI foam earplug plus circumaural earmuff.[4]

IV. LIMITS TO ATTENUATION: CURRENT VERSUS PRIOR DATA

The data in Fig. 10 compare the four previously published directly measured estimates of the BC limits to those from the current study, for the head-not-covered condition. The current data and the Berger (1983) values were measured in a diffuse field; the others were frontal incidence free field.

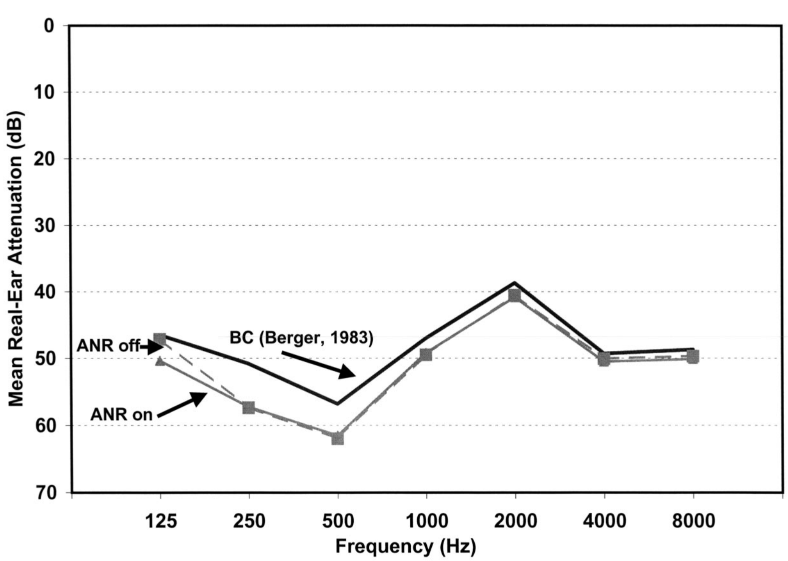

FIG. 8. Real-ear attenuation of the ANR earmuff worn in combination with a deeply inserted foam plug (DI), with the ANR turned on or turned off, as compared to Berger's (1983) prior BC estimates.

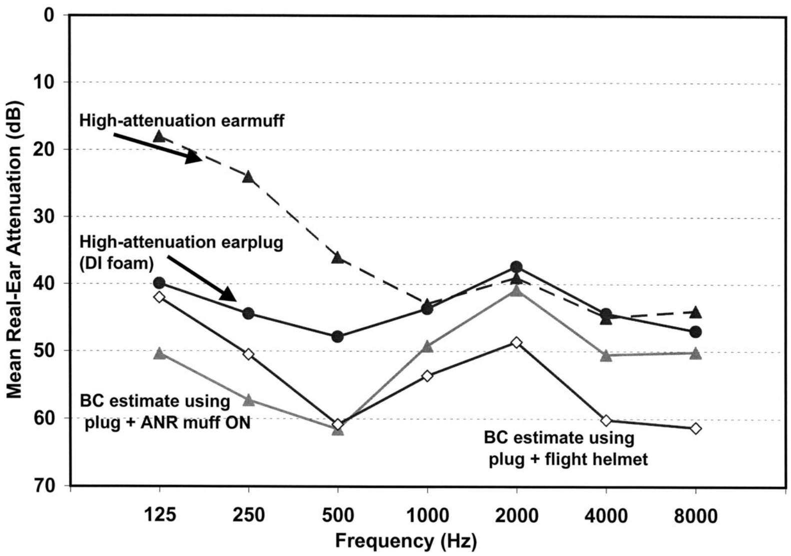

FIG. 9. Various estimates of the BC limits with head exposed versus covered, from the current study.

Zwislocki, who used very rigid deeply fitted plugs with muffs, obtained the greatest noise reduction, i.e., his are the highest estimates of the BC limits at most frequencies. However, Zwislocki relied upon only three to six subjects depending upon the test frequency and test condition. One could take the few subjects with the greatest attenuation in the current study and construct an estimate within a couple of decibels of the Zwislocki data at all frequencies. In all likelihood, the reason that the data of Nixon and von Gierke diverge from the other studies below 2 kHz is that the premolded earplugs available to them in the 1950s could not provide the high levels attenuation, especially in the low-frequencies, that are available using today's foam earplugs. (In fact Nixon and von Gierke also discounted the validity of their BC estimates below 2 kHz because of how their values compared to the prior work of von Gierke and Warren, and of Zwislocki.) The unusually low value for Schroeter and Els at 500 Hz may be attributed to their decision not to have the subjects wear an earplug underneath the attenuation boxes for the frequencies below 2 kHz. Berger's data and the current study present the highest values at 8 kHz, perhaps due to their sound field conditions that differed from the other studies.

Because of the extreme measures employed by Zwislocki, with different types of plugs used to acquire data at different frequencies, and the type of seal he achieved in the bony meatus, his data may indeed define the true BC limits with uncovered heads. However, in all likelihood his values are unachievable for groups of subjects with any single combination of wearable HPDs, regardless of how well they are fitted.

Figure 11 addresses the question of the gains to be made if we cover the head so that sound conduction pathways of the skull are blocked to varying degrees, thus limiting the flanking energy to transmission via the lower face, the neck, or the torso. In the current study we shielded the head by using a flight helmet and were able to measure data across the entire range of conventional test frequencies. This provided increases in the BC limits at and above 1 kHz, with substantial improvements of from 4 to 11 dB available at and above 1 kHz. Nixon and von Gierke accomplished the goal of shielding the head by wrapping it in ‘‘medical cotton

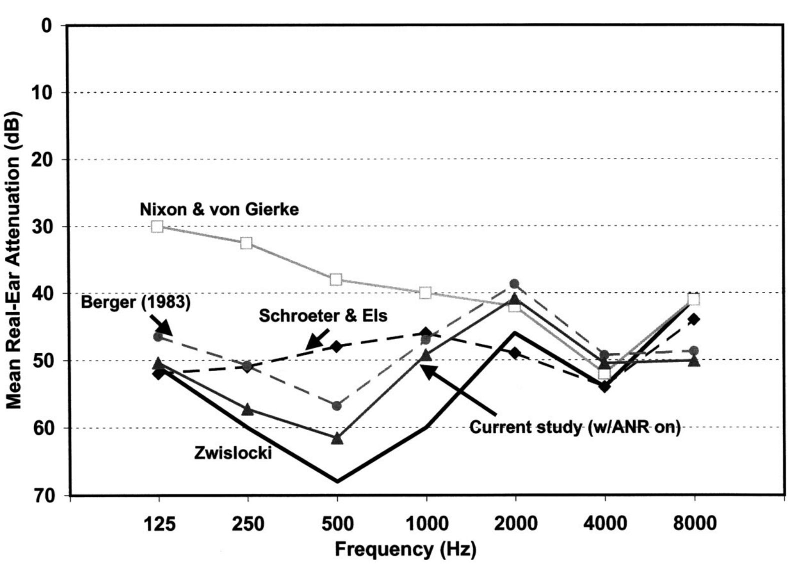

FIG. 10. Various estimates of the BC limits to attenuation with head exposed, from this study and the prior literature.

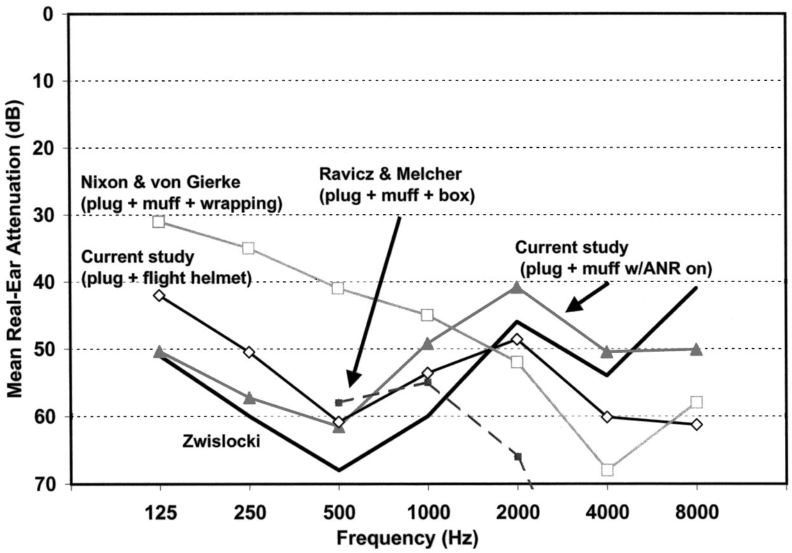

FIG. 11. Various estimates of the BC limits to attenuation with head exposed versus covered from this study and the prior literature.

wicks'' to a thickness of 2 to 3 in. As mentioned with regard to Fig. 10, below 2 kHz their data should be disregarded because of limitations of the earplugs they utilized. However, from 2 kHz and up, note in Fig. 11 the relatively close correspondence between the cotton-wrapping technique and the flight helmet used in this study, though the cotton appears to perform better at 4 kHz.

Another effect observed in the current study was the importance of covering the face (even the partial covering provided by the flight helmet) when listening in a diffuse sound field with a helmet shielding the remainder of the head. This was illustrated by testing the attenuation on one subject (the first author) with the DI foam plug plus helmet, with the visor in the up and down positions. A loss of attenuation of from 2–4 dB at and above 2 kHz was observed, suggesting that even with a helmet, facial shielding is important for maximum protection.

The most extreme approach to shielding the head was taken by Ravicz and Melcher (2001), who utilized a helmet completely separate from an earmuff worn beneath it. The helmet fully enclosed the head and chin, sealing to the neck. Even when the muff and helmet (box) were worn in combination with earplugs, the earplugs provided the dominant portion of the attenuation in the lower frequencies. Because the fit of the foam plug utilized by Ravicz and Melcher was closer to PI than DI, this led to estimates of attenuation at 500 and 1000 Hz that were unremarkable. However, at the very high frequencies where each of the three hearing protectors was an effective sound block, the noise reduction achieved by the combination became exceptional; 68 dB at 2 kHz and 82 dB at 2.8 kHz. The gain they demonstrated of about 20 dB, by ''completely'' shielding the head at 2 kHz, is in close agreement with the prediction of von Gierke and Warren (1953) which was based upon irradiating small segments of the forehead and sternum with sound from a metal tube in order to measure hearing thresholds.

A few additional points merit discussion. When maximum attenuation is desired, the concern may also arise regarding sound transmission through the nose and mouth into the eustachian tube. Would it be necessary to keep the mouth shut and plug the nose in extreme environments? Von Gierke and Nixon addressed that question by doing all testing with mouths shut, and then including a condition when the subjects heads were shrouded in cotton, of a threshold with nostrils pinched shut. The average effect was no more than about 2 dB at any frequency, although one subject experienced a 5-to-7-dB effect. We hypothesize that might have been due to a patent eustachian tube. To examine this, one subject (first author) was tested in a dual-protection mode with mouth closed and nose open, mouth closed and nose closed, and mouth open and nose open. No differences of more than 2 dB were observed at 500, 1000, 2000, and 4000 Hz.

Since the first author can also open his eustachian tubes at will, he listened in high-level broadband noise with mouth shut and nose open or pinched shut, while opening his tubes. Subjectively, attenuation was clearly affected in the high-frequency range when his tubes were open, while he alternately pinched or opened his nostrils. The effect seemed in the range of 5 dB. An attempt was made to measure this effect by tracking his $\frac{1}{3}$-octave band thresholds, but this was problematic due to an artifact. When he opened his eustachian tubes it also caused his acoustic reflex to trigger, thereby increasing his low-frequency physiological noise and to some extent masking and elevating his thresholds. The measured threshold shifts yielded an apparent 5-dB reduction in attenuation at 2 kHz and an increase at 4 kHz, but these results may have been contaminated due to the physiological-noise artifact. The effects were of the magnitude reported by Nixon and von Gierke for their ''worst case'' subject. Recall, however, that for maximum attenuation the head must be shielded, as in using a helmet, and under such conditions the nose would be shielded as well.

V. A WORD ABOUT SINGLE NUMBER ATTENUATION VALUES—NRRs AND SNRs

To place the BC limits in perspective relative to the type of attenuation factors that are often associated with HPDs, the noise reduction rating (NRR) and the single number rating (SNR) were computed for several of the measurements.

These ratings are weighted attenuation values averaged across frequencies and then adjusted to represent what 98% of the test subjects obtained in the case of the NRR (a 2-standard-deviation adjustment), and what 84% obtained (a 1-standard-deviation adjustment) in the case of the SNR (Berger, 2000). NRRs and SNRs are subtracted from C-weighted sound exposures to estimate the protected exposures in terms of an A-weighted value. For the head-not-covered conditions, the NRR for the BC limits is approximately 34 dB, and the SNR about 42 dB. With the head covered the values increase approximately 7 dB, to about 41 and 49 dB respectively.

VI. DISCUSSION

Although in most noise exposure scenarios, the protection afforded by an individual well-fitted earplug or earmuff will be sufficient, there are exposures in which more protection is either desired or required. For example, while shooting handguns, a single well-fitted high-attenuation earplug or earmuff is generally adequate, but for reasons of comfort or to reduce flinching and improve shooting, the additional protection gained from two products is desirable. In 1983 Berger explored various combinations of HPDs in detail and provided suggestions for the selection of each of the devices to be used in a dual protection scenario. In this paper we reexamined this issue to uncover the absolute maximum hearing protection that can be provided, in order to address current and new military systems as well as exceptional occupational situations with extreme noise exposures. And we also asked, "if the dual protection isn't enough, is there anything that can be done about it?"

We have reviewed the handful of extant studies published during the past 50 years and added recent data of our own, to provide insight into the limits to attenuation. The answers are summarized in Figs. 10 and 11 and in Table I.

For a head-not-covered condition at frequencies from 125 Hz to 4 kHz the best estimate of the limits to protection is the range of values encompassed by the Berger (1983) data and the Zwislocki (1957) results, though it is unlikely that with any single set of plugs and muffs that the Zwislocki data can be achieved. For a realistic estimate, in a diffuse field with a single pair of well-fitted muffs and plugs, a better choice would be encompassed by the data of Berger (1983) and the current study. These values range from 45 to 60 dB, except at 2 kHz where a minimum is observed that hovers around 40 dB. At 8 kHz the Berger and recent estimates provide the highest measured BC limits, and also those most appropriate for diffuse sound fields.

The prior work of Nixon and von Gierke (1959) and the recent test on one subject by the current authors suggests that whether or not the nostrils are open or closed is unimportant. However, anecdotal evidence provided by McKinley (2002) raises another issue. He has observed in high-level steady-sound fields of approximately 140 dB SPL that those wearing dual protection notice a difference in the perceived sound level depending on whether they have their mouths shut, with jaws either relaxed or clenched. He observed that clenching the teeth increases the sound transmission, i.e., reduces the attenuation. The authors of the current paper were unable to confirm this finding based on two of our authors listening in a 90-dB SPL diffuse sound field, but the third author did experience a small effect of clenching his jaws, estimated to be in the range of 3 to 4 dB in the middle frequencies.

For the absolute maximum in attenuation, the head must be acoustically shielded in addition to wearing dual protection. Shielding with a lightweight acoustically leaky barrier such as a safety helmet has no effect. What is required is a tight fitting helmet, like a standard military flight helmet, that encloses the entire skull and face. In such cases, one can achieve the values in Fig. 11 as obtained in this study with the DI foam plug plus flight helmet. Note that this requires the visor be down, otherwise 2–4 dB of protection may be sacrificed in the upper frequencies. Figure 11 also shows that with the highest-attenuating plug and muff combination (head not covered), attenuation is lowest at 2 kHz. Since extreme noise levels tend to be dominated by energy at higher frequencies, increasing the attenuation of noise reaching the skull should be a primary objective in providing protection in these environments. The authors are not aware of much engineering effort to date to develop a head and face noise-shielding helmet that can be practically worn (i.e., while addressing comfort considerations). Clearly, better shielding by a more complete helmet can potentially achieve much improved protection in the high frequencies, as demonstrated by Ravicz and Melcher (2001).

An estimate of the "best possible protection," i.e., with the ears plugged and the head and neck fully enclosed, can be made by combining the low-frequency limits (125–500 Hz) for the DI foam plug plus ANR earmuff in the on-mode, with the Ravicz and Melcher data for 1 and 2 kHz. To extend the estimate upwards in frequency, we make the conservative presumption that if 82 dB can be achieved at 2.8 kHz, then at least 75 dB can be obtained at 4 and 8 kHz. NRRs and SNRs computed from such data using the standard deviations for the DI foam plus flight helmet (see Table I) are 46 and 55 dB, respectively. This represents an increase of 12 to 13 dB over the DI foam plug plus ANR earmuff. In an actual noise spectrum typical of that to which crew are exposed while launching modern Navy jets on an aircraft carrier (a relatively flat spectrum with a difference of about +1 dB between the C-weighted and A-weighted sound levels), the noise reduction computed for 84% of the test population (a minus 1-standard-deviation adjustment) is about 41 dB for the DI foam plug plus ANR earmuff. The computed noise reduction for the best possible protection (see Table I) with a complete head enclosure is 51 dB. Whether or not such a "theoretical" protector could ever be fielded and worn in practice, it provides a benchmark for the limits to protection.

Finally, we turn to the thorny question of the real world. All of the forgoing measures were based upon compliant and well-trained subjects, willing to undergo some degree of discomfort while optimally fitting perfectly functioning hearing protection or special test devices, for short periods of time under pristine laboratory conditions. Temperatures were moderate and no exertion was required. When occupational situations or adverse military conditions, such as servicing jet aircraft with noise levels of 150 dB, are repeatedly en-

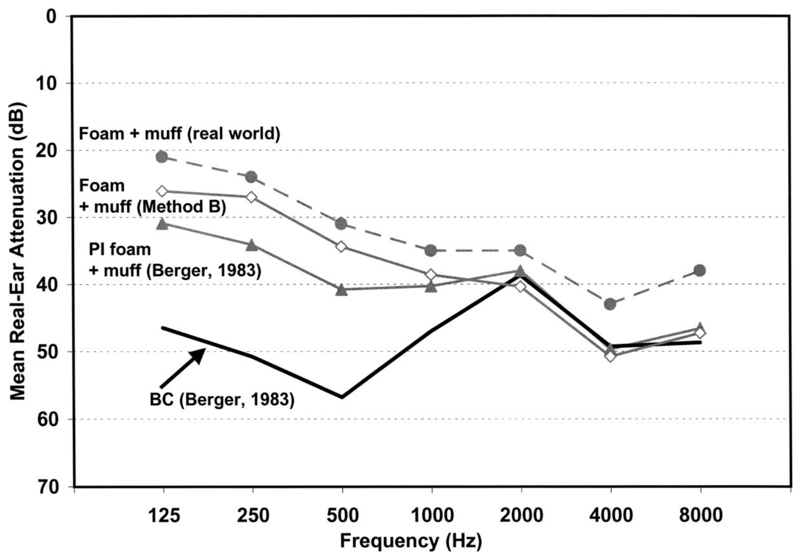

FIG. 12. Comparison of laboratory-estimates of the BC limits versus the actual protection in the real world for a foam plug plus small earmuff.

countered throughout a workshift, the situation is very different. Little indication of the real-world performance of maximum-attenuation dual hearing protection systems is available based on field studies, except for one report by Hachey and Roberts (1983) based on ten employees in a textile plant. Another indication is provided by testing to the new American National Standard S12.6-1997 under its Method-B procedure, which uses subject fit with naïve listeners to provide an estimate of field performance. Both of these data sets are compared in Fig. 12 to the PI foam plus small-volume muff data from Berger (1983) as well as to the Berger estimate of BC, which is the most conservative of the estimates provided in Fig. 10.

Both the real-world measurement and the real-world estimate (Method B) demonstrate attenuation up through 1000 Hz that is substantially less than even the PI foam plug under an earmuff, let alone the BC limits. Above 1000 Hz the disparity is somewhat reduced. Nevertheless the message is clear: approaching the laboratory BC limits in actual occupational situations is going to require very serious motivation, training, supervision, and enforcement. Whether it can be achieved is open to question.

As we move forward to examine new technologies we should be well grounded in the learnings from existing data and experiments. For example, ANR technology in earmuffs with communication systems provides certain advantages, but in terms of maximum protection when combined with a well-fitted earplug, there appears little to be gained. Efforts are moving forward to develop custom-molded earplugs with built-in transducers for communication, or ANR earplugs, or even helmets that might cancel skull vibrations. One should be aware of the limitations. Custom inserts may provide enhanced comfort and in some cases approach the attenuation of deeply inserted foam earplugs, but they will not exceed it. Also, alternative technology using foam eartips already exists and has been shown to afford excellent performance (Ribera *et al.*, 1996). ANR earplugs may extend the effectiveness of noise cancellation into higher frequencies than ANR headsets, but this is unlikely to have an important effect on the high-frequency BC limits, which Khanna *et al.* (1976) showed are due to flanking pathways that are largely outside the reach of earcanal-based systems. Helmet-based ANR systems that cancel bone vibration will probably be very complex to devise, and although they may produce some reductions in the BC-transmitted energy in the high frequencies, as can be seen in Fig. 4, even the best of helmets provide little attenuation at the frequencies below 1000 Hz. Furthermore, above 1000 Hz, the simple shielding that a flight helmet currently offers may be all that is needed, unless one takes the next step of fully encasing the chin and neck (see Fig. 11).

In the meantime, the best gains available to us today may well be in the motivation, training, and supervision of the users of hearing protection to make sure they get the most out of the devices that are currently available.

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[1]REAT measurements are conducted at low sound levels that never exceed about 80 dB SPL even for HPDs with maximum values of attenuation. Thus, the BC pathways are stimulated within the range in which they have been shown to behave linearly (Khanna *et al.*, 1976). However, in the extreme 150-dB environments in which maximum dual protection is warranted, one might ask whether the BC response is still linear. We are aware of no data that bear on this question. In all likelihood, if the responses are nonlinear, the effect would be compressive since mechanical systems that are overdriven tend to produce less output, i.e., as the sound field excitation increases, the increase in the BC response does not keep pace. In this case, the cochlear excitation would be less than predicted by simply subtracting the REAT-measured attenuation of the HPD from the noise exposure. Therefore, any errors arising from application of the method of this report would tend in a conservative direction, meaning that noise-induced hearing loss associated with sound transmitted via the BC pathways, while hearing

protection is being worn, would be expected to be no more severe, but possibly less severe than otherwise anticipated. Experiments are underway to address these questions (McKinley *et al.*, 2003)

[2]For those familiar with threshold procedures such as REAT measurements it might seem problematic to conduct an REAT measurement on an ANR system, since active devices emit low residual noise levels even when operating in extremely low ambient noise levels. This was not a problem, however, in the measurement of our dual-protection system as discussed in Sec. III A.

[3]Subsequent to the experiments conducted for this paper an additional ten-subject study was completed to evaluate the effects of ANR when worn with a PI foam plug. This was to answer questions of ANR performance in dual-protection scenarios, not to address the focus of the current study. As anticipated due to the lower attenuation of the PI fitting in the low and middle frequencies, the benefit of ANR was more pronounced at 125 Hz (7 dB vs. 3 dB with the DI fitting) and was still statistically significant at 250 Hz showing a 2-dB effect. As in the DI condition no benefits were observed above 250 Hz.

[4]Note the unexpected observation that in the on-mode, the ANR muff-plus-plug combination outperforms the helmet-plus-plug combination at 125 and 250 Hz. The same applies in the off-mode for the ANR muff-plus-plug versus the helmet-plus-plug (see Table I). This is true, even though as can be noted in Table I, the attenuation of the ANR muff in the off-mode, and the helmet, are essentially identical at those two frequencies. Our hypothesis is that at the low frequencies, the helmet acts as an acoustic antenna that couples the sound-field vibrations more closely to the earmuff cup than is the case for conventional circumaural cups directly exposed to a sound field, and this bone-conducted energy flanks the earplug. At the frequencies above 500 Hz, the foam/hook and loop pads that affix the cups to the helmet shelleffectively decouple the helmet from the cup, allowing for increased attenu-ation.

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