Understanding Noise-Induced Auditory Damage

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Noise-induced hearing injuries are pervasive in the United States, and large numbers of military personnel who train or carry out missions under high-noise conditions have particularly been affected. Understanding the impacts of noise on the auditory system and developing metrics to predict the likelihood and severity of these impacts are key to developing hearing protection devices that will prevent, or mitigate, hearing impairments. Lincoln Laboratory is performing R&D to improve this understanding and to create computational models that can inform the development of effective hearing protection for warfighters.

Noise-induced hearing injuries (NIHI), including tinnitus and hearing loss, are among the most common disabilities suffered by active-duty warfighters and reported by military veterans. Although noise-exposure risks can vary greatly across military branches and occupations, most service members are regularly exposed to noise hazards through training and operational duties during their careers, and the resulting auditory impairments are unfortunately common. For example, tinnitus and hearing loss have long been, and continue to be, the two most common disabilities compensated by the Department of Veterans Affairs (VA). In fiscal year 2016 alone, the VA awarded 149,429 new tinnitus compensation benefits and 77,622 new hearing loss disability benefits [1].

Hearing impairment and auditory-perceptual dysfunction resulting from noise exposure can reduce situational awareness by degrading sound detection thresholds, localization accuracy, and speech intelligibility. For the warfighter, these impairments not only can threaten mission success and survival, but also can adversely affect post-service life. Ironically, hearing protectors can induce similar detrimental effects through the attenuation that they provide, leading to their lack of acceptance by warfighters. A small number of studies have investigated the effects of hearing impairment or hearing protection on military operational performance during training and simulated operations, and in all cases, these studies found not only reductions in the subjects' ability to communicate, detect, and localize sounds [2] but also lethality in a combat exercise [3].

Such effects may result directly from peripheral auditory impairments that reduce input to the brain, indirectly from central deficits that compromise the neural processing of auditory information, or from a combination of both.

To limit noise exposure and reduce the risk of NIHI, hearing-protection devices (HPDs) are made available to warfighters in training exercises and during operational duties. However, these devices often are disliked and consequently are not worn, in part because the devices can compromise auditory situational awareness and degrade speech intelligibility [4]. Another challenge in addressing the risk of NIHI is the difficulty in accurately quantifying the noise exposure experienced by a warfighter during a mission or training exercise, or even in a standard work day. Noise-dosimetry devices are intended to measure noise exposure levels, but a variety of confounding factors—including device placement, environmental noise conditions, suboptimal hardware components, and inadequate predictions of auditory risk—can result in the dosimeters producing a poorly informed assessment of NIHI risk for a given situation.

Auditory Physiology and Damage Mechanisms

While high sound-pressure-level events such as explosions are capable of causing damage to the outer or middle ear, e.g., a ruptured tympanic membrane, i.e., ear drum, NIHIs are most often considered with respect to the inner ear, or cochlea. Mechanical signals (transmitted from sound-pressure wave energy in the middle ear) are transduced into electrochemical signals within the cochlea, and these signals are communicated, in turn, as nerve impulses via the auditory nerve to the brain (Figure 1). On the basilar membrane within the cochlea is the organ of Corti, lined by specialized primary auditory receptor inner and outer "hair" cells. Hair cells play a crucial role in hearing: in response to mechanical vibration that displaces their hair-like stereocilia, they transduce that energy into electrical current to the auditory nerve. Noise-induced hearing loss can occur when strong pressure waves (from very loud sounds) damage or destroy hair cells and their transduction machinery, or their associated neural connections. In humans, damaged and lost hair cells cannot regenerate, so the resulting hearing impairment is permanent.

Historically, hair-cell loss has been the focus of most research efforts to explain noise-induced hearing loss. Inner hair cells transform sound waves into signals for relay to the auditory nerve, brainstem, inferior colliculus, thalamus, and auditory cortex. Each inner hair cell is innervated by many afferent (ascending) nerve fibers—approximately 30,000 per ear—which are activated by neurotransmitters released across the synaptic space. Stimulation of the inner hair cells is the primary trigger for acoustic information to be sent to the brain. Inner hair cells can be damaged directly by loud noise and have additional susceptibility for noise-induced damage at their synapses with the auditory nerve.

Outer hair cells function primarily to amplify sound in a frequency-dependent, compressive, nonlinear

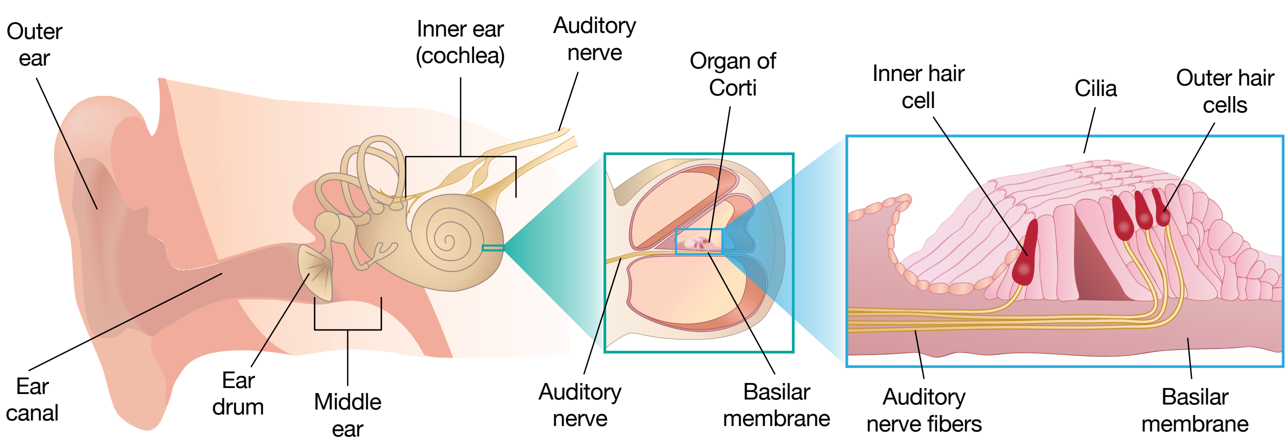


FIGURE 1. The auditory periphery includes the outer, middle, and inner ears (left). At center is an exploded diagram of the cochlea, and at right is a cross-section of the organ of Corti within the cochlea (reused from [5]).

manner such that a wide range of sound pressures are encoded by small displacements of the inner hair cells. In this way, outer hair cells "fine-tune" frequency selectivity along the basilar membrane. Outer hair cells also receive efferent (descending) inhibitory innervation from the brain stem; this innervation attenuates and reduces the frequency selectivity of afferent signals. The interplay of afferent and efferent innervation acts as a feedback loop, allowing the brain to adjust hearing sensitivity on the basis of frequency and sound level. Outer hair cells are thus critical to the reception of quiet sounds, sound frequency discrimination, and perception of sound in noisy environments. Loss of outer hair cells leads to what is most commonly thought of as the noise-related damage that affects general "hearing ability," but can also manifest as hearing impairment beyond changes in audibility, e.g., reduced speech comprehension.

Hearing ability typically is assessed as changes in audiometric thresholds at various frequencies in clinical tests known as audiograms, which are indicative of hair-cell damage or loss. When hearing function tests appear normal, the conclusion is often made that no significant or permanent injury has occurred. However, recent research in animal models suggests that well before noise exposure causes hair-cell death, it can cause extensive injury to auditory nerve synapses that is not reflected in any appreciable audiogram changes. In animal experiments, researchers at the Massachusetts Eye and Ear Infirmary have shown that dramatic cochlear synaptopathy (loss of synaptic connections) can occur after exposure to continuous noise [6–9]. These studies indicate that as many as half of the auditory nerve fibers are lost after noise exposure at levels that cause neither permanent damage to hair cells nor permanent threshold elevations as measured by an audiogram. The initial synaptic loss can be observed within hours of noise exposure while subsequent death of the spiral ganglion (auditory nerve) cell bodies continues slowly over months or years. Despite histological degeneration of the auditory nerve, sensory function (as measured by an audiogram) remains normal. However, transfer of auditory information to the brain may be compromised, resulting in diminished perceptual processing. In particular, synaptopathy is thought to be associated with difficulty in understanding speech in noisy backgrounds. This difficulty occurs because the type of synapses that are most susceptible to damage encode suprathreshold sound information. Since an audiogram measures hearing thresholds, i.e., the softest sounds a person can hear, it cannot detect synaptic loss. It is not currently possible to noninvasively observe synaptic damage in living human subjects, so the prevalence and severity of this form of auditory damage in the population remain uncertain.

Noise Exposure

Hazardous noises arise from a wide variety of sources. In the military, some common sources of noise include aircraft, land vehicles, and naval vessels, as well as weapons fire and other explosive blasts. To understand noise characteristics and their potential effects on auditory health, it is useful to classify noise into two general categories as illustrated in Figure 2: continuous and impulsive. Continuous noise is relatively uniform over time, exhibiting only minor fluctuations in level or frequency content. Engine rooms and aircraft cockpits are examples of military work environments in which loud continuous noise is a concern. Impulse noise is characterized by a sharp burst of acoustic energy with rapid rise and decay times. Weapons fire and other explosions are examples of impulse noise. Heterogeneous combinations of both categories are known as complex noise.

Beyond categorizing the type of noise, alternative approaches have been used to quantify the noise to which an individual is exposed and to set guidelines for safe exposure. To those ends, noise dosimetry involves measuring sound-pressure levels in an environment to estimate an individual's exposure throughout a day, work shift, or event of interest. Noise dose typically is estimated in terms of acoustic energy in conjunction with the equal-energy hypothesis (EEH), which assumes that accumulated noise energy is sufficient to determine risk of NIHI without consideration of the underlying temporal or spectral characteristics. Under the EEH, two exposures are equivalent if the respective average noise levels and durations comply with a specified exchange rate. For example, a 3-decibel (dB) exchange rate often is employed such that a halving or doubling of the exposure time is accommodated with a +3 or −3 dB adjustment, respectively, to the allowable noise level.

To conserve individuals' hearing in industrial and military settings, regulatory agencies, such as the National Institute for Occupational Safety and Health,

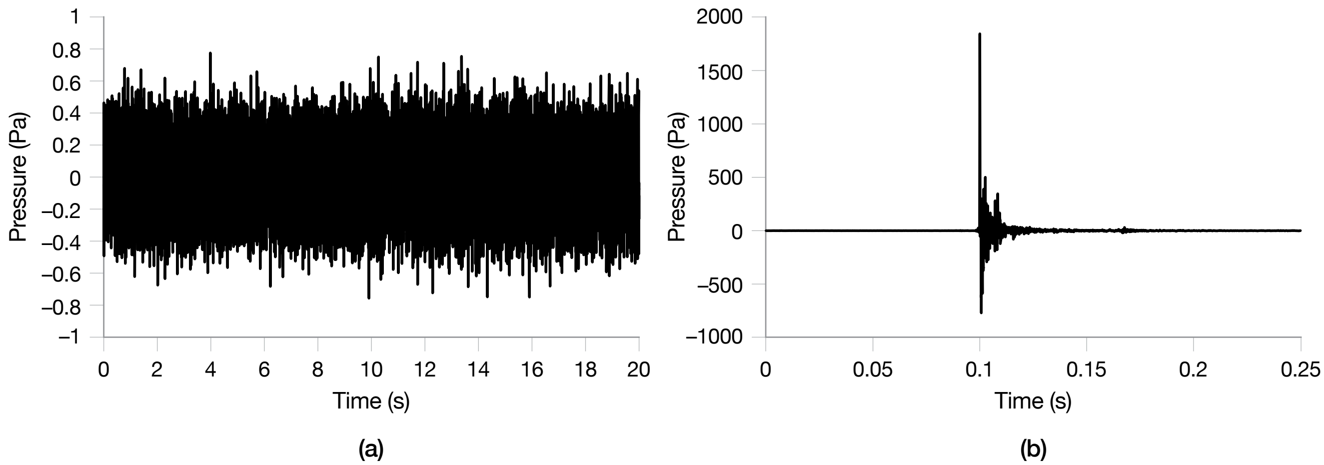


FIGURE 2. The waveforms depicted illustrate the relatively uniform level of continuous noise (a) and the short rise time, high peak level, and rapid decay of impulsive noise (b).

and military branches under the Department of Defense (DoD) Hearing Conservation Program have recommended guidelines on the maximum allowable daily noise exposure. For example, the current military standard sets a limit of 85 dBA[1] for a duration of eight hours for continuous noise exposure, where the exposure duration and level may be traded off to satisfy an equal-energy criterion using a 3 dB exchange rate.

Noise-exposure measurements can be compared against established limits or other criteria to determine a need for hearing protection or to predict the risk of hearing loss. Numerous damage-risk metrics have been proposed to quantify harmful aspects of noise exposure, but risk-assessment metrics remain an active area of research because no single metric is considered adequate across the spectrum of noise conditions. The most common damage-risk metric is a time-weighted average of the A-weighted noise level:

$$L_{\mathrm{Aeq,T}} = 10\log_{10}\left[\frac{1}{T}\int_T \frac{p_A^2(t)}{p_0^2}dt\right]$$

where T represents the exposure duration, p_0 = 20 μPa is the reference pressure level, and $p_A(t)$ is the A-weighted pressure-time waveform. As noted previously, the military noise exposure limit is 85 dBA for an eight-hour period, that is, $L_{\mathrm{Aeq,8h}}$ R 85 dBA [10].

While $L_{\mathrm{Aeq,8h}}$ has wide acceptance as a damage risk metric for continuous-noise exposure, many concerns have been raised that it is not adequate for predicting hearing damage from complex or impulsive noise [11]. One concern is that $L_{\mathrm{Aeq,8h}}$ and other energy-based metrics ignore much of the temporal and spectral structure of the noise; yet, evidence suggests that some of these features influence the damage severity from impulsive and complex noise. Hamernik et al. [12] showed that $L_{\mathrm{Aeq,T}}$ under-predicts hearing damage when continuous and impulsive noise are combined, and other studies suggest that impulsive exposures with predominantly low-frequency energy may be less hazardous than an equal-energy impulse dominated by higher frequencies [13]. Furthermore, the linear relationship between energy and permanent auditory threshold shifts only holds for noise levels up to about 140 dB [14]. Above this level, nonlinear operations may be necessary to translate the energy metric into auditory damage. In response to these concerns, several complementary or alternative metrics have been proposed for impulsive noise.

Recently, the military adopted the Auditory Hazard Assessment Algorithm for Humans (AHAAH) as part of the noise-limits design standard, MIL-STD-1474E [15]. The AHAAH electro-acoustic model developed by the U.S. Army takes an impulsive noise waveform as input and calculates an output value in auditory risk units (ARUs) that represents energy reaching the inner ear

[1] The A in dBA indicates that a filtering process meant to mimic the varying sensitivity of the human auditory system with respect to frequency, known as A-weighting, is used in the acoustic energy calculation.

[16, 17]. While AHAAH has been adopted in the standard for military acquisitions, the military acknowledges some limitations of this model, including the need for further validation. Zagadou et al. [18] recently suggested a number of modifications to the AHAAH parameters and hazard assessment calculation that yield a better fit to existing human blast exposure data. Another area of active research on AHAAH is evaluating the assumption that the middle-ear muscle contraction can provide significant protection against acoustic damage [19]. The so-called "warned" condition of the AHAAH assumes that this acoustic reflex is engaged prior to exposure and that the associated muscle contraction reduces energy transfer to the inner ear by as much as 20 dB, compared to the "unwarned" condition [20].

A limitation with the aforementioned metrics is that while $L_{\mathrm{Aeq,8h}}$ is intended for assessing continuous noise and AHAAH is for impulsive noise, no guidelines exist for combining the predicted risk of complex noise in which both continuous and impulsive noise pose a hazard. Several recent studies have sought to model auditory damage from complex noise exposures that may be more realistic to military and industrial settings. One concept to address the inaccuracy of the equal-energy hypothesis is a kurtosis correction factor for time-weighted average noise levels [21–23]; compared to the uncorrected, this concept has been shown to improve correlation against permanent auditory threshold shifts in chinchillas. Recently, Sun et al. [24] proposed an alternative kurtosis-based energy metric that adaptively elevates the effective energy in impulsive noise environments and reverts to the conventional A-weighted calculation in continuous noise environments. This approach is promising because it aims for a unified metric that appropriately adapts to the noise environment, but further study is needed to validate kurtosis-corrected energy metrics over more datasets, including those from complex military noise environments.

Noise Dosimetry for Military Environments

Noise dosimetry involves the collection of environmental noise data and the calculation of exposure levels with free-field, on-body, and/or in-ear devices. Free-field noise surveys typically characterize the noise levels of an environment, but accurately translating such a survey to the dose for an individual can be challenging. For example, sound-pressure levels at the ear drum can differ dramatically from those in an arbitrary free-field location (by 10 dB or more), depending on the exact positioning of body and ear relative to the noise sources [25]. Many modern, small-form-factor commercial off-the-shelf (COTS) dosimeters can be worn on the body (preferably in close proximity to the ear) to directly measure the dose in the vicinity of an individual, but they typically lack the dynamic and frequency ranges necessary for military use. Another potential measurement location for noise is in the ear canal. This measurement can be achieved by integrating a microphone into a hearing-protection device to allow for characterization of the noise exposure when hearing protection is worn. This configuration is referred to as an in-ear measurement, in contrast with the on-body measurement that may follow the individual but does not account for the noise attenuation of the hearing-protection device.

Hardware requirements for a dosimetry device vary for different noise types and environments. Military noise environments are complex, with both continuous and impulsive noise. The latter, for example from weapons fire, typically is the most demanding with respect to dosimeter design because of its highly dynamic nature and extreme levels. This challenging set of characteristics drives the need for a broadband dosimetry device with a high sampling rate and a wide dynamic range to avoid clipping or distortion from large blasts. Typical commercial noise dosimeters operate up to 140 dB peak sound-pressure level (SPL) and cover a frequency range similar to that of human hearing [26]. However, weapons fire, blasts, and other impact noises can exceed this SPL limit, and impulses can exhibit acoustic bandwidths extending well beyond the audio spectrum because of their short durations [27, 28].

Size, weight, and power are important considerations in designing a dosimetry device that is wearable and capable of measuring noise for a significant length of time (more than eight hours). For a small package suitable for an on-body or in-ear system, the trade-off typically will be between recording fidelity (driven by sample rate and bit depth) and recording duration (driven by battery life and memory capacity). While many commercial devices are available in a wearable form factor, they are designed for industrial noise environments and do not have adequate microphones or sampling rates to characterize impulsive noise with peak levels above 140 dB. Portable commercial audio recorders are one alternative that can be used to capture high-fidelity noise exposures with external microphones capable of measuring high-SPL noise; however, such devices are often bulky and have many settings and stressing power requirements to support the high-SPL external microphones. Furthermore, audio recorders do not support onboard processing to calculate noise metrics in real time. Smartphones have been successfully used to measure some noise environments [29] but are limited by a low sample rate and dynamic range. External microphones can be paired with a smartphone to increase the maximum SPL, but these require separate power and circuitry.

A comparison of some important characteristics for dosimetry devices is shown in Table 1. The COTS dosimeter column represents a state-of-the-art commercial dosimeter designed for industrial noise environments. The limitations on peak SPL and sample rate motivate alternative setups for exposure measurements in military environments. Two alternatives are explored in the middle columns and are described later in more detail. The final column represents a notional ideal device based, in part, on the impulse noise measurement requirements from the military specifications in MIL-STD-1474E. The ideal peak SPL of 175 dB is guided by the peak levels expected across a variety of military ordnance.

Lincoln Laboratory Dosimeter Prototyping

In response to the gap in available COTS devices to support noise measurements in military environments, Lincoln Laboratory began designing and prototyping a noise dosimeter in 2013 to support an operational noise collection on U.S. Marines in Afghanistan. The study was fielded by the Marine Expeditionary Rifle Squad (MERS) as part of a joint protocol with the U.S. Army Research Institute of Environmental Medicine.

Table 1. Performance Characteristics for Various Noise-Dosimetry Hardware Options

METRIC	COMMERCIAL OFF-THE-SHELF (COTS) DOSIMETER	AUGMENTED COTS RECORDER	LINCOLN LABORATORY SECOND-GENERATION DOSIMETER PROTOTYPE	IDEAL DEVICE FOR MILITARY DOSIMETRY
Peak sound-pressure level	140 dB	172 dB with external microphones	175 dB	175+ dB
Sample rate	20 kHz	96 kHz	128 kHz	192 kHz
Form factor	Clip to clothing	Small pouch	Small pouch	Clip to clothing
Maximum collection time	24+ hours	7 hours	8 hours	8+ hours
Measure in-ear and on-body noise	On-body only	Both with custom microphone housings	Both	Both
On-board processing	Yes, but limited metrics	No	Yes	Real-time calculations of noise exposure metrics

Excellent | Acceptable | Low-quality solution

The Laboratory's first-generation prototype had a primary objective of supporting the collection of data from small-arms fire through increasing the maximum SPL and sample rate. While this prototype was an improvement over the existing COTS dosimeters, the initial prototype fell short on some of the ideal requirements listed in Table 1. Building on the lessons learned from the first-generation system, Laboratory researchers developed a second-generation noise dosimeter designed to be more portable and capable of satisfying the instrumentation specifications of MIL-STD-1474E. Both generations of prototypes are shown in Figure 3. The second-generation prototype was funded jointly by MERS and the U.S. Army Combat Capabilities Development Command Soldier Center (formerly known as the U.S. Army Natick Soldier Research, Development, and Engineering Center). Several auxiliary sensors were integrated into the second-generation device to add a capability for simultaneously measuring other environmental data, including location (GPS), temperature, barometric pressure, and acceleration. The second design also included onboard data processing via a Xylinx Zynq, which has a field-programmable gate array and a dual-core ARM processor, as well as co-located accelerometers at each microphone to help detect and screen out microphone artifacts in a post-processing step.

The measurement quality of the second-generation dosimeter was verified by directly comparing the dosimeter to a reference laboratory-grade data-acquisition system (National Instruments, 24-bit, 200 kHz sample rate) for a series of high-SPL impulse-noise events. Figure 4 shows the test configuration and an example measurement that illustrates the close agreement between the prototype and the reference system for a 161 dB impulse generated from a compressed-air shock tube.

The peak SPL, $L_{\mathrm{eqA,100ms}}$ and median difference in the 1/3-octave-band levels of the prototype were within ±1.5 dB of the reference system for eight blast measurements with peak SPLs in the 160–179 dB range.

Augmented COTS Audio Recorder

During the development of the second-generation dosimeter prototype, several field collection opportunities arose that required an interim solution for measuring noise during military exercises. An augmented COTS recorder was constructed using a two-channel, 24-bit TASCAM DR100-MKIII recorder with a sampling rate of 96 kHz (Figure 5). With the addition of dual microphones and custom housings to support in-ear and on-body noise measurements, this device supports a peak SPL of 172 dB and approaches several of the ideal requirements of Table 1. The device is primarily limited by a lack of onboard processing and a bulky form factor. Laboratory tests were performed to validate the measurement quality through a comparison with the reference laboratory-grade National Instruments data-acquisition system.

To enable in-ear measurements with one of the COTS microphones connected to this augmented recorder, Lincoln Laboratory designed a custom, 3D-printed

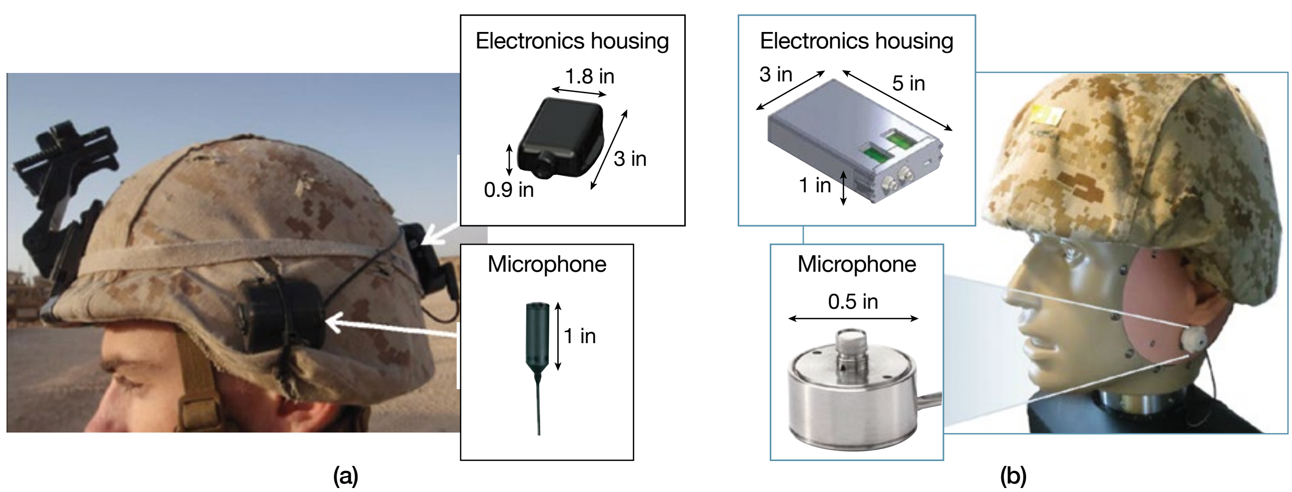
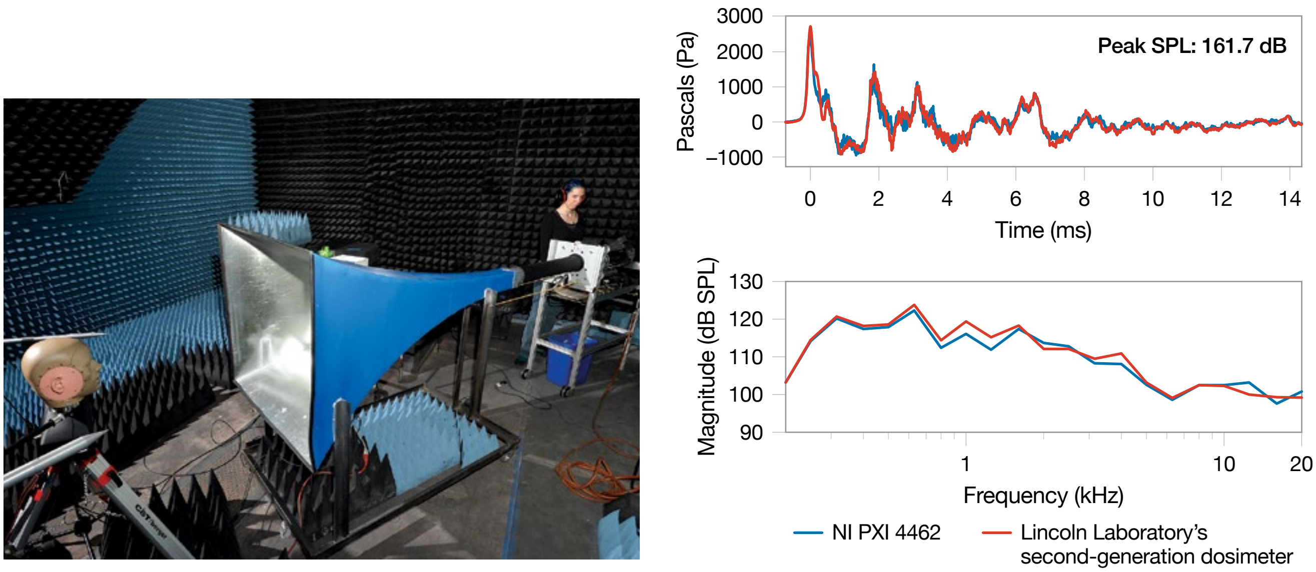


FIGURE 3. The first- (a) and second-generation (b) Lincoln Laboratory noise dosimeters are depicted.



(a) (b)

FIGURE 4. The shock tube measurement setup is seen in (a), and an example comparison of Lincoln Laboratory's second-generation dosimeter to the reference data-acquisition system measurements is depicted in (b). Note that the two systems have comparable performance in terms of peak amplitude and overall blast waveform shape (upper b) and blast spectral content (lower b).

housing that couples with a military-grade hearing protector ear-tip (Figure 6). Prior to test subjects' using this device as hearing protection, the impulse peak insertion loss was measured in accordance with the ANSI HPD testing standard [30], repeating the same shock tube and acoustic test fixture setup illustrated in Figure 4a.

The blast tests indicated that the in-ear microphones integrated with the foam ear-tip HPDs achieved an impulse peak insertion loss 45 dB or higher for blasts in the range of 148–172 dB. This level of suppression provides sufficient protection for many military exercises, including rifle training. Furthermore, the measurement quality of impulse noise captured by the augmented COTS recorder is very similar to the measurement quality of the reference laboratory-grade data acquisition system. Over a series of 18 blasts, a root-mean-square (RMS) error of 1.2 dB was found between the peaks measured by the augmented COTS recorder and those measured by the reference system.

One opportunity to collect and characterize military noise exposure occurred during Marine rifle training in which instructors are subjected daily to thousands of rounds of rifle fire. The augmented COTS recorder with both on-body and in-ear microphones was used to collect noise exposure measurements for seven rifle instructors over a two-day period at Marine Corps Base Camp Pendleton.

FIGURE 5. As an alternative to the second-generation dosimeter, Lincoln Laboratory augmented a commercial noise recorder (above) with dual microphones and custom housings so that it could be used during a military exercise.

Representative results of the daily noise exposure for two rifle range instructors are shown in Table 2. Several noise-exposure metrics were calculated on the basis of data from both the in-ear and on-body microphones. Subjects were asked to wear the in-ear microphone in place of their normal hearing protection during the

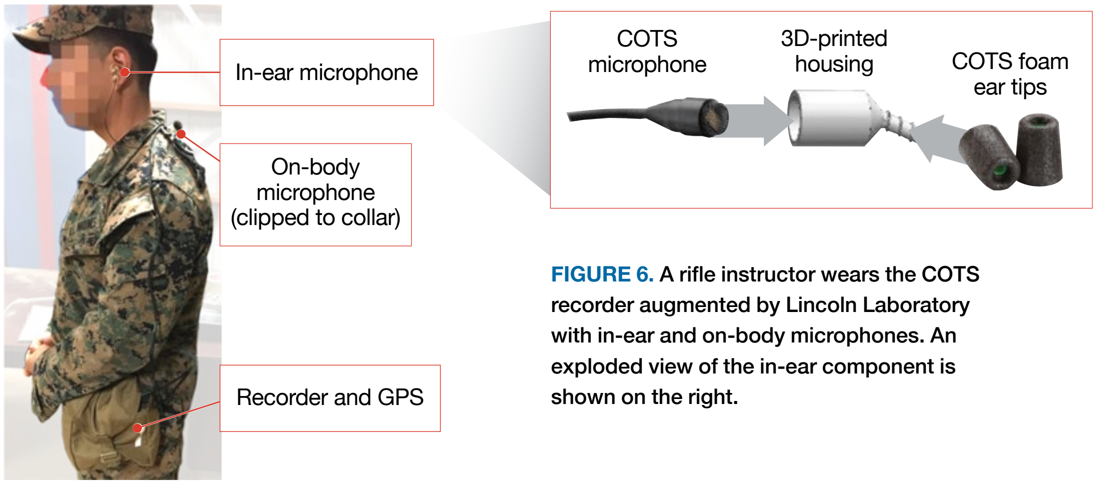


FIGURE 6. A rifle instructor wears the COTS recorder augmented by Lincoln Laboratory with in-ear and on-body microphones. An exploded view of the in-ear component is shown on the right.

collection. The on-body microphone measurements represent the potential noise exposure that individuals would have experienced had they not worn hearing protection; the in-ear measurements represent the actual noise that reached the ear canal with the HPD in place. Eight-hour equivalent energy and the corresponding dosage were computed along with the AHAAH ARU under the unwarned assumption. Without hearing protection, each instructor would be exposed to more than 1,000 impulses per day, each with peak SPL above 140 dB. The eight-hour equivalent noise energy on the rifle range without HPD is more than 15 dB above the DoD limit, corresponding to exposure dosages that are 30 to 45 times higher than the DoD limit. With hearing protection, the exposure metrics are considerably reduced. However, variability in the individual fit of the HPD can lead to dramatic differences in protection.

Of the two instructors represented in Table 2, Subject 5 achieved better overall suppression of the noise energy because of a better fit of the HPD, and he received a dose well below the limit. However, at a few points during the day, he briefly removed the HPD and, consequently, was exposed to a few rifle shots without hearing protection. This handful of unprotected impulses was enough to raise his AHAAH ARU count above the daily limit, despite his overall energy dosage being within the limit. In contrast, Subject 4 did not attain an equally good HPD fit, and as a result, the accumulated acoustic energy that reached his ear canal was more than twice the daily limit. However, this subject wore his HPD throughout the day, avoiding any unprotected impulse exposures. This consistent protection throughout the day held the AHAAH ARU at zero, despite the overall energy exceedance.

The on-body noise exposure measurements from the Marine rifle instructors indicated high-risk noise conditions for both the continuous and impulsive metrics. Wearing hearing protection reduced the exposure risk, but the level of protection can vary dramatically, depending on how well the HPD was inserted in the ear canal or whether the HPD was removed, even briefly, during the day.

Future Noise Dosimetry Considerations

Capturing both in-ear and on-body noise measurements is important for developing relevant noise-exposure models. In addition, collecting coordinated audiometric tests on warfighters during military operations or training could generate important datasets for evaluating existing noise metrics and validating new ones. To date, most military noise-exposure standards are typically validated on animal tests of blast overpressure exposures or a single human study of blast overpressure exposures conducted on more than 200 military volunteers in the early 1990s [31]. Additional dosimetry and audiometric collections during military training or operations could provide valuable data to help validate exposure metrics and standards over a wide variety of military noise conditions. In addition to improving noise-exposure standards, future data collections of this type may help to inform individual susceptibility for NIHI by including other physiological and genetic factors.

Table 2. Representative Metrics for Two Rifle-Range Instructors at Camp Pendleton

NOISE EXPOSURE METRIC	WITHOUT HEARING-PROTECTION DEVICES: SUBJECT 5	WITHOUT HEARING-PROTECTION DEVICES: SUBJECT 4	WITH HEARING-PROTECTION DEVICES: SUBJECT 5	WITH HEARING-PROTECTION DEVICES: SUBJECT 4	DEPARTMENT OF DEFENSE LIMITS
$L_{Aeq,8h}$	100 dBA	101 dBA	80 dBA	88 dBA	≤85 dBA
Dose	2995%	4479%	33%	219%	≤100%
Auditory Hazard Assessment Algorithm for Humans auditory risk units (ARU) (unwarned)	152,129	161,291	1004	0	≤200 ARU
Number of impulses ≥140 dB	1145	1393	6	0	n/a

DoD noise limits were exceeded (red: Without HPD Subject 5 and Subject 4 for $L_{Aeq,8h}$, Dose and ARU; With HPD Subject 4 for $L_{Aeq,8h}$ and Dose; With HPD Subject 5 for ARU)

Within DoD noise limits (green: With HPD Subject 5 for $L_{Aeq,8h}$ and Dose; With HPD Subject 4 for ARU)

Hearing Protection

Warfighters often do not use hearing protection provided to them for a variety of reasons—comfort, poor integration with other headgear, and compromised auditory situational awareness. Many of these concerns are echoed by participants in noisy, nonmilitary jobs [32], so a deeper understanding of HPD performance and limitations, with the ultimate goal of improving their usability and efficacy, is relevant to a wide swath of the population.

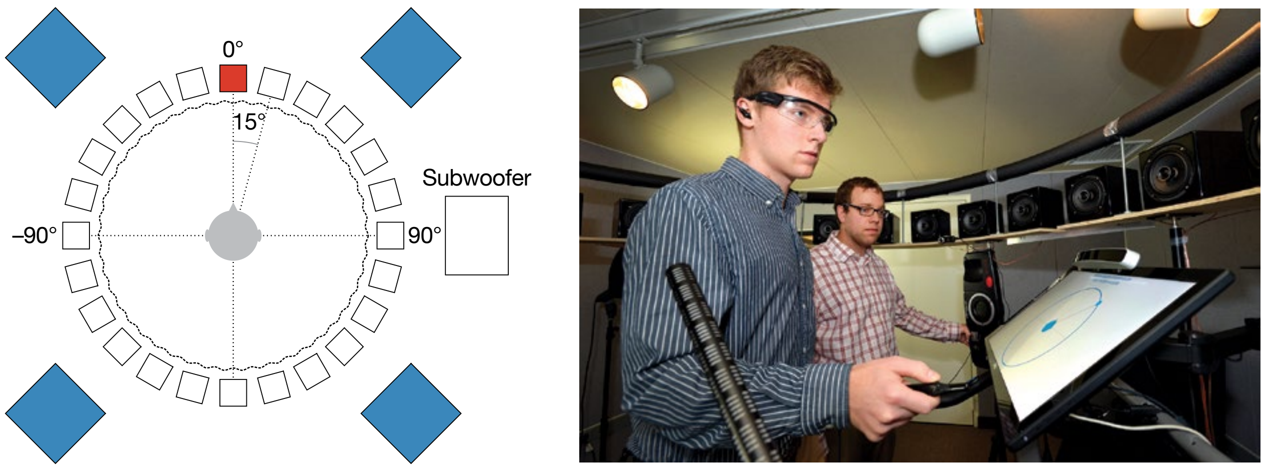
Studies of the effects of hearing protection on auditory situational awareness typically include some combination of tasks related to sound-localization performance, detection thresholds, and speech intelligibility [33, 34]. We are focusing on the first of these, along with a novel assessment of the additional cognitive load (listening effort) induced by wearing an HPD.

Devices and Subjects

Hearing-protection devices fall into one of two main categories: passive or active. Passive HPDs rely on occlusion of the ear canal, along with acoustic absorption and/or impedance mismatch, to prevent sound from impinging on the eardrum. The form factor may be over the ear, e.g., an earmuff, or in the ear, e.g., an ear plug. Active HPDs also rely on occlusion to limit the acoustic energy entering the auditory system, but in addition they contain an outward-facing microphone and an inward-facing loudspeaker, along with relevant signal processing algorithms, to enhance their performance. The most basic mode of an active HPD involves attenuating the signal at the microphone and passing it through to the loudspeaker otherwise unchanged. More sophisticated devices provide features such as nonlinear dynamic-range compression, active noise cancellation, and modes designed to address different noise types (e.g., impulse noise or continuous noise). Our HPD evaluation included two passive and three active devices, with the open ear as a reference.

Data-Collection Platform

Data were collected in a double-walled sound-isolating booth in which 24 loudspeakers were positioned in an eight-foot-diameter ring approximately five feet above the floor, as shown in Figure 7. The speakers were evenly spaced and pointed toward the central listening position; they were covered by an acoustically transparent cloth so their locations cannot be determined visually. Additionally,



(a) (b)

FIGURE 7. The experiment setup (a) includes 24 stimulus speakers evenly spaced in a ring with a diameter of eight feet. Larger speakers for background-noise output are shown in blue outside the ring, with a subwoofer shown on the right. The dashed line represents an acoustically transparent screen. The red-shaded speaker marks the direction directly in front of the listener, i.e., 0 degrees azimuth. The photograph in (b) shows the actual system, including the touch screen, with the acoustically transparent cloth removed to expose the loudspeakers.

four large speakers were placed in the corners of the booth to generate background noise if desired. Each test subject was seated in the center of the ring, positioned such that his/her ears were approximately at speaker level, with a 20-inch touchscreen positioned to display instructions and capture the subject's responses to tasks.

Sound Localization

The target stimulus used for the localization task was a recording of an AK-47 rifle cocking, an acoustic signal of approximately one second in duration with a broad spectrum that has been used in previous HPD localization studies [34]. The stimulus playback was calibrated at the listening position so that its level was matched across all 24 speakers and presented at 65 dBA SPL. Three background noise conditions were used for the sound localization task: quiet (<20 dBA ambient noise) and a Blackhawk helicopter noise calibrated at 60 dBA and at 80 dBA. These noise levels were chosen to span the range from "clearly audible" to "very difficult to localize" for the stimulus. The stimulus was played from one speaker at a time in random order, and each speaker was used twice for a total of 48 trials per background-noise level. The subject used a positional tracking device to point in the direction from which she/he heard the sound on each trial. The subject was shown the direction that she/he was pointing on a computer screen in real time to prevent errors in the pointing direction caused by any obstruction of the sensor by the body. Localization performance was assessed by measuring the mean angle error (MAE) between the target speaker and the subject's response in degrees azimuth. Quadrant errors, which manifest as front/back or left/right confusions, also were quantified.

Figure 8 shows the MAE in localization for the open ear and five hearing-protection devices across all subjects for each of the three background-noise conditions. The mean error for each subject was computed across the 48 trials per noise condition and HPD. In addition to the degradation of localization accuracy, quadrant errors also were evaluated as shown in Figure 9. The space around the listener is divided into four quadrants defined as: right-front (0° R azim. < 90°); right-rear (90° R azim. < 180°); left-rear (180° R azim. < 270°); left-front (270° R azim. < 360°). Quadrant errors occur when a subject's localization estimate is in a different quadrant from the actual stimulus loudspeaker (typically a front/back error rather than a left/right error), and the difference between the estimated and actual azimuth is greater than 30 degrees.[2] A

[2] The 30-degree minimum constraint was imposed to differentiate actual quadrant errors from localization blur near the quadrant boundaries.

two-way analysis of variance (ANOVA) performed on the mean localization MAE revealed that there was an effect of HPD ($F (4, 47) = 45.0$, $p < 0.001$), noise level ($F (2, 23) = 104.5$, $p < 0.001$), and a significant interaction ($F (8, 95) = 2.189$, $p < 0.035$) between the two. Once quadrant errors were removed, the interaction effect was no longer significant ($F (8, 95) = 0.315$, $p = 0.95$). A comparison of Figure 8 and Figure 9 suggests that the dominant cause of degraded localization performance with an HPD is the prevalence of quadrant errors. The implications of localization errors in terms of warfighter performance are an active area of research.

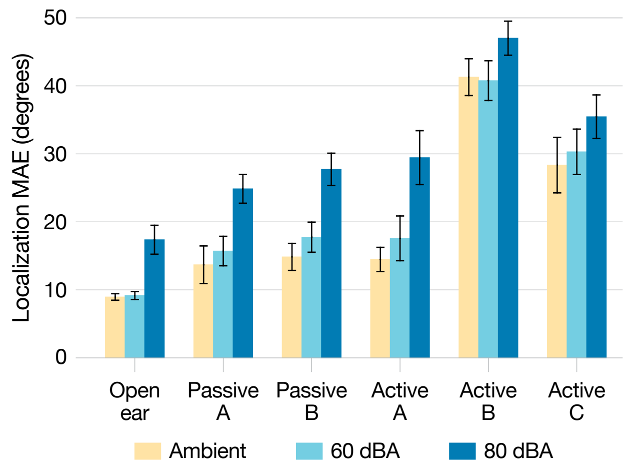


FIGURE 8. The localization performance in terms of mean angle error (MAE) is given for the three noise conditions (ambient, or quiet; helicopter noise at 60 dBA and at 80 dBA) for the open-ear condition and for five hearing protection devices (two passive HPDs and three active HPDs). Performance with all devices degrades with increasing background noise. The large errors induced by the Active B and C devices, caused mainly by quadrant errors as shown in Figure 9 below, are particularly detrimental to auditory situational awareness.

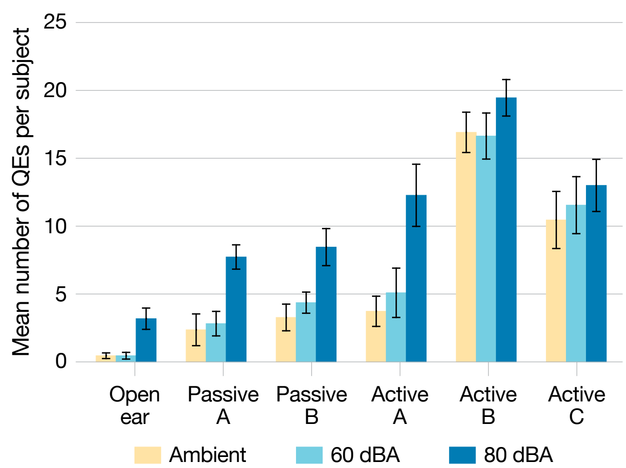


FIGURE 9. Localization performance in terms of quadrant errors (QEs) is given for the three noise conditions (ambient, or quiet; helicopter noise at 60 dBA and at 80 dBA) for the open-ear condition and five hearing protectors (two passive HPDs and three active HPDs).

Cognitive Load

A dual-task paradigm was employed to evaluate the effect of the HPDs on cognitive load. The primary task involved the evaluation of speech intelligibility in 75 dBA noise by using the modified rhyme test (MRT), a six-alternative forced-choice task for which each trial consists of the carrier phrase “Please select the word” followed by a target word [35]. The target word, played from the speaker directly in front of the subject, rhymed (i.e., either the first or last consonant varied) with a number of word selections presented on the touch-screen response pad. The subject had three seconds to select the word he/she heard before the next trial began. The secondary task required the subjects to remember the five previously selected words in the MRT. In addition, visual reaction times in response to the lighting of a red LED light above the touch screen were measured throughout the experiment. Visual reaction time has previously been shown to be sensitive to mental effort [36] and has been established as a metric for predicting cognitive load.

To control for variations in individual reaction time and alertness over the multiple days of testing, reaction times were acquired without the MRT or word-recall task. Noise and stimulus levels were kept in the same configuration as during the dual-task testing, but subjects were instructed to ignore the speech and noise, and only respond to the visual stimulus as quickly as possible. Dual-task MRT blocks were alternated with reaction-time-only blocks in an A-B-A-B fashion. The median reaction time over both reaction-time-only blocks was taken as the single-task visual reaction-time baseline for each hearing-protector condition. Reaction time and word recall were initiated at random intervals (every 7 to 12 trials) following an MRT trial so as to not overlap directly with the primary task.

Dual-task reaction times were better on average for the open ear than for HPD conditions, and word recall was the highest on average for the open-ear condition

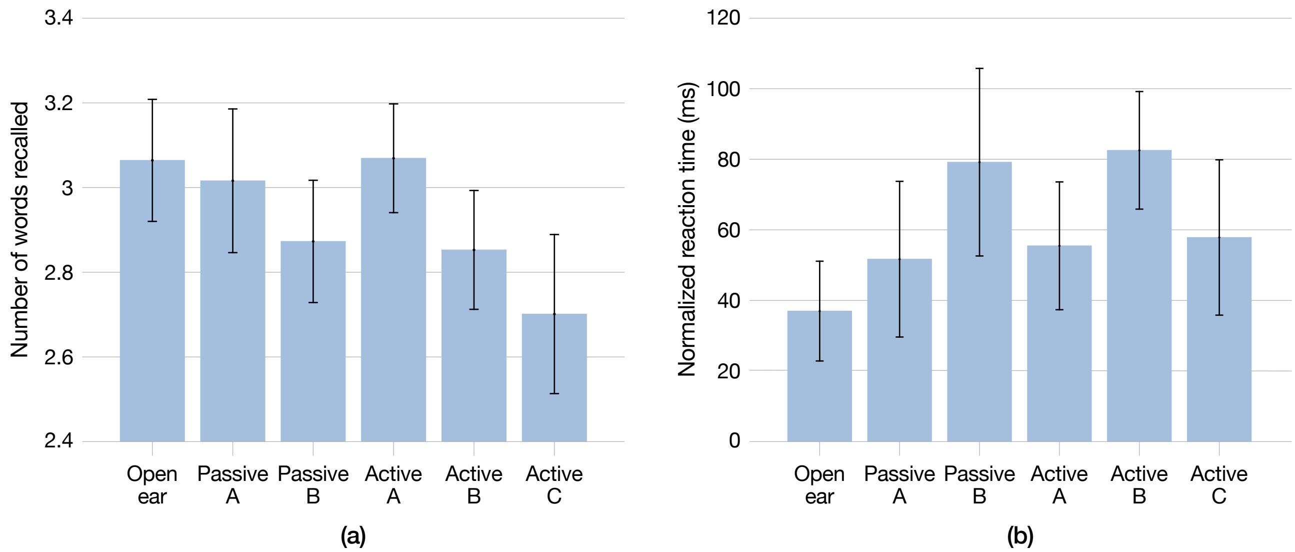


FIGURE 10. The graphs depict the measures of cognitive load induced by the evaluated hearing protection devices (HPDs). Word recall is seen in (a), and visual reaction time is plotted in (b).

(Figure 10a). A two-way ANOVA (with HPD as the fixed factor and subjects as a random factor) was performed on the mean reaction times after subtracting the median baseline time from the same session. This analysis revealed that there was an effect of HPD ($F(4,48) = 3.76$, $p < 0.038$). Post hoc Tukey multiple comparisons ($\alpha = 0.05$) revealed a significant difference between the open condition and the Active B HPD. These results suggest that additional effort required to process speech while wearing hearing protection may have an effect on cognitive resources required to execute the visual reaction-time task.

A separate two-way ANOVA performed on the mean number of words recalled (Figure 10b) revealed that there is no main effect of HPD ($F(4, 45) = 1.65$, $p < 0.178$). However, there is a weak negative correlation ($R = -0.26$, $p < 0.0725$) observed between the reaction time and the number of words recalled across all subjects and hearing-protection devices. This trend indicates that increased visual reaction time and reduced cognitive-load task performance are related, as both are indications of fewer available processing resources.

Computational Models of Auditory Situational Awareness

Computational models of the auditory pathway have been in use since at least the early 1980s and provide a useful tool for testing our understanding of how the auditory system functions and how damage to the cochlea affects both auditory system function and consequently an individual's perception and situational awareness. Functional models based on the physiology or mechanics of the ear, such as the AHAAH model [37], have been adopted by the U.S. Army in MIL-STD-1474E for evaluating impulse-noise exposure and predicting temporary threshold shifts. The models typically share common components: a linear filter for the transfer function of the outer ear and a bandpass filter bank that represents the frequency sensitivity of the cochlea. This filter bank has a number of nonlinear properties that affect the filter width and gain, depending on the acoustic stimulus input level and frequency. While the AHAAH model is designed to predict the risk of hearing damage, Lincoln Laboratory has developed models to predict the effect of hearing damage on auditory performance. A model schematic, which includes a neural component, is shown in Figure 11, where the output is either a measure of speech intelligibility (percentage of words correctly identified) or sound localization (angle error).

Modeling Sound Localization

Because auditory localization performance was effective at discriminating HPDs in the analysis described earlier, a subsequent modeling effort focused on reproducing the human-subject data with a computational localization model. The purpose of the model is to speed up HPD evaluations, reducing the need for human subject testing and minimizing the time between design cycles. To simulate the original protocol for the model-based analysis, the human listeners were replaced with a GRAS 45CB Acoustic Test Fixture (ATF), a humanoid head-and-shoulders manikin with integrated in-ear microphones (Figure 12). The ATF was fitted with each HPD, and data were collected by playing the AK-47 stimulus twice from each loudspeaker and recording binaural signals with the in-ear microphones. For the evaluation, these binaural signals were used as input to an offline localization algorithm that employed an auditory model to generate angle estimates for each source location. The performance of such a model can be quantified in terms of various error metrics, e.g., the number or percent of front/back confusions, and compared to those same metrics computed on the human-subject data to assess the accuracy of the model-based HPD evaluations.

Once data were collected with the ATF, localization modeling was done in two steps. First, a frontal lateralization angle between −90 degrees and 90 degrees was computed using the approach described by May et al. [38].[3] Second, the localization process was augmented with a front/back disambiguation step to extend the estimate range to cover the full circle around the listener, i.e., −180 degrees to 180 degrees. Interaural level differences (ILDs, i.e., differences in loudness and frequency distribution between the two ears) are distinct, particularly at high frequencies, for mirrored front-to-back source locations, and are due mainly to direction-dependent shadowing and reflection effects from the pinnae (part of the ear outside the head).

In a preprocessing step, we generated the expected ILDs from an acoustic source located in a series of positions with fine angular resolution around the ATF. Then, during the estimation process, our algorithm followed these steps when provided with binaural signals from an unknown source location:

1. Compute a frontal azimuth estimate θ in the range (−90 degrees to 90 degrees).
2. Compute the ILD spectrum for the input.

[3] Localization estimates often are restricted to the frontal hemisphere because some relevant cues, such as interaural time differences, have front/back ambiguities caused by the use of only two ears as sensors.

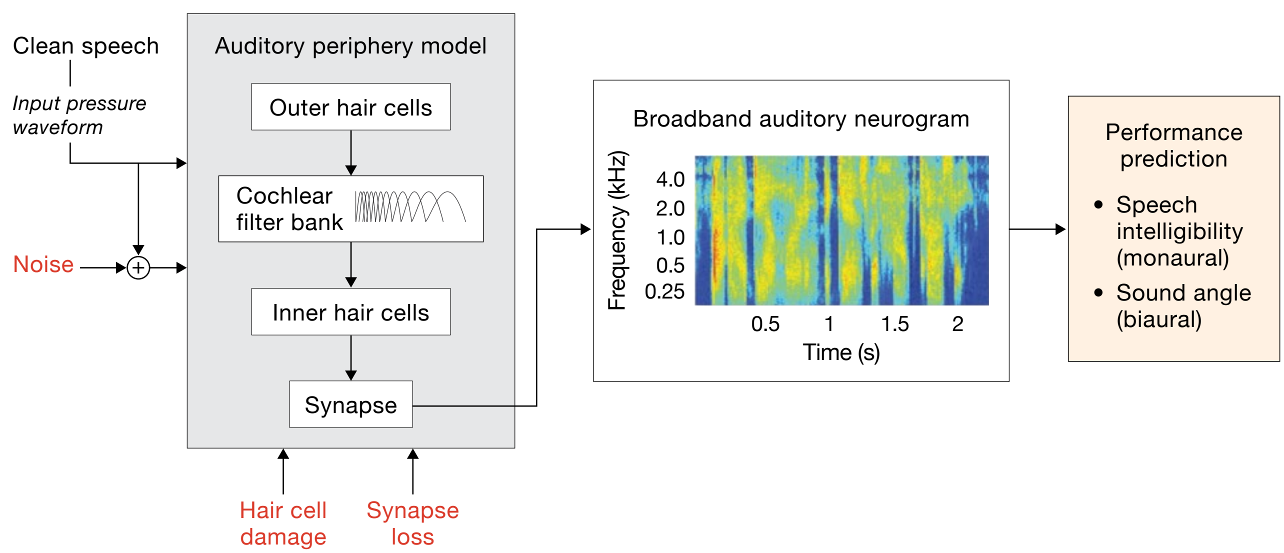


FIGURE 11. The schematic diagram depicts a model to predict speech intelligibility or sound localization performance. The input to the model is a pressure waveform. First the signal is passed to a cochlear periphery model, which is composed of inner and outer hair cells (IHC, OHC) and synapses to generate the auditory nerve output. By simulating the model at various frequencies along the filter bank, a spectrogram-like auditory neurogram output can be generated. Finally, these neurograms can be compared to a reference to estimate intelligibility or across the two ears to estimate the angle of arrival. Other features of the model, shown in red, include the ability to degrade the input with background noise or to simulate hearing loss.

3. Use the azimuth to look up the expected ILD spectra for the corresponding frontal (θ) and rear (180 – θ degrees) positions.
4. Compare the measured ILD spectrum from step 2 to the expected ILD spectra from step 3 and compute the RMS error for each of the two comparisons.
5. Choose the final output azimuth θ (source in front) if the RMS error between the measured ILD spectrum and the expected frontal ILD spectrum is the smaller of the two errors. Otherwise, choose the final output azimuth, 180 degrees to θ (source in back).

Data related to two example angle estimates are shown in Figure 13. In Figure 13a, the measured ILD spectrum closely matches the expected ILD spectrum for a source in the front at 35 degrees azimuth, so the algorithm correctly chooses the frontal location. In Figure 13b, the ILD spectrum is distorted by a hearing protector mounted on the ATF, causing an incorrect choice of the rear position, i.e., a front/back quadrant error similar to one a human listener might make. Preliminary localization modeling results are shown in Figure 14 (the corresponding human subject results are shown in Figure 9). The modeling approach tends to overestimate the number of quadrant errors relative to the number that the human subjects made, but the performance trend among the HPDs is captured.

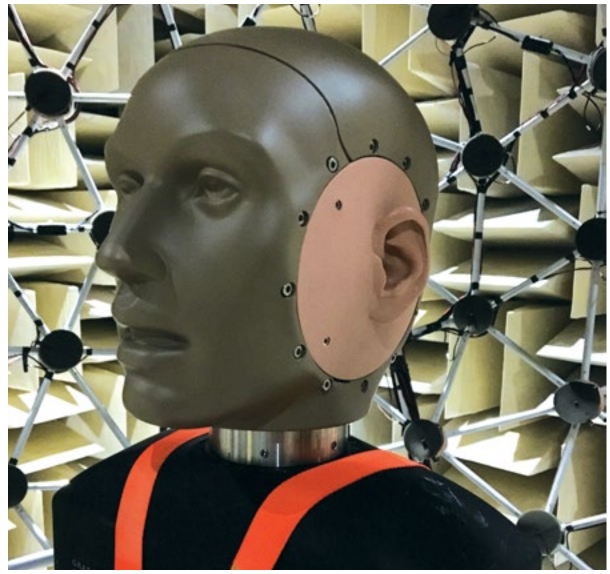


FIGURE 12. The GRAS 45CB Acoustic Test Fixture was used for the model-based hearing-protection device localization assessment.

Modeling Speech Intelligibility

Previously mentioned in this article is the recent discovery that noise exposure can cause a permanent loss of cochlear synapses that is undetectable by a traditional audiogram [6]. The type of synapses that are targeted are those that encode high-level sounds and are associated with low spontaneous rate (LSR) auditory nerve

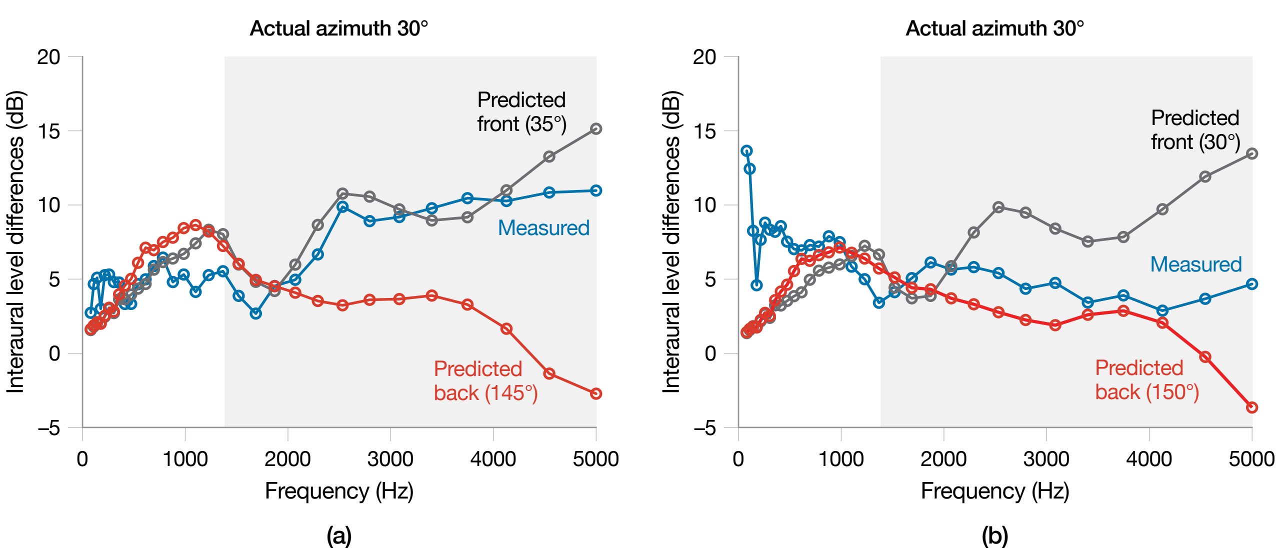


FIGURE 13. The graphs show examples of front/back disambiguation using measured and predicted interaural level differences (ILDs). The data in (a) are for an open-ear test; in (b) are results for an active hearing-protection device (HPD). Note the distortion (blue line in [b]) induced by the HPD.

fibers. This phenomenon is believed to create difficulties for humans to understand speech in noise, although no established clinical or noninvasive technique verifies this assumption. In addition, it is not well understood how LSR loss might interact with the medial olivocochlear reflex (MOCR), a feedback mechanism that controls outer-hair-cell gain. The MOCR is thought to provide protection and signal enhancement for a listener who is in background noise, and it is also used as a predictor of susceptibility to noise injury [39]. To ultimately make predictions of auditory performance, Lincoln Laboratory utilized a computational model of the auditory periphery and cortex to study the effect of LSR auditory-nerve fiber loss and MOCR strength reduction on the cortical representation of speech intelligibility in noise.

We used the auditory-periphery model of Zilany et al. [40, 41] and Smalt et al. [42] to make predictions of auditory-nerve responses to speech stimuli in noise. One hundred auditory-nerve fibers were simulated for each of 32 frequencies along the cochlea between 100 Hz and 8 kHz. The resulting cochlear neurogram, a spectrogram-like output based on auditory-nerve fiber outputs, then became the foundation for measuring the fidelity of speech encoding. At the level of the auditory nerve, this measurement is achieved by computing the Neural Similarity Index (NSIM, [43]), and at the level of the auditory cortex by computing the Spectro-Temporal Modulation Index (STMI, [44]). Both of these techniques essentially compare two copies of the model output: the first is a clean copy of the acoustic target, and the second can be degraded by noise, hearing loss, or both. The strength of the MOCR and the percentage of lost LSR auditory-nerve fibers (i.e., the degree of synaptopathy) can be adjusted in the model, in which the effect on predicted speech intelligibility can be observed. Both STMI and NSIM are normalized measures, where 0 indicates no intelligibility and 1 indicates perfect intelligibility.

Simulations of speech intelligibility using the neurogram revealed that both the MOCR and LSR loss have minimal effect on STMI speech intelligibility predictions in quiet (ΔSTMI < 0.04). In background noise, however, the effect of the MOCR and LSR loss was more significant ($\Delta STMI_{MOCR} = 0.23$, $\Delta STMI_{LSR} = 0.07$). This effect can also be observed qualitatively in the neurogram images (Figure 15). In these simulations, a complete loss of the LSR fibers in the population response was used to

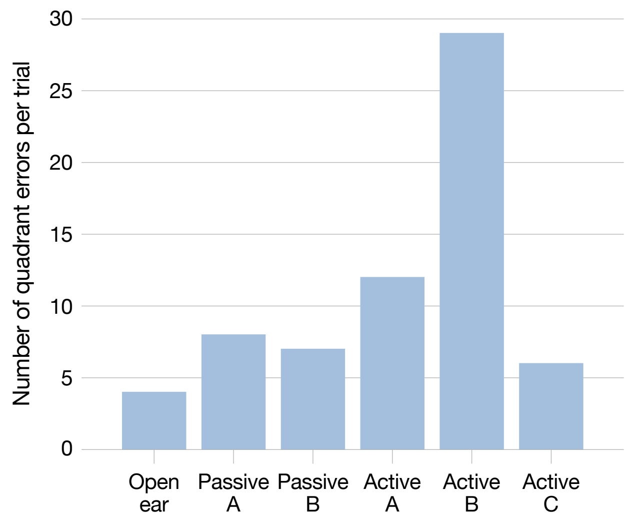


FIGURE 14. Preliminary results for tests using a localization model are similar to those derived from human-subject testing (Figure 9), as indicated by the average number of quadrant errors (QEs).

indicate the maximum possible damage that could occur. Preliminary results on a 50-subject human test showed a better correlation with speech intelligibility in noise using STMI and MOCR ($r = 0.38$) as compared to the audiogram alone ($R = 0.3$).

These results suggest that while LSR loss can have some impact directly on neural signal fidelity in noisy environments, efferent feedback such as the MOCR has a stronger effect on speech perception. One possible explanation for this observation is that the LSR fibers are indirectly responsible for reducing MOCR strength, which in turn causes communication difficulties. This hypothesis is supported by the fact that LSR fibers typically encode louder sounds, and it is known that they drive the input to the middle-ear reflex. Further simulations using the auditory-nerve model may help researchers understand human susceptibility to noise-induced hearing loss, the impact of damage to low spontaneous rate auditory-nerve fibers, and the health and performance risks of noise exposure to the warfighter.

Discussion

Implications of Compromised Hearing

One outstanding question related to auditory health and situational awareness is how degradations caused by hearing loss or hearing protection affect the warfighter's

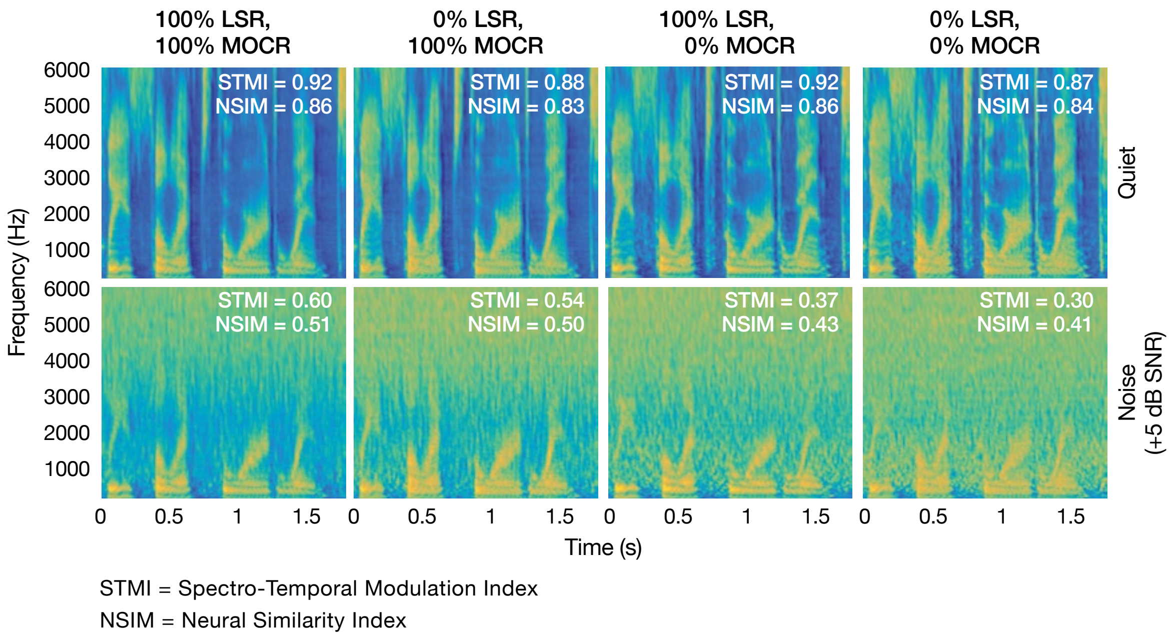


FIGURE 15. The neurograms of speech in quiet and in broadband background noise show results for a healthy cochlea, 100 percent low spontaneous rate (LSR) auditory nerve fibers, and 100 percent medial olivocochlear reflex (MOCR) strength. The signal-to-noise ratio of the neurogram is reduced significantly because of synaptic loss (LSR and efferent feedback loss [MOCR]) for speech in background noise but not in quiet. This simulation may explain why the predominant hearing issue people have is listening in background noise; it is likely to be a result of synaptic loss in the cochlear nerve from noise exposure.

operational performance. Put another way, at what point does a change in detection threshold, or localization accuracy, or speech intelligibility, or cognitive load, or some combination of these, become a hindrance to mission success or survivability? This complex issue has been studied only to a limited degree thus far in laboratory and field experiments, and through surveys of active-duty soldiers.

Field Studies

Casali et al. [45] studied the effects of three HPDs on in-field performance of reconnaissance and raid missions through subjective and objective measures of auditory detection and identification of threats and through communication with participants in the experiments. Casali et al. found varied performance over missions and ratings and no indication of an optimal HPD among the ones included in their experiments. While they did receive subjective feedback from participants about aspects of situational awareness, such as localization abilities, they did not provide measurements of HPD performance.

Clasing and Casali [2] studied detection and identification of auditory threats (gunshot, spoken Arabic, and weapons preparation) with five HPDs and open ear; they found performance varied among the HPDs, with occasional performance improvements (from one HPD) but mainly detriments. Again, no measurements of HPD performance were provided.

Talcott et al. [46] studied azimuthal localization of and response time to gunshots with four HPDs and the open ear. They found that localization was generally worse with all HPDs (none preserved "normal" performance), but they did not comment on the impact any of the performance degradations might have.

Sheffield et al. [3] used hearing-loss simulation systems to study the effect of hearing loss on operational performance in a dismounted combat scenario (a last-man-standing paintball competition). While their results indicated a reduction in offensive effectiveness with increased hearing impairment but little effect on survivability, they did not map their simulated hearing profiles to metrics such as localization or intelligibility.

Laboratory Studies

Peters and Garinther [47] studied the effects of degraded speech intelligibility on the performance of tank operators in a simulator. Speech intelligibility was modulated by electronically distorting the speech in the simulator, and the performance of two-man crews was quantified in terms of mission time, completion, and error, as well as gunner accuracy. All performance metrics other than accuracy were shown to degrade with reduced intelligibility.

Similarly, Mentel et al. [48] evaluated the relationship between speech intelligibility and operational performance in a simulation of an Aegis Combat System Command Information Center. Using a modified rhyme test, they found performance, in terms of the percentage of successfully accomplished tasks, decreased significantly when modified rhyme test scores fell below 65 percent correct. Neither of these studies involved HPDs as a variable, although our results suggest that noise level is more important than HPD choice when assessing speech intelligibility.

Surveys

By surveying military personnel with infantry or combat-support roles with a questionnaire, Semeraro et al. [49] identified nine mission-critical auditory tasks (seven communication, one detection, and one localization) to be included in future tests of auditory fitness. However, while they asked the participants to comment on the consequences of poor performance for each task, they neither inquired about nor tried to define acceptable performance levels.

Auditory Fitness for Duty

All of these studies can be classified as attempts to evaluate what is known as auditory fitness for duty, or "the possession of hearing abilities sufficient for safe and effective job performance" [50]. Whether auditory performance is limited by hearing impairment or the use of HPDs, the effects of such limitations should be linked to job performance more directly. Such study of this linkage will help determine the suitability of people with or without hearing protection or enhancement for the tasks to which they are assigned. Efforts to help define tests for auditory fitness for duty have been discussed for military [49, 51] and law-enforcement [52] personnel, but significant work remains to be done in this area to understand the relative contributions of detection, localization, speech intelligibility, cognitive load, and other factors and to define appropriate testing protocols.

Continuing Research

Auditory health is a significant concern within the DoD and VA because typical high-noise military environments, whether associated with training or operational activities, are responsible for hearing loss and tinnitus affecting large numbers of active-duty service personnel and veterans. Reductions in hearing ability are known to affect the operational performance of a soldier, both on and off the battlefield, in ways that must be better understood to determine the fitness of a person for a specific job and to develop devices that provide ample hearing protection. Lincoln Laboratory researchers are involved in research to improve dosimetry devices that address the complexity of military noise environments comprising continuous and impulse noise, often with moving sound sources and listeners. We have investigated damage-risk metrics needed to estimate the likelihood and severity of noise-induced hearing injuries, and we have developed computational models of the auditory pathway to evaluate detrimental HPD-induced effects on auditory perceptual tasks, such as localizing sounds and understanding speech. In the future, we will continue to seek advancements for evaluating and maintaining auditory health that will lead to improved warfighter performance and reduced hearing-related disabilities. ■

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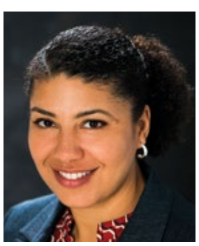

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