Physiological Insights into Hearing Loss and Optimal Amplification

By Ian C. Bruce, Ph.D.

Since the introduction of electric hearing aids over 100 years ago, the search has been on for a way to fully compensate for all aspects of hearing impairment by hearing aid amplification. To date, this goal has not been achieved. However, as hearing aid technology has developed from the era of carbon aids to the present-day digital signal processors, much progress has been made in making optimal use of the available technology. The vast majority of improvements in amplification strategies have been achieved by trial and error using empirical testing; the physiological basis for optimal hearing aid amplification is generally not well understood.

Consider the development of prescriptions for linear amplification. Early studies suggested "mirroring of the audiogram," such that 1 dB of gain is applied for each 1 dB of hearing loss at each frequency. While this appeared to be satisfactory for conductive hearing losses, it was found to amplify loud sounds excessively for sensorineural losses. Consequently, it was proposed that a constant could be subtracted from the gain to bring it down to an acceptable level. Watson and Knudsen (1940) refined this idea by suggesting that the gain should be reduced additionally by a factor depending on most comfortable equal loudness curves. Shortly thereafter, Lybarger (1944) suggested the alternative "half gain rule" that formed the basis for many subsequent linear amplification prescription schemes.

The goal of these linear prescriptions is to optimise audibility, comfort, and speech intelligibility; the differences between the various prescription schemes reflect different weightings of the importance of these desired perceptual outcomes. But can we determine the physiological basis for a linear gain prescription? Bondy et al. (2004) used a computational model of the auditory periphery to determine the linear gain-frequency responses that would optimally restore *normal average levels of auditory nerve activity* for a range of audiograms. The model predictions very closely match the NAL-R prescription (Byrne and Dillon, 1986). This result indicates that in applying amplification that gives more normal auditory nerve activity *on average*, an optimal mix of audibility, comfort, and speech intelligibility is obtained for speech *on average*.

We then attempted to extend these results to multiband compression on a phoneme-by-phoneme basis, rather than just for speech on average. It is known from empirical studies that by adjusting the compression characteristics, it is possible to avoid distorted and uncomfortably loud signals, to reduce the intensity differences between phonemes or syllables, to provide automatic volume control, to increase sound comfort, to normalize loudness, to maximize intelligibility, or to reduce background noise (Dillon, 2001). However, the required compression parameters vary substantially among these goals; consequently, any one compression scheme tends to provide benefit in some but not all aspects of compensating for hearing impairment. Consistent with the empirical observations, using the auditory periphery model it was *not* possible to find one set of compression parameters that normalized the auditory

Figure 1.

Schematic of "neurocompensator" hearing aid training.

The block "**H**" indicates a model of normal hearing, while the block "**Ĥ**" indicates a model of impaired hearing.

The "**Nc**" block is a trainable hearing aid algorithm or neurocompensator.

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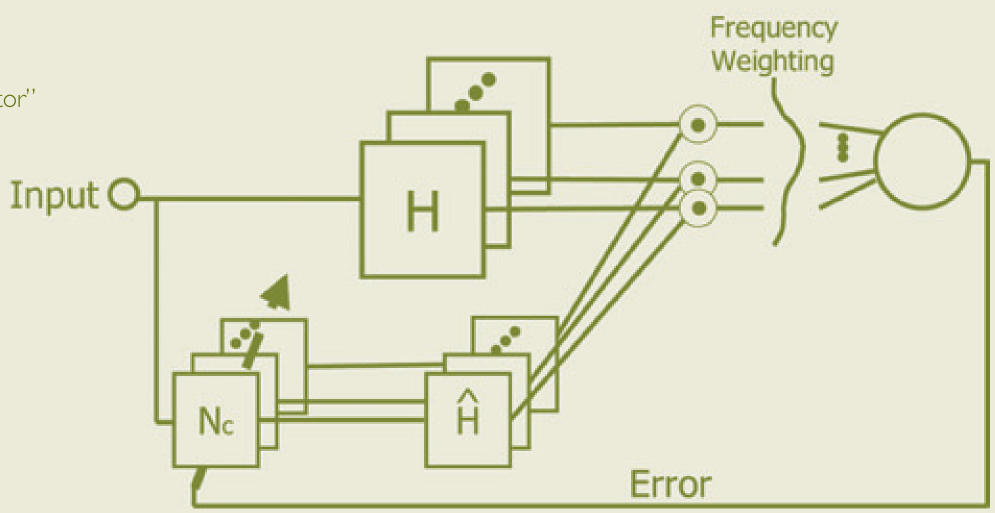

nerve response to all phonemes. Even so, the model does predict that the *maximum gain* over all phonemes should be subject to a compression ratio of around 1:1 in regions of no hearing loss up to 2:1 in regions with substantial loss—ratios that are fairly consistent with most nonlinear prescriptions for wide dynamic range compression.

The failure thus far to find, either empirically or with the auditory model, an optimal set of parameters for a simple compression scheme indicates that more sophisticated nonlinear amplification is required. One group has developed just such an algorithm, referred to as adaptive dynamic range optimization (ADRO; Blamey, 2005). ADRO is a *slow-acting* automatic gain control that uses sophisticated logic to determine the optimal gain for a specific acoustic environment. However, the normal cochlea makes use of fast-acting nonlinearities such as amplitude compression, spectral suppression, and temporal adaptation. Loss of these nonlinearities in the impaired cochlea contributes to distortion in the neural representation of speech (Bondy et al., 2003; Bruce et al., 2003; Sachs et al., 2002). What forms of amplification could compensate for such distortions?

A number of groups have investigated spectral sharpening schemes aimed at counteracting the broadened tuning and loss of spectral suppression that comes with cochlear hearing loss. Unfortunately, the improvement in speech intelligibility obtained with these algorithms has proven to be very little or none. Consistent with these empirical results, it appears from animal and modelling studies (Bruce et al., 2003; Sachs et al., 2002; Giguère and Smoorenburg, 1998) that the broadening of tuning and loss of suppression is so substantial that sharpening of spectral peaks in speech may have little effect on the neural representation. In contrast, animal and modelling studies suggest that amplification to adjust the *relative amplitudes* of spectral peaks may help overcome broadened tuning and loss of suppression (Bruce, 2004; Bruce et al., 2003; Sachs et al., 2002). This scheme, referred to as contrast enhancing frequency shaping (CEFS), has also been shown to be compatible with multiband compression (Bruce, 2004), unlike spectral sharpening schemes.

However, the question still remains: what is the optimal amplification strategy to restore the normal neural representation of speech? Chabries et al. (1995) developed a simplified auditory model that could be inverted, such that it could be included directly in an amplification algorithm to compensate for the modelled hearing loss. Unfortunately, in order to make the model invertible, important features such as spectral suppression and temporal adaptation could not be included. An alternative approach to finding an optimal amplification strategy is to use machine learning algorithms, in which an amplification scheme is trained so that it minimizes the difference between the output of a normal model in response to unprocessed speech and the output of an impaired model in response to speech processed by the hearing aid (Kates, 1993; Bondy et al., 2004; Chen et al., 2005). One such scheme, illustrated in Figure 1, makes use of a neural network in the amplification block, and consequently is referred to as a "neurocompensator."

Preliminary results with these trainable amplification schemes are promising, but the search continues to find the best metric for measuring distortion of the neural representation of speech and the best amplification framework for the algorithm to optimize. When the answers to these questions are discovered, we may finally be sure that we have the optimal amplification strategy to compensate for cochlea hearing loss.

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