Compensation for Sensorineural Hearing Loss with Adaptive Control Methods

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***Abstract*—Many methods used to identify the particular ear of a person require a degree of speculation and approximation through examination. A feedback control system could be used in particular hearing aid devices, but also has potential to be applied in neurological systems that could interface with neural connections between the inner ear and the brain. This work introduces a novel approach for hearing assistance with accurate system identification based on monitoring of hearing response and adaptive compensation. Specifically, model-reference adaptive control techniques can be used to actively make adjustments to the input of the ear in order to compensate for a specific perception of one's defective ear. Further learning techniques can be utilized in implementation in order to precisely adjust for each subject's unique limitation.**

***Index Terms*—hearing compensation, control, bioelectricity**

I. Introduction

The improvement of hearing loss compensation techniques is an ongoing area of research in medical and engineering fields. Many physical devices have been invented to overcome permanent sensorineural hearing loss using various acoustic methods, signal processing techniques, implementation processes, and biomechanical modifications.

A. Model of Human Ear

It is important to consider perceived sound in the frequency spectrum since human hearing is over a limited range of frequencies (approximately 20Hz-20kHz) [1] [2]. The unit phon is used to measure the perceived loudness sounds. Ideally, the loudness level (in phon) for a given individual is flat for all ranges of frequencies [1]. For a given individual with approximately perfect hearing, the human ear might be modeled similar to a bandpass filter:

$$H(j\omega) = \frac{K\dfrac{\omega_c}{Q}}{s^2 + \dfrac{\omega_c}{Q}s + \omega_c^2} \tag{1}$$

where Q is the ratio of center frequency to bandwidth : $Q = \omega_c/\beta$.

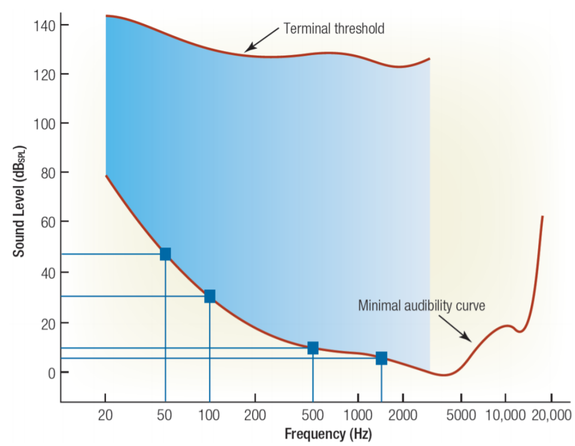

Fig. 1. Frequency response for normal equal-loudness-level contour at 0dB (minimum audibility). This is the relationship used to derive the approximate model of the ideal human ear. [3]

1) Mathematical Model for Minimum Audibility Curve (from ISO 226:2003): Loudness level is the linear metric for perceived magnitude of acoustic pressure for which a relationship exists with actual sound pressure level (SPL). It is important to note that $1Pa = 94dB\ SPL$ and $20\mu Pa = 0dB\ SPL$ [2]. The relationship between SPL and loudness is the following [1]:

$$L_p = \left(\frac{10}{\alpha}\log_{10}(A_f)\right) dB - L_U + 94dB \qquad (2)$$

and the minimum audibility weighting (shown in figure 1 can be applied with this mathematical curve equation:

$$\begin{aligned} A_f &= 4.47 \times 10^{-3} \times \left(10^{0.025 L_N} - 1.15\right) \\ &+ \left[0.4 \times 10^{\frac{T_f + L_U}{10} - 9}\right]^{\alpha_f} \end{aligned} \qquad (3)$$

This equation will probably be easier to model in the frequency domain. We can also use the value given in ISO 226 for the specific frequencies in order to approximate the model as a LaPlace-domain transfer function that can be used to derive an accurate system model for the human ear [1].

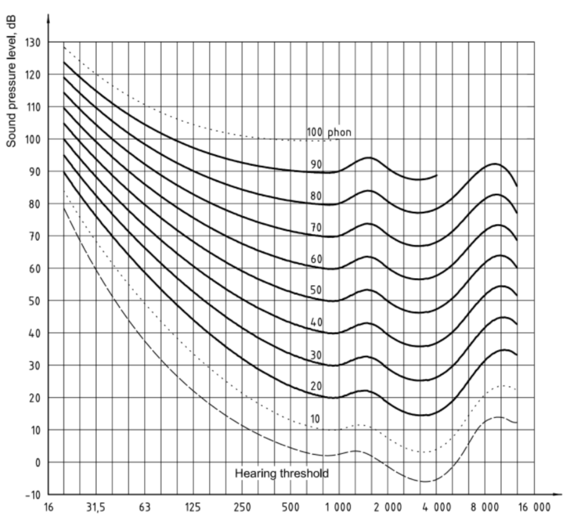

Fig. 2. Frequency response for normal equal-loudness-level contours at different loudness levels. [1]

2) Equal-Loudness-Level used for Comfortable hearing: In Fig. 2, the equal-level-loudness for multiple levels is shown. The solid-lined relationships were found experimentally, the others are estimated. Each relationship relative to each phon level is almost exactly the same relationship as the minimal audibility curve shown in Fig. 1. Consider that the scale is from 0 to 100 phon (which is below the threshold of pain), and 100 phon is considered the maximum loudness before discomfort, 50 phon can be regarded as the optimal perceived loudness comfort level.

II. Control System Techniques for Hearing Compensation

Before designing a controller for the defective ear, the system will be identified and a model derived for active comparison and reference.

A. Finding Frequency Domain Transfer Function of Human Ear

From the equal-loudness-contours in Figure 2, the 50 phon curve can be regarded as the optimal perceived loudness comfort level. The proposed adaptive controller must be designed to produce a system response that closely matches this 50 phon curve. In order to design such a controller, the LaPlace domain transfer function of the 50 phon curve must be found. To do this, the curve was inverted and normalized to be 0dB at 1kHz. This inverted curve, shown in Figure 3, shows the sensitivity of the ideal human ear to specific frequencies.

In Figure 3, the blue data points were taken from the 50 phon equal-level-loudness contour in ISO 226. A moving average filter was applied to this data to obtain the dotted red curve. This filter is used to smooth the curve to produce a more simplified transfer function. A simplified transfer function will improve the processing time of the controller and will reduce delays within the system. This will provide a better user experience to the listener.

The transfer function of the human ear can be approximated using a simple bandpass filter, with passband frequencies between 20 Hz and 20 kHz. The poles and zeros of a simple bandpass filter can

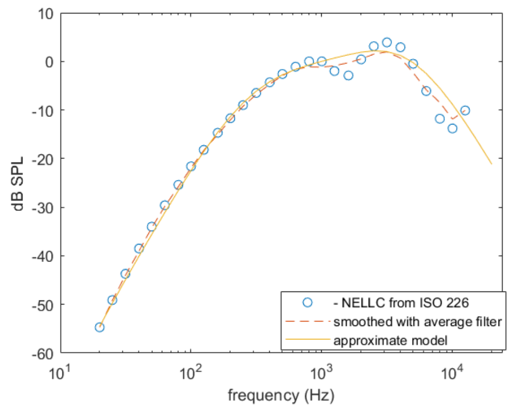

Fig. 3. Sensitivity Curve for Ideal Human Ear.

be adjusted to produce a magnitude plot, shown in yellow, that closely resembles the smoothed sensitivity curve.

Equation 5 shows the derived poles and resulting transfer function of the ideal human ear.
Poles:

$$p_{1,2,3,4,5} = [40, 1800, 22000, 25000, 10^6] \tag{4}$$

$$P = \frac{(4.53 \times 10^{10})s^2(s + 12000)}{(s + p_1)(s + p_2)(s + p_3)(s + p_4)(s + p_5)} \tag{5}$$

The decibel level of the frequency response can be shifted up or down by varying the gain of the transfer function. The gain of this transfer function was derived to produce a magnitude of 0dB at 1 kHz.

B. *Design of Controller*

Refer to Figure 4 where the system plant is the model for the ear. In figure 4, the intuitive control system is taken to ensure tracking of the output of the model. No matter the magnitude of the error between the outputs, the system compensates by injecting a factor of the error into the input of the defective ear, thus ”aiding” the hearing of the listener.

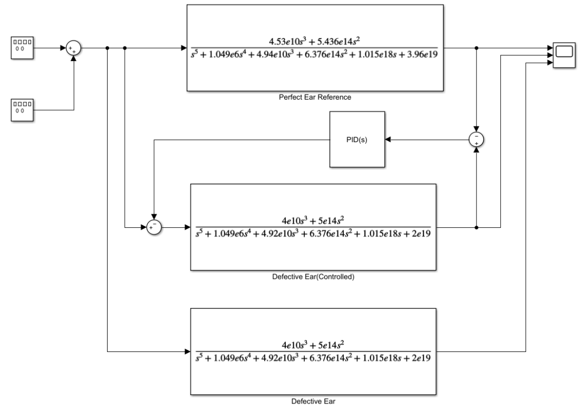

Fig. 4. Basic system model for an applied feedback control system under ideal implementation conditions. This system is used to compare three systems: the ideal ear, the defective ear, and the controlled defective ear.

III. Analysis

A. *Error Analysis*

Since the type of inputs for this system is a broad range of mixed periodic sound pressure variations, the type of error with the largest concern is tracking error. With the human ear as a plant, the desired controlled output should track that of the output of the model of the ideal ear. In order to assess this characteristic of the system we can test a range of of inputs for the system and observe the output compared to the output of the ideal model. Because of this particular application, much information can be observed from the bode plot of the system. MATLAB Simulink was used in order to compare the bode plots of each system. The output of the controlled defective ear should track with the ideal ear. During the testing of the controller, the defective ear was altered by applying a certain magnitude of attenuation and shifting the poles for a less ideal response.

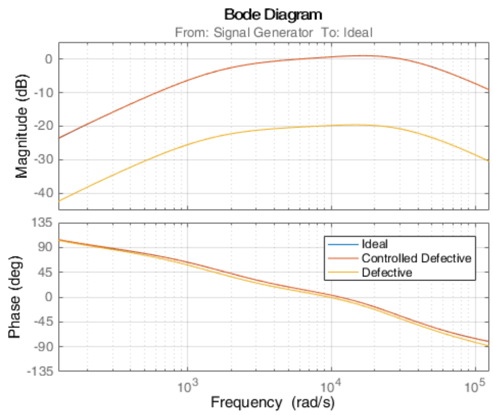

Fig. 5. From this bode plot of a sample defective ear, it's controlled output, and the output of the ideal ear, the compensations do the the feedback control is verified to be quite precise.

B. Stability Analysis

First, let's observe the plant of the system. Because of the nature of it's model's derivation, we know that all the poles are strictly-left-hand-plane (SLHP). Since the closed-loop system is extremely complex, we will use MATLAB to plot the Nyquist diagram of the controlled system. From the Nyquist plot shown in figure 6 we can see this stability even under extreme hearing loss in the simulation.

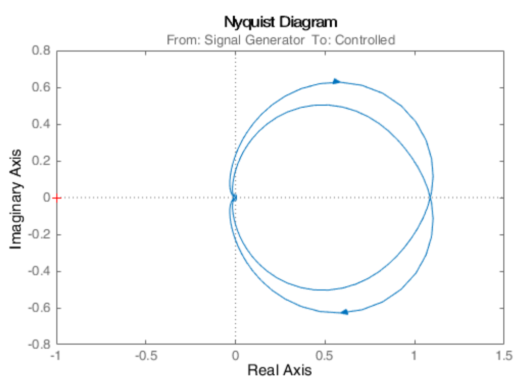

Fig. 6. From this Nyquist diagram of the controlled defective ear system, the stability of the system becomes obvious.

IV. Conclusion

Under the assumption that the actual physical output of one's perception can be monitored, using feedback to control the input in order for compensation can be quite practical. However, the technology does not yet exist to accurately and actively measure how one's perception compares to actual sound pressure levels. It is also important to note that the presented design doesn't account for any type of processing time for the controller to work although in application, the measured output would need to be processed in order to determine the exact error.

A. Moving Forward

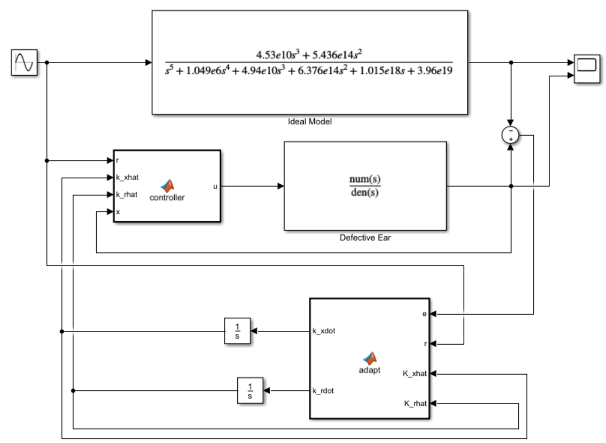

Fig. 7. Simulink block diagram of MRAC-controlled hearing compensation.

Figure 7 shows a model-reference adaptive control (MRAC) design is implemented. This design improves upon the simple PID-controlled error because the system is able to adapt for variations in the particular defects of the plant (ear). In addition to this adaptation, the controller can identify the system as it changes. This feature is vital when a processor is used with particular function that compensates for the particular defects of an unhealthy ear. Instead of compensating without any regard to the plant, a system utilizing MRAC can update the controller which is tuned precisely for a specific ear.

References

[1] International Standard. *Acoustics - Normal equal-loudness-level contours*, 2nd ed. ISO 226:2003(E).

[2] National Research Council (US) Committee on Disability Determination for Individuals with Hearing Impairments. *Hearing Loss: Determining Eligibility for Social Security Benefits*. Dobie RA, Van Hemel S, editors. National Academies Press (US). 2004. www.ncbi.nlm.nih.gov/books/NBK207834/

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