COMPUTER MODELING OF ACOUSTICAL ELEMENTS OF A HEARING AID

G. SZWOCH [1], B. KOSTEK [1,2], A. CZYŻEWSKI [1]

[1] Technical University of Gdańsk
Sound and Vision Engineering Department
(80-952 Gdańsk, ul. Narutowicza 11/12)

[2] Institute of Physiology and Pathology of Hearing
(01-943 Warsaw, Pstrowskiego 1)

In this paper, application of computer modeling methods to the process of hearing aid fitting is described. A computer model of the acoustical system of a hearing aid is presented. Exemplary results of the experiments are presented and compared with measurement data. The model proved to behave similarly to the physical system. Further improvements to the model are discussed.

1. Introduction

Computer modeling methods are commonly used when there is a need to examine properties of a given system before it is constructed. Computer simulations, however, were not applied so far to the process of fitting of acoustical elements of hearing aids. Currently, the fitting process is based on choosing the acoustical elements from a variety of configurations that differ in their acoustic properties. This is often a trial-and-error process, which is time-consuming and is tiresome for a patient. Therefore, using modeling methods in order to design a system having desired properties seems reasonable [5][6].

A computer model is intended to examine acoustic properties of systems with various sets of parameters (dimensions, shape, materials, etc). Particularly, the computer model enables one matching its transfer function to the type of the hearing loss of a patient. Using the computer model, the user is able to compare results of simulations performed for various sets of parameters of the model and choose the most suitable one to create the acoustical system of a hearing aid, well-matched to the patient's needs.

In this paper, the main part of the computer model of the acoustical system of a hearing aid is presented. The proposed model simulates the acoustical system as a set of cylindrical sections of different sizes (the waveguide). Experiments were performed in order to examine the influence of changing parameters of the model on its transfer function. The exemplary results of these experiments are included in this paper. The simulation results were compared with the measurements results of the real acoustical elements of a hearing aid. Finally, some improvements of the model are discussed which may augment its accuracy and effectiveness

2. Computer Model of Acoustical Elements of a Hearing Aid

The task of the acoustical system of a hearing aid is to transmit sound waves produced by the receiver of the hearing aid to the ear canal of the user of the apparatus [2]. In the Behind-The-Ear (BTE) hearing aid type, this system comprises three main elements (Fig. 1). An earhook, made of hard plastic, protects acoustical converters of the hearing aid. A tubing is a long and narrow elastic tube which connects earhook to the earmold. An earmold is the most important and most complicated part of the system. Many earmold types are available, differing in their shape and material they are made of, as well as in shape and dimensions of the canal through which sound waves are transmitted to the ear [4][5]. In the miniaturized types of hearing aids (In-The-Canal types), the acoustical system is reduced to the small earmold.

Several attempts to model the acoustical system of a hearing aid were done by various researchers in the past. Most of them were based on the analogies between acoustical and electrical systems. These models are complicated and computing the transfer function of such models is time-consuming and requires a great amount of computer resources. Moreover, it is difficult to change the model parameters, such as dimensions or size of the system. Therefore, a different approach is proposed in this paper. It is based on the physical modeling methods.

Physical modeling has been successfully implemented to model acoustical systems such as the human vocal tract [6] or the organ pipe [1]. The acoustical system of a hearing aid is similar to the above-mentioned systems. The system can be represented as a set of cylindrical sections that differ in diameter and length. The model composed of cylindrical tubes is called the waveguide model [9]. This model is valid as long as the propagation of sound waves in the system is described by the one-dimensional wave equation. In other words, no transversal modes in the waveguide exist. This is true only for frequencies below the critical value given by [9]:

$$f_c = 1.84 \frac{c}{2\pi a}, \tag{1}$$

where c is the velocity of sound (c = 343 m/s) and a is the radius of the waveguide. Since the frequency band in hearing aids is limited, one can assume that f_c is equal to 11.025 kHz, thus based on Eq. 1 the radius a must not exceed 9.111 mm. This condition is fulfilled in the acoustical system of the hearing aid, hence physical modeling methods may be used.

The physical (waveguide) model of the acoustical system can be based on the Markel-Gray model, already applied to modeling of the vocal tract [6]. This model was slightly modified in order to remove the arbitrary limit on the length of each cylindrical section [7][8]. This does not affect the efficiency of the model as long as its transfer function is computed mathematically (off-line mode). The model may be also implemented to work in real-time. However, digital interpolators are required in order to obtain fractional delays [9].

The general block diagram of the waveguide model is presented in Fig. 2. Propagation of sound waves through the cylindrical sections is simulated using the delay lines d. Length of the delay line depends on the length of the tube, which is given by the expression:

$$d = \frac{f_S}{c} L, \tag{2}$$

where L is the length of the cylindrical segment in meters, c is the velocity of sound, f_S is the sampling frequency. The value of d in the model described in this paper is not an integer in most of the cases.

Since each two adjacent cylindrical segments differ in diameter, there is a discontinuity in the acoustic impedances at the point of connection. Therefore, the partial reflection of the sound waves occurs. This phenomenon is described in terms of reflection coefficients [4][7]:

$$r_i = \frac{S_i - 1}{S_i + 1} \tag{3}$$

In the formula presented above, S_i is:

$$S_i = \left(\frac{a_{i+1}}{a_i} \right)^2 \tag{4}$$

and a_i is the radius of the i-th cylindrical segment.

The most difficult problem to solve in modeling process is the simulation of the interaction between the waveguide and the outer media. In the model of the acoustical system of a hearing aid this means the interaction between the earhook and the receiver, as well as between the earmold and the ear canal and tympanic membrane. The modeling of these interactions can be done by setting the values of the input and output reflection coefficients, shown in Fig. 2 as r_0 and r_N respectively. Generally, these coefficients model the energy losses of the sound waves leaving the waveguide. Values of these coefficients depend on the diameter of the waveguide at its terminations, the density of air inside the waveguide and the acoustic impedance of the outer medium. However, input and output reflection coefficients may incorporate other energy losses in the model, resulting from physical phenomena such as viscous friction between the air and the walls of the waveguide, heat conduction through the walls or vibration of the walls [6]. Thus, the proper modeling of all these energy losses becomes a complicated matter. Furthermore, since most of these phenomena are frequency-dependent, constant input and output coefficients have to be replaced by digital filters.

In order to examine acoustical properties of the model, its transfer function has to be calculated and plotted. This can be achieved using a method similar to the one applied by Rabiner and Shaffer to the original Markel–Gray model [6]. The transfer function of the model shown in Fig. 2 has the following form:

$$H(z) = \frac{\prod_{i=1}^{N} (1 + r_i)}{\begin{bmatrix} 1 & -r_0 \end{bmatrix} \prod_{i=1}^{N} \begin{bmatrix} z^{di} & -r_i z^{di} \\ -r_i z^{-di} & z^{-di} \end{bmatrix} \begin{bmatrix} 1 \\ 0 \end{bmatrix}} \tag{5}$$

where N is the number of cylindrical sections, di is the length of the i–th delay line ($i = 1...N$), r_i is the reflection coefficient ($i = 1...N-1$), $r_0 \equiv F_{in}(z)$, $r_N \equiv F_{out}(z)$. Eq. 5 shows that the waveguide model can be treated as the N–th order digital filter, having only poles.

3. Experiments

A simple computer model of the acoustical system of the hearing aid, described in the previous section, was used in experiments. The earhook and the tubing were both modeled as single cylindrical tubes. The earmold canal was modeled as a number of cylindrical sections, depending on the modeled configuration. Due to the complicated nature of physical phenomena resulting in energy losses, they were not directly implemented at this stage of the research. The constant value $r_0 = 0.3$ was chosen in order to obtain smooth shape of the transfer function curve. First-order lowpass IIR filter was used as the output reflection filter r_N:

$$r_N \equiv F_{out}(z) = \frac{a\left(1+z^{-1}\right)}{1+b\,z^{-1}}. \tag{6}$$

Coefficients of this filter: a = 0.226, b = –0.547, were chosen experimentally so that frequency bandwidth of the transfer function of the model matches the one of a real acoustic system.

All experiments were performed using the Mathematica computer system running on a typical personal computer. The transfer function of the model, given by Eq. 5, was computed and plotted, after substituting $exp(j2\pi f / f_S)$ for z, where f_S is the sampling frequency. Since the frequency range of most hearing aids does not exceed 11 kHz due to the limitations of the acoustical converters, the sampling frequency of 22.05 kHz was chosen as sufficient.

The transfer function of the model was plotted for various values of its parameters and the obtained results were compared. The length and the diameter of the earmold canal sections and the tubing were changed, as well as the number of sections comprising the earmold canal. The size of the earhook was not changed, its length was equal to 17 mm, its diameter – 1.8 mm. Dimensions of all elements of the system were adopted from real acoustical elements of a hearing aid [3][5]. Exemplary simulation results in a form of logarithmic plots of the transfer function magnitude versus frequency are included in this section (Figures 3 to 9). As seen from plots obtained from computer simulations, the main feature of plots is the presence of evident maxima (peaks). The frequency of the first, main peak is about 1 kHz. The frequency components above 1 kHz are attenuated due to the energy losses modeled by the filter r_N.

Experiments were performed in order to examine how the change of one of the model parameters, eg. size of one of its sections, affects the transfer function of the model. The obtained results of the simulations were not only compared with each other, but also with the results of measurements of the real acoustical elements of a hearing aid. These measurements were not performed by the authors, but they have been obtained from the manufacturers' data of modeled elements [3][4]. Due to comparison with the measurement results the authors were able to examine whether the computer model changes its parameters analogically as the real acoustical system does. The results are summarized below.

Fig. 3 illustrates the effect of increasing the length of the earmold canal from 10 mm to 20 mm. The size of the tubing was not changed, its length was equal to 45.8 mm and its diameter to 2 mm. In the performed simulations, the peaks are shifted towards lower frequencies and the height of the main peak is slightly increased (Fig. 3b). Similar effect can be noticed in the measurement plot [3], a decrease of the high frequencies, however, is more noticeable. The effect of shortening the canal to 2 mm causes opposite results (Fig. 4). In both measurement and simulation plots the peaks are shifted right on the frequency axis.

In another experiment, the diameter of the earmold canal was changed. The length of the canal was equal to 10 mm, the size of the tubing was the same as in the previous experiment. In simulations, decreasing the diameter of the canal from 2.4 mm to 1.1 mm (Fig. 5b) attenuates higher frequencies significantly and decreases the resonant frequencies. If the diameter is increased to 3 mm (Fig. 6b), the peaks are shifted to higher frequencies and amplitude of the transfer function in the high frequency range increases. These results are also consistent with measurement plots (Fig. 5a and 6a).

Results of another experiment, in which the size of the tubing was changed, are also in accordance with measurements of real acoustic elements. In theory, increasing the length of the tubing causes shifting the peaks towards lower frequencies, while decreasing the tubing length has the opposite effect [4]. Similar relationships were found during simulations. Shortening the tubing from 45.8 mm to 40 mm increases the resonant frequencies (Fig. 7a) while shortening the tubing to 50 mm decreases them (Fig. 7b). The diameter of the tubing was equal to 2 mm and was not changed.

In the next experiment, diameter of the tubing was changed. According to measurement results [4], increasing the diameter of the tubing shifts peaks to higher frequencies and the amplitude of the main peak increases. Decreasing the tubing diameter does the opposite. These effects are also evident in the plots obtained from computer simulations. If the length of the tubing is constant (equal to 45.8 mm) and its diameter is increased to 2.5 mm (Fig. 8a), the frequencies of the peaks increase and the height of the main peak rises. The opposite effect – lowering the height of the main peak and shifting the peaks to lower frequencies – was obtained by reducing the diameter of the tubing to 1.5 mm (Fig. 8b).

The examples presented above show that it is not possible to obtain significant amplification of higher frequencies by changing the length and diameter of tubing and earmold canal. Therefore, special modifications have to be introduced to the acoustic system. The most commonly used solution is implementation of the "horn effect" – using the canal of the increasing diameter, which results in amplification of the higher frequencies (as an example a Libby horn earmold can serve) [3][5]. In order to implement this effect in the model, the earmold canal was divided into two or three cylindrical sections of different diameter (Fig. 9). Results of the simulation show that when the diameter of each consecutive cylindrical segment increases, the main peak is shifted towards higher frequencies and the frequency band above 1 kHz is significantly amplified (Fig. 9a). This effect is more significant if three such segments are used (Fig. 9b). In addition, the reverse horn effect can be used if the attenuation of higher frequencies is needed (Fig. 9c) The length and diameter of the tubing and hook, as well as filter coefficients, were the same as in previous examples.

4. Conclusions

The computer model of the acoustical system of a hearing aid was presented in this paper. This model was used in experiments in which its transfer function was computed and then plotted. The results of the experiments are generally consistent with the results of measurements obtained by the manufacturers of the acoustic elements. Changing the model parameters has similar influence on its transfer function as changing the corresponding parameters of the acoustical system. Additionally, changing the shape of the earmold canal in a proper way enables one to amplify or attenuate high frequency components. Therefore, it may be concluded that the computer model proposed and described in this paper properly simulates the acoustical system of a hearing aid. The major differences concern the magnitude values on the transfer function plot. Therefore, a normalization in experiments is required in order to obtain the magnitude values in the decibel scale the same as in measurements.

It should be, however, noted that the model described in this paper was simplified. In order to achieve greater accuracy of the model, some improvements should be introduced.

The most important problem is developing the method of simulating energy losses in the model. Moreover, the most important modifications used in real acoustical system of a hearing aid, such as venting canals and dampers, should be implemented in the model. These problems will be the subject of the next stage of research.

The results of experiments presented in this paper are promising. When the above-mentioned features are incorporated into the model, it will become the basis of the computer system which might be useful in the process of fitting of the hearing aid. Based on the computer simulations, one will be able to compare the acoustical properties of different waveguide systems, change the model parameters until proper transfer function is obtained and then use the simulation results to create the acoustical system of a hearing aid well-fitted to the user's needs. Such a system is not intended to replace the physicians, but to optimize their work by providing the fast and efficient method of designing acoustical elements of a hearing aid.

Acknowledgements

The research is sponsored by: Committee for Scientific Research, Warsaw, Poland, grant No. 8 T11E 008 19; Foundation for Polish Science, Warsaw, Poland; partially by Institute of Physiology and Pathology of Hearing, Warsaw, Poland.

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Figure captions

Fig. 1. The main parts of BTE hearing aid: (a) overview, (b) earmold [5]

Fig. 2. General model of the acoustical system of the hearing aid; r denotes reflection coefficient, d – delay line length

Fig. 3. Increasing the length l of the earmold – plots of transfer function: (a) measurements of the real earmold [3]: l = 10 mm (dashed line) and l = 20 mm (solid line); (b) computer simulations: l = 10 mm (thin line) and l = 20 mm (thick line)

Fig. 4. Decreasing the length l of the earmold – plots of transfer function: (a) measurements of the real earmold [3]: l = 10 mm (dashed line) and l = 2 mm (solid line); (b) computer simulations: l = 10 mm (thin line) and l = 2 mm (thick line)

Fig. 5. Decreasing the diameter d of the earmold – plots of transfer function: (a) measurements of the real earmold [3]: d = 2.4 mm (dashed line) and d = 1.1 mm (solid line); (b) computer simulations: d = 2.4 mm (thin line) and d = 1.1 mm (thick line)

Fig. 6. Increasing the diameter d of the earmold – plots of transfer function: (a) measurements of the real earmold [3]: d = 2.4 mm (dashed line) and d = 4 mm (solid line); (b) computer simulations: d = 2.4 mm (thin line) and d = 4 mm (thick line)

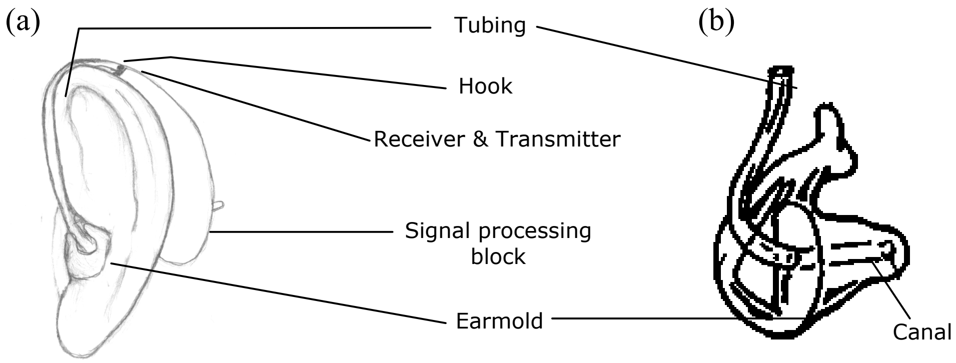
Fig. 7. Influence of changing the length l of the tubing on plots of transfer function in simulations: (a) l = 45.8 mm (thin line) and l = 40 mm (thick line); (b) l = 45.8 mm (thin line) and l = 55 mm (thick line)

Fig. 8. Influence of changing the diameter d of the tubing on plots of transfer function in simulations: (a) $d = 2$ mm (thin line) and $d = 2.5$ mm (thick line); (b) $d = 2$ mm (thin line) and $d = 2.5$ mm (thick line)

Fig. 9**.** Plots of transfer function of the computer model for varying shape and dimensions of the earmold canal; thin line: cylindrical canal of length $l = 10$ mm, diameter $d = 2.4$ mm; thick line: (a) $l_1 = l_2 = 5$ mm, $d_1 = 2.4$ mm, $d_2 = 4$ mm,

(b) $l_1 = l_2 = 3$ mm, $l_3 = 4$ mm, $d_1 = 2.4$ mm, $di = 3.5$ mm, $di = 5$ mm,

(c) $l_1 = l_2 = 3$ mm, $l_3 = 4$ mm, $d_1 = 5$ mm, $d_2 = 3.5$ mm, $d_2 = 2.4$ mm



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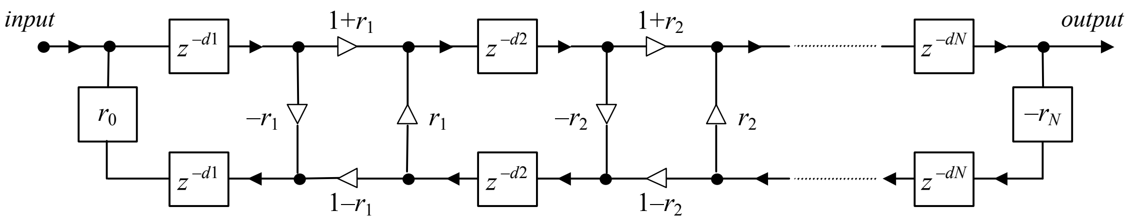

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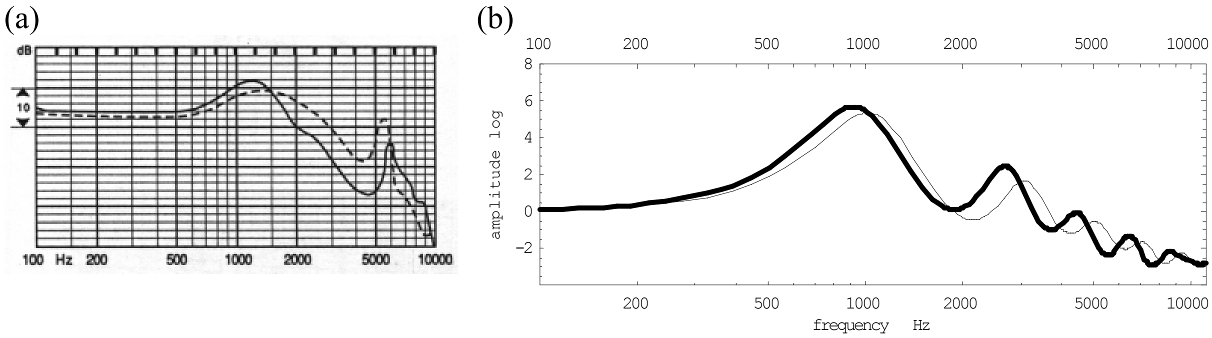

Fig. 3. Increasing the length l of the earmold – plots of transfer function:
(a) measurements of the real earmold [3]: $l = 10$ mm (dashed line) and $l = 20$ mm (solid line)
(b) computer simulations: $l = 10$ mm (thin line) and $l = 20$ mm (thick line)

(a) (b)

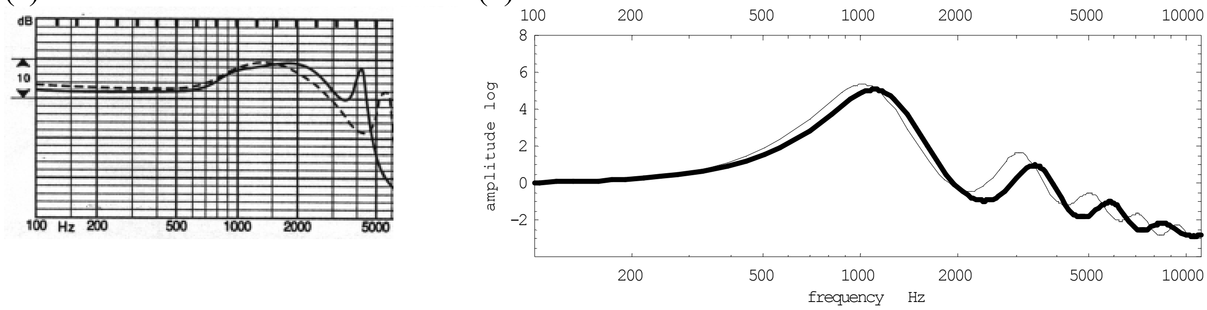

Fig. 4. Decreasing the length *l* of the earmold – plots of transfer function:
(a) measurements of the real earmold [3]: *l* = 10 mm (dashed line) and *l* = 2 mm (solid line)
(b) computer simulations: *l* = 10 mm (thin line) and *l* = 2 mm (thick line)

(a) (b)

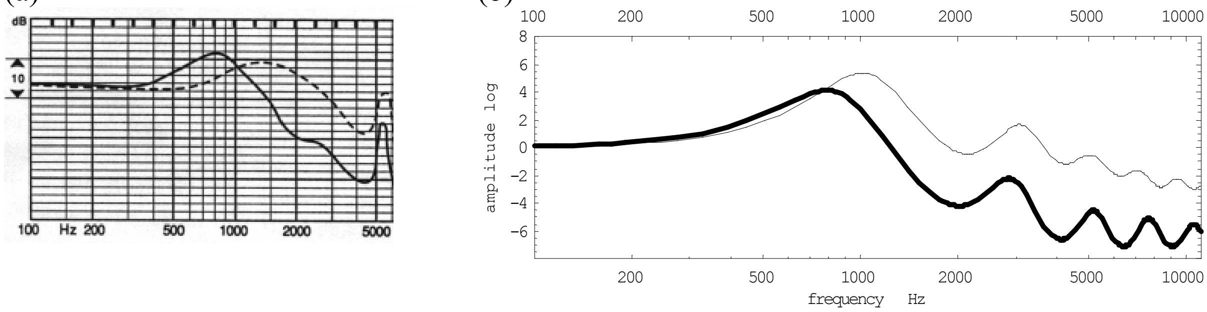

Fig. 5. Decreasing the diameter *d* of the earmold – plots of transfer function:
(a) measurements of the real earmold [3]: *d* = 2.4 mm (dashed line) and *d* = 1.1 mm (solid line)
(b) computer simulations: *d* = 2.4 mm (thin line) and *d* = 1.1 mm (thick line)

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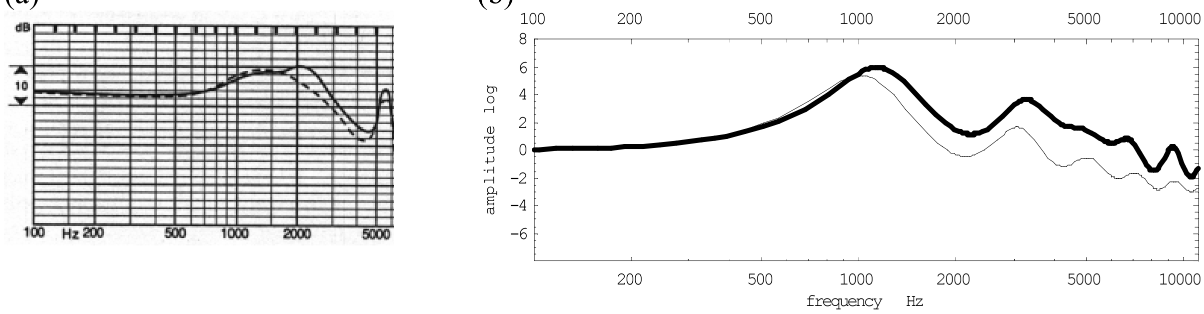

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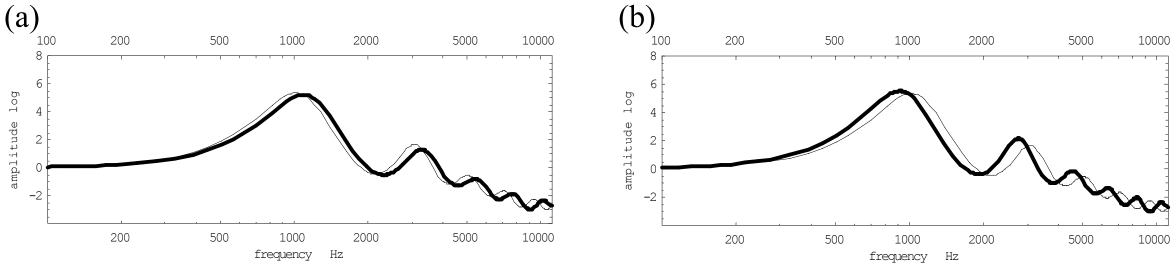

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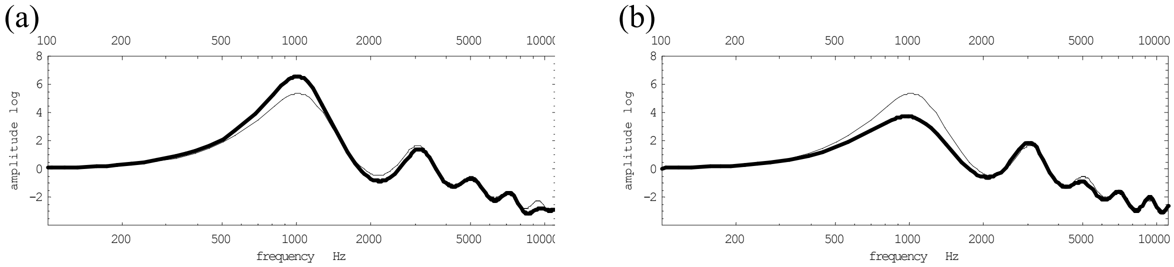

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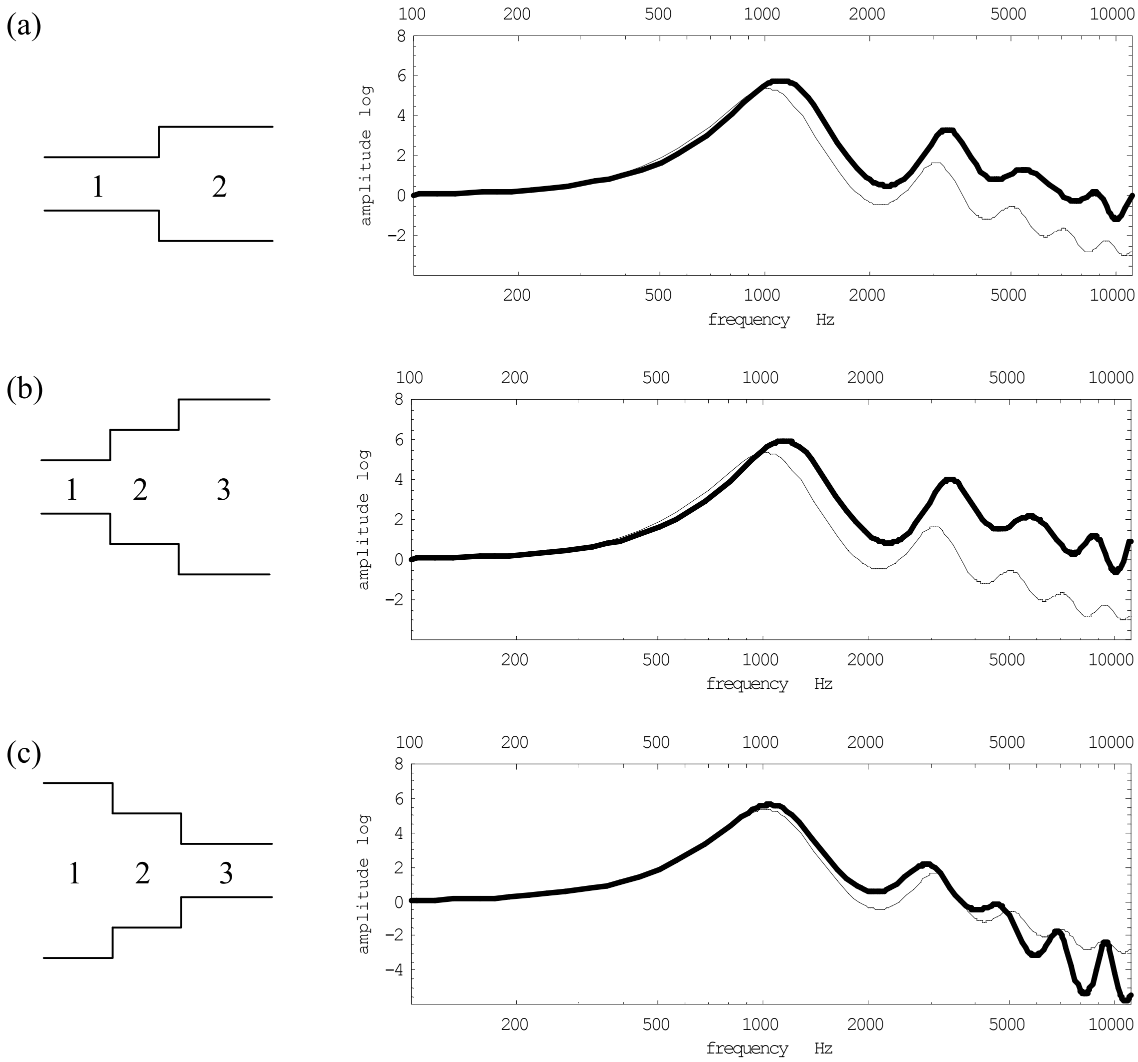

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(b) $l_1 = l_2 = 3$ mm, $l_3 = 4$ mm, $d_1 = 2.4$ mm, $di = 3.5$ mm, $di = 5$ mm,
(c) $l_1 = l_2 = 3$ mm, $l_3 = 4$ mm, $d_1 = 5$ mm, $d_2 = 3.5$ mm, $d_2 = 2.4$ mm