JOURNAL OF NANO- AND ELECTRONIC PHYSICS
Vol. **16** No 3, 03030(8pp) (2024)
ЖУРНАЛ НАНО- ТА ЕЛЕКТРОННОЇ ФІЗИКИ
Том **16** № 3, 03030(8cc) (2024)

REGULAR ARTICLE

Human Inner Ear Implant with an Acoustic Information Transmission Channel

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(Received 25 March 2024; revised manuscript received 22 June 2024; published online 28 June 2024)

The paper considers an implant of the human inner ear as an electroacoustic system. The implant serves to partially restore human hearing by direct impact on the nerve endings of the coil with electrical impulses. Electrical stimulation of the auditory nerve, in turn, leads to auditory sensations. A block diagram of an implant using an ultrasonic communication line between the external and internal parts of the system is selected. A theoretical study of the influence of parameters on the frequency properties and sensitivity of the ultrasonic line of acoustic information transmission was carried out, which is based on a mathematical algorithm that implements the calculation of the sensitivity of the transmission channel depending on the frequency and channel parameters. A multilayer model of the communication line was chosen for calculations, which contains emitting and receiving piezo plates, two protectors to ensure safe contact between the receiving and emitting piezo plates and bio tissue, an intermediate layer of bio tissue, two dampers that are adjacent to the emitter and receiver, respectively. At the same time, it is considered that the dampers have limited dimensions. As a result, the influence of individual design parameters of the transmission line on the maximum sensitivity of the line and its frequency characteristic is theoretically researched.

Keywords: Acoustics, Electronics, Implant, Electroacoustic transducer, Human inner ear, Emitting mode, Reception mode, Frequency response.

DOI: 10.21272/jnep.16(3).03030 PACS number: 43.38. + n

1. INTRODUCTION

It is known that in some cases the disease disables that part of the hearing apparatus of a person that precedes the formation, transmission, and subsequent processing of electrical impulses in the central nervous system. At the same time, the hearing apparatus, starting with the efferent nerve fibers, remains intact. For example, the function of hair cells, which transform mechanical vibrations in the inner ear into electrical ones, which are then transmitted to the "input" of efferent nerve fibers and further into the central nervous system, may be impaired. Then, to partially restore hearing, it is possible to use a direct effect on the nerve endings with an electrical impulse. Electrical stimulation of the auditory nerve leads to auditory sensations. A human inner ear implant serves this purpose. For its implementation, an electroacoustic system was chosen, the block diagram of which is shown in Fig. 1.

The system with electrode implantation consists of three main parts (Fig. 1): microphone 1; sound processing devices (blocks 2-9); a row of electrodes 10.

The system operation principle is as follows. Sound vibrations are converted into electrical signals using a microphone 1, which is attached using an ear hook. These signals are then guided to a signal processing device worn on the body. They are fed to the microphone amplifier 2 with an adjustable gain. Depending on the level of the signal at its output, the dynamic range of signals at the output of compressor 3 changes.

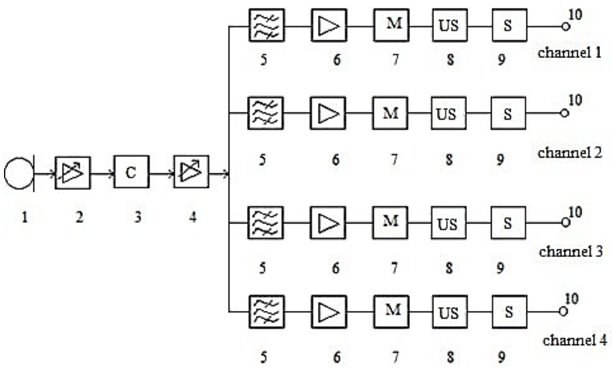

Fig. 1 – Block diagram of a human inner ear implant

The compressor compresses signals into a narrow dynamic range, which is typical for most patients. From the compressor, the signals are sent to the amplifier 4, changing the amplification factor, which affects the volume level. Then the signals are separated depending on the frequency using 4-band filters 5. After that, the signals are fed to the modulator 7.

Two versions of the system are possible: without encoding the speech signal and with its encoding [1-3].

In the first variant, the speech signal is changed by the parameters of the modulator. The carrier frequency is chosen equal to the operating frequency of the piezo elements used in the acoustic communication line, namely, 2 MHz. Depending on the initial transmitting

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Cite this article as: S.A. Naida et al., *J. Nano- Electron. Phys.* 16 No 3, 03030 (2024) https://doi.org/10.21272/jnep.16(3).03030

signal, the modulator changes the amplitude of high-frequency oscillations (amplitude modulation). At the same time, an analog signal is applied to the electrodes.

In the second variant, the speech signal is converted into a digital form, and a sequence of pulses (pulse coding) is applied to the electrodes. Both options differ only in the device of the modulator block 7.

The modulated signal is returned to the ear hook and fed to the ultrasonic (US) communication line 8. The function of this element of the system is to transmit the signal in the coil, that is in the implanted part of the system. In the implant, the signal enters the demodulator 9 (amplitude detector), which emits a useful speech signal. After that, the signal is sent to the final element of the system – a row of electrodes 10, which consists of 6 platinum spherical elements. Each electrode uses a common ground located in the parietal part. This is the so-called unipolar excitation, in which the threshold is twice as low as in bipolar, when each electrode has its own ground. As a result, we have the opportunity to excite the electrodes with a smaller current.

In this system, the principle of place is used, which consists in the fact that a certain place in the curl of the inner ear of a person corresponds to a certain frequency of sound. Thus, the low-frequency electrode is located at 20-24 mm from the round window, and the high-frequency electrode is located at 8-12 mm from the round window.

Loudness in the analog version of the system is coded by a change in the amplitude of the output current, that is an increase or decrease in excitation. In the second, pulse variant, the volume is coded by a change in the charge density. The charge density is defined as the product of the pulse width and the amplitude. It should be noted that the dependence of sound pitch on the frequency of electrical excitation complicates intensity coding.

The system uses synchronous excitation of the electrodes.

A simplified model can be used to establish the possibility of pulse transmission of the required amplitude and duration, which encode acoustic information, from the external generator to the stimulating electrodes of the implant. Such model contains in addition to the mandatory elements, such as the emitting and receiving piezo plates, and the intermediate layer of bio tissue, two more semi-infinite dampers, which are adjacent to the emitter and the receiver, respectively. However, the real acoustic path of the ultrasonic transmission line (one of its channels) is more complicated. Firstly, the dampers have limited and very small dimensions: the total thickness of each block of the ultrasonic line (external - emitting and implanted - receiving) cannot be more than 5-6 mm. This size includes the thicknesses of the piezo plates, the damper, and the intermediate layer between the piezo plate and the bio tissue - the so-called protector (which here plays a different role than the protector in flaw detectors and medical ultrasonic diagnostic devices). The need for a protector as part of the acoustic path is due to the need to ensure safe contact between the receiving and emitting piezo plates and bio tissue.

Such complication of the acoustic path requires solving a few issues at the stage of construction development. Namely, what should be the thickness and parameters of the material of the dampers, the thickness, and parameters of the material of the protectors, what kind of cover should be used to close the damper on the reverse side: acoustically flexible or acoustically rigid? Since, as shown at the previous stage, there are thickness resonances in the ultrasonic line, the choice of the above parameters can affect both the frequency properties of the line and the value of absolute sensitivity, that is the important technical parameters of the ultrasonic line.

The stated considerations justify the need for a theoretical study of the influence of parameters on the frequency properties and sensitivity of the ultrasonic line for the transmission of acoustic information. This theoretical study is based on a mathematical algorithm that implements the calculation of the sensitivity of the transmission channel depending on the frequency and channel parameters. The justification of the mathematical algorithm, some calculation results and their analysis are presented below.

2. CALCULATION OF A MULTILAYER ULTRASONIC LINE FOR THE TRANSMISSION OF CODED ACOUSTIC INFORMATION

2.1 Justification of the Calculation Algorithm of the Multilayer Structure of the Acoustic Path of the Ultrasonic Transmission Line

The schematic acoustic path is shown on Fig. 2.

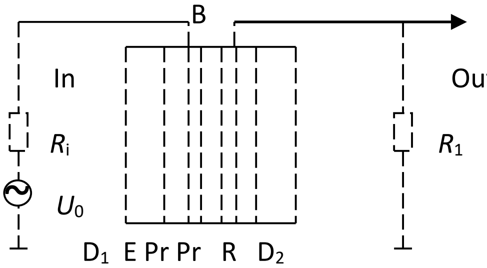

Fig. 2 – Acoustic path of the ultrasonic transmission line

The following designations are used in Fig. 2:

U_0 – emf developed by the output amplifier of the electronic unit;

R_i – the internal resistance of the output amplifier;

D_1, D_2 – emitter and receiver dampers;

E – emitting piezo plate;

Pr – protectors;

R – receiving piezo plate;

B – bio tissue;

R_1 – the load resistor.

Acoustic vibrations in each of the layers shown in Fig. 2, can be imagined as the sum of two opposing waves with amplitudes not determined in advance. In general, there are 7 unknown quantities in the given problem. Contact and boundary conditions will provide us with the appropriate number of equations. As we can see, the direct approach can lead to a rather time-consuming computational procedure for solving a system of 7 equations with 7 unknowns, of which we may need only one to analyze the situation. Therefore, it is advisable to build the calculation differently, dividing the entire ultrasonic transmission line into three blocks (Fig. 3).

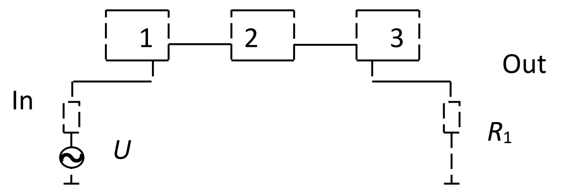

Fig. 3 – US transmission line

Unit 1 includes an emitting piezo transducer and its damper. At the same time, the damper is considered as an acoustic load on the left edge of the emitting piezo electric plate. Let us denote it by z_1. By definition, z_1 is the ratio of pressure (or mechanical stress, taken with the opposite sign) to the oscillating speed of the left edge of the emitting piezoelectric plate. Similarly, block 3 includes a receiving piezo plate with its damper, which is characterized by high acoustic resistance z_2. Block 2 contains a three-layer acoustic system, which is further intended to be considered as a three-chain acoustic long line.

Oscillations of each of the piezoelectric layers can be imagined as the sum of two opposing plane waves propagating in the piezoelectric ceramic. Particle oscillatory deflection of emitter [4]:

$$u_t = A_1 \exp\left(ik_k x\right) + A_2 \exp\left(-ik_k x\right), \tag{1}$$

and receiving device:

$$u_r = B_1 \exp\left(ik_k x\right) + B_2 \exp\left(-ik_k x\right), \tag{2}$$

where k_k is the wave number of ultrasound in piezoceramic material.

The amplitudes of waves A_1, A_2 and B_1, B_2 are pairwise related by the boundary conditions of their contact with the damper. This relation can be specified below after the equations for the waves in the piezo plate are obtained. As a result, A_2 and B_2 will be expressed through A_1 and B_1. The passage of ultrasonic waves through protectors and bio tissues is described using the transmission matrix of the equivalent long line, which depends only on the parameters in the corresponding layers and does not contain new unknown values. Thus, the computational procedure will be reduced to solving a system of two equations with two unknowns A_1 and B_1. As will be seen below, the value of the current in the load of the receiving piezo plate, which interests us, can be expressed in terms of B_1.

Let's move on to the equations describing elastic oscillations in electrically loaded piezo plates. Let's consider them on the example of an emitting piezo plate.

We will proceed from the well-known equations of the piezo effect, which relate the values of mechanical stress, oscillatory displacement u_t, electric intensity E and electric deflection D [5-7]:

$$\sigma = c^E_{33} du_t / dx - e_{33} E; \tag{3}$$

$$D = e_{33} du_t / dx + \chi^E_3 E. \tag{4}$$

Inside the piezo plate (for the region of ultrasonic frequencies) the integration of equation (4) allows us to relate the electric deflection D with the potential difference (voltage drop) on the piezo plate:

$$D \cdot h = e_{33}\left(u_t(h) - u_t(0)\right) + \chi^E_3\left(\phi(0) - \phi(h)\right), \tag{5}$$

where h is the thickness of piezo plate.

On the other hand, the voltage drop across the piezo plate can be found from Ohm's law for an electric generator:

$$U_0 = IR_i + \left(\phi(0) - \phi(h)\right). \tag{6}$$

Since the current through the piezo plate $I = dD/dt \cdot S = -i\omega DS$, and combining (6) and (5), we get:

$$D = \frac{e_{33}}{h}\frac{1}{1 - i\omega\tau_i}\left(u(h) - u(0)\right) + \chi^E_3 \frac{U_0}{h\left(1 - i\omega\tau_i\right)}. \tag{7}$$

where $\tau_i = R_i C_0$, C_0 is the static capacitance of the emitting piezo electric plate.

Eliminating from (1) and (2) intensity E we get:

$$\sigma = \rho_k c_k^2 du_t / dx - e_{33} D / \chi^E{}_3 .$$

After substituting the D values from (7), we obtain the equation for the oscillations of an electrically charged emitting piezo plate (equivalent to Hooke's law):

$$\sigma = i\omega\rho_k c_k \left[\frac{1}{ik} du_t / dx - a_i\left(u_t(h) - u_t(0)\right)\right] - \sigma_0, \tag{8}$$

where the following notations are introduced for convenience:

$$a_i = \frac{K}{1 + K^2}\frac{1}{ikh}\frac{1}{1 - i\omega\tau_i};$$

$$\sigma_0 = \frac{e_{33} U_0}{h\left(1 - i\omega\tau_i\right)}.$$

The appropriate expression for the receiving piezo plate differs only in that $\sigma_0 = 0$:

$$\sigma = i\omega\rho_k c_k \left[\frac{1}{ik} du_r / dx - a_r\left(u_r(h) - u_r(0)\right)\right]. \tag{9}$$

We will use relations (8) and (9) to find oscillatory deflections and mechanical stresses on the surfaces of piezo plates adjacent to the protectors, expressing them in terms of A_1 and B_1.

Substitute (1) into (8) and then consider that at the boundary of the emitting piezo plate and its damper, the mechanical stress and oscillatory deflection are related by the boundary condition:

$$\sigma = i\omega z_1 u_t . \tag{10}$$

From (10):

$$A_2 = A_1 r_1 - \sigma_0^{'} ,$$

where

$$r_1 = \frac{\rho_k c_k \left[1 - a_i\left(\exp(ikh) - 1\right)\right] - z_1}{\rho_k c_k \left[1 + a_i\left(\exp(ikh) - 1\right)\right] + z_1};$$

$$\sigma_0^{'} = \frac{\sigma_0}{\rho_k c_k \left[1 + a_i\left(\exp(ikh) - 1\right) + z_i\right]}.$$

Similarly for the receiving piezo plate we get: $B_2 = B_1 r_2$, where r_2 differs from r_1 only by replacing z_1 with z_2 and a_i with a_r.

After substituting the value of A_2 into formula (1), we find the distribution of deflections of the emitting piezo plate:

$$u_t = A_1\left[\exp(ikx) + r_1\exp(-ikx)\right] - \sigma_0^{'}\exp(-ikx),$$

and with the help of the θ - distribution of mechanical stresses in the piezoelectric plate:

$$\begin{gathered}\sigma = i\omega\rho_k c_k A_1[\left(\exp(ikx) - a\left(\exp(ikh) - 1\right)\right) - \\ -r_1\left(\exp(-ikx) + a\left(\exp(-ikh) - 1\right)\right)] + \\ +i\omega\rho_k c_k \sigma_0^{'}\left[\left(\exp(-ikx) + a\left(\exp(-ikh) - 1\right)\right)\right] - \sigma_0.\end{gathered}$$

According to these formulas, the voltage and deflection on the right border of the emitting piezoelectric plate are equal to:

$$\begin{gathered}u_t = A_1\left[\exp(ikh) + r_1\exp(-ikh) - \sigma_0^{'}\exp(-ikh)\right], \qquad (11)\\ \sigma_t = i\omega\rho_k c_k A_1[\left(\exp(ikh) - a_i\left(\exp(ikh) - 1\right)\right) - \\ -r_1\left(\exp(-ikh) + a_i\left(\exp(-ikh) - 1\right)\right)] + \\ +i\omega\rho_k c_k \sigma_0^{'}\left[\left(\exp(-ikh) + a_i\left(\exp(-ikh) - 1\right)\right)\right] - \sigma_0.\end{gathered}$$

Similarly, deflection and voltage on the left border of the receiving piezo plate, which is adjacent to the protector, are equal to:

$$u_r = B_1\left[\exp(ikh) + r_2\exp(-ikh)\right]. \qquad (12)$$

The magnitudes of mechanical deflections and stresses on the closely located faces of piezo plats are physically related due to the fluctuations of parts of the three-layer system "protector – bio tissue – protector". Let's establish this relationship using the matrix method.

Let $w_1 = \{u_1, \sigma_1\}$ is a vector whose projections are the deflection and voltage at the input of a certain layer, and $w_2 = \{u_2, \sigma_2\}$ is a similar vector at the output of the layer. It is known (and it is easy to show) that these two vectors are linearly related by the transition matrix through the layer M:

$$w_2 = M \cdot w_1,$$

where

$$M = \begin{Bmatrix}\cos(kh) & \sin(kh)/\omega\rho c \\ -\omega\rho c\sin(kh) & \cos(kh)\end{Bmatrix}. \qquad (13)$$

If the system consists of N layers, then the vector "deflection - voltage" at the output of the system with the same vector at its input is connected by a matrix that is the product of the transition matrices through individual layers. In our case, there are 3 such layers. Therefore, the vectors $w_r = \{u_r, \sigma_r\}$ and $w_t = \{u_t, \sigma_t\}$ are related as:

$$w_r = M_n \cdot M_\sigma \cdot M_n \cdot w_t,$$

where the transition matrices through the protector M_n and bio tissue M_σ are determined by the formula (13), if the parameters *ρ, c, k* and *h* correspond to appropriate indexes. Let's mark $M = M_n \cdot M_\sigma \cdot M_n$, and the elements of the matrix M through m_{11}, m_{12}, m_{21}, m_{22}. Then:

$$\begin{gathered}u_r = m_{11}u_t + m_{12}\sigma_t,\\ \sigma_r = m_{21}u_t + m_{22}\sigma_t.\end{gathered}$$

By substituting the expressions (11) and (12) here for u_t, σ_t and u_r, σ_r through A_1, B_1, we obtain a system of two equations with two unknowns. Having solved it, we will find the value of B_1:

$$\begin{gathered}B_1 = \frac{1}{D}\{[m_{21}q_1 + m_{22}(i\omega\rho_k c_k)s_1][-m_{11}\sigma_0^{'}\exp(-ikh) + \\ +m_{12}(i\omega\rho_k c_k)s_3] - [m_{11}q_1 + m_{12}(i\omega\rho_k c_k)s_1]\cdot \qquad (14)\\ \cdot[-m_{21}\sigma_0^{'}\exp(-ikh) + m_{22}(i\omega\rho_k c_k)s_3]\}.\end{gathered}$$

Determinant of the system of equations:

$$\begin{gathered}D = q_2\left[m_{21}q_1 + m_{22}(i\omega\rho_k c_k)s_1\right] - \\ -i\omega\rho_k c_k s_2\left[m_{11}q_1 + m_{12}(i\omega\rho_k c_k)s_1\right].\end{gathered} \qquad (15)$$

In formulas (14) and (15) for simplification, the following definitions are introduced:

$$\begin{gathered}q_1 = \exp(ikh) + r_1\exp(-ikh),\\ q_2 = \exp(ikh) + r_2\exp(-ikh),\\ s_1 = (\exp(ikh) - a_i\left(\exp(ikh) - 1\right) - \\ -r_1\left(\exp(-ikh) + a_i\left(\exp(-ikh) - 1\right)\right)),\\ s_2 = (\exp(ikh) - a_r\left(\exp(ikh) - 1\right) - \\ -r_2\left(\exp(-ikh) + a_r\left(\exp(-ikh) - 1\right)\right)),\\ s_3 = \sigma_0^{'}\left(\exp(-ikh) + a_i\left(\exp(-ikh) - 1\right)\right) - \sigma_0/(i\omega\rho_k c_k).\end{gathered}$$

The calculation *of* B_1 is an important, but intermediate result that allows us to find the output effect of the US transmission line, namely the current in the load (current through the exciting electrode).

First, we will find the electrical deflection in the receiving piezo plate using a formula like (7):

$$D = \frac{e_{33}}{h}\frac{1}{1 - i\omega\tau_r}\left(u_r(h) - u_r(0)\right),$$

where $\tau_r = R_1 C$ is the time constant in the receiver circuit.

The difference in deflections of the faces of the receiving piezo plate $u_r(h) - u_r(0)$, which is included in the formula for D, is determined using formula (12):

$$u_r(h) - u_r(0) = (q_2 - (1 + r_2)) \cdot B_1 .$$

Considering that $I = dD/dt \cdot S = -i\omega DS$, we finally get:

$$I = -\frac{e_{33}}{\chi_3^E} \frac{i\omega\tau_r}{1 - i\omega\tau_r} (q_2 - (1 + r_2)) \cdot B_1 , \qquad (16)$$

where B_1 is determined by the formula (14).

Further, we will additionally provide formulas for the input resistances of the dampers z_1 or z_2 , which were included in the previous expressions. If the rear wall of the damper is not loaded, then (omitting indexes 1 or 2):

$$z = -i(\rho c)_\partial tg(k_\partial h_\partial) .$$

When the wall of the damper is rigidly supported:

$$z = i(\rho c)_\partial ctg(k_\partial h_\partial) ,$$

where $(\rho c)_\partial$ is the wave resistance of the damper material;

k_d and h_d are the wave number in the damper and the thickness of the damper.

Usually, the material of the damper is distinguished by high absorption of ultrasound [6]. This is considered, like absorption in other line elements, by adding an imaginary part to the value of the speed of sound, and hence the wave number.

The above formulas served as the basis for developing a program for calculating the frequency dependence of the current at the output of the transmission line. The developed program allows us to arbitrarily change the material parameters of the line and study their influence on the value and amplitude of the current at the output of the line. Some results of calculations and their analysis are given in the next part of the paper.

2.2 Analysis of Calculation Results

The purpose of the calculations was to establish the influence of individual design parameters of the transmission line on the maximum sensitivity of the line and its frequency characteristic. Desirable properties of the line include, aside from operational and technological requirements, the ability of the line to transmit pulse signals without significant distortions which the calculated information is encoded. At the same time, the duration of the pulses can vary from 20 μs to 408 μs. Therefore, the bandwidth in the range of maximum sensitivity should be at least 50 kHz. Since the pulses of the code sequences are transmitted through the line at a certain carrier frequency, the choice of a specific value of the carrier frequency should be considered. Moreover, this frequency should be high enough so that:

a) Radiation beyond the implant was effectively absorbed by bone tissue and did not affect nerve cells. Considering the low level of ultrasonic signals, the situation here is far from critical, and the measures taken are aimed only at establishing the possibility of harmful influence to the maximum extent.

b) The dimensions of piezo transducers (emitters and receivers), as well as dampers, were small enough to fit into the limited volume of the implant and the loaded (emitting) part of the line.

c) The high frequency component after the detection of radio pulses of the code sequence could be filtered with the help of capacitors of a sufficiently small capacity, which, in the caseless version, could be placed in the implant.

d) With a certain Q-factor of the line, determined by its design and material properties, to ensure a sufficient bandwidth.

Frequencies in the range of 1-3 MHz meet these requirements. Finally, maximum transmission line sensitivity is desired. Its lower limit is determined by the condition that when a voltage with an amplitude of 10 V is applied to the emitting transducer, the amplitude of the current in the load, the value of which is 1 kΩ, should be at least 1.5 mA. This requirement can be weakened if there is no electrical contact of the emitter with the bio tissue. In this case, it is acceptable to apply high voltages to the emitter to obtain the required maximum current amplitude of 1.5 mA.

The changing parameters of the line, in its mathematical modeling within the framework of the model described above, include physical constants of the materials of the dampers, piezo plates and transition layers (protectors), their longitudinal dimensions (thickness), the method of fixing the rear side of the dampers: rigid fixing and no fixing (free edge). The bio tissue parameters are given: density $\rho_2 = 1064\ \text{kg/m}^3$, sound speed $c_2 = (1610 - i \cdot 10)\ \text{m/s}$ (adding an imaginary part to the value of the sound speed allows us to take into account the absorption of ultrasound in the bio tissue layer). The given values correspond to the acoustic properties of the skin. The thickness of the bio tissue layer was considered equal to 2 mm. The amount of current that a receiving-emitting pair of piezo plates can create in the load depends on the area of the contact surface. The last is limited by constructive circumstances. During the calculations, the value of the area was taken $151{,}71 \cdot 10^{-6}\text{m}^2$. Also, the influence of the output resistance of the final electrical cascade is known, which feeds the emitting piezoelectric plate. It was taken equal to 10 Ω. Already the first calculations showed that the difference between the accepted model and reality should be considered. This difference lies in the fact that in the mathematical model, the transverse dimensions of all layers in a multilevel structure were considered infinite. Under this condition, the layered structure, which is a complex multiresonant device, is a closed type resonator: ultrasound does not go beyond the resonator, and losses in it are caused only by dissipation in the material of the layers.

The graphs shown in Fig. 3, 4 show that, considering only internal losses, the transmission line has sharp resonance properties. The graph in Fig. 3 corresponds to the contact of the damper with a rigid cover, and the graph in Fig. 4 corresponds to the free rear side of the damper. At the same time, PZT-8 material was used for piezo plates ($\rho_n = 7600\,\text{kg/m}^3$, $c_n = (4580 - i \cdot 12)\,\text{m/s}$, relative dielectric constant $\varepsilon = 260$, coefficient of electromechanical coupling K = 0.51) [4]. The thickness of the piezo plate is 1.145 mm. The 4 mm thick layers of compound without filler were used as a damper ($\rho_d = 1160$ kg/m^3, $c_d = 2330 - i \cdot 15$ m/s), and layers of teflon with a thickness of 0.2 mm ($\rho_1 = \rho_3 = 900$ kg/m^3, $c_1 = c_3 = 1340 - i \cdot 18$ m/s) as a protector. The choice of Teflon is due to its compatibility with bio tissues. Analyzing Fig. 3, 4, it can be concluded that the case of the free rear side of the damper has advantages.

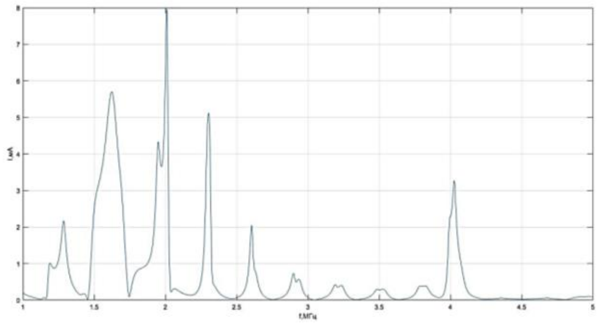

Fig. 3 – Frequency response for the case of damper contact with a rigid cover

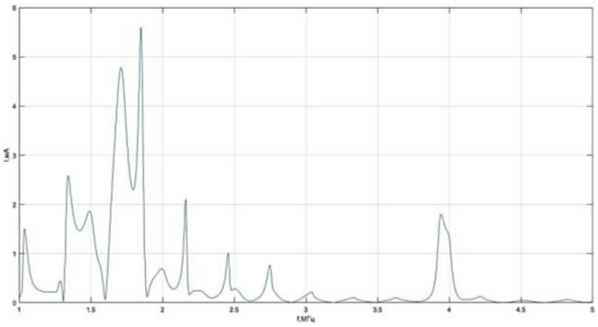

Fig. 4 – Frequency response for the case of a free rear side of the damper

The transmission line as a structure with the layers is an open type resonator in which additional losses are caused by sound radiation from the resonator volume. This radiation is due to the diffraction of sound at the finite dimensions of piezo plates and protectors, it is more noticeable the longer the wavelength (lower frequency) and increases when the faces of the plates are not parallel. The exact calculation of these effects is a rather complex and little-studied task. Therefore, in the calculations it was assumed that the bio tissue material has some equivalent additional absorption, which was considered by introducing an imaginary addition of the sound speed in the bio layer, which increases inversely to the frequency. At a frequency of 2 MHz the speed of sound in the bio layer was $1610 - i \cdot 30$ m/s (weak absorption) or $1610 - i \cdot 100$ m/s (strong absorption).

The study of the input mechanical impedance of the damper showed that at the maximum allowable thickness of the damper for design reasons, which is 4 mm, even for a highly absorbing material, it has noticeable resonant properties. Therefore, it is advisable to make the outer surface of the damper non-parallel to the surface of the piezo plate. Practically, this widely used measure reduces resonant phenomena. As a corresponding equivalent, an additional absorption was introduced into the damper material (an additional imaginary part to the speed of sound in the damper material, which was three times the imaginary part corresponding to the sound absorption in the damper material).

Fig. 5 shows the frequency characteristics of the ultrasonic line under the condition that the outer surfaces of the dampers are free, and Fig. 6 - that they are rigidly fixed. The material parameters and all dimensions are the same as previously mentioned, but an additional weak absorption has been introduced.

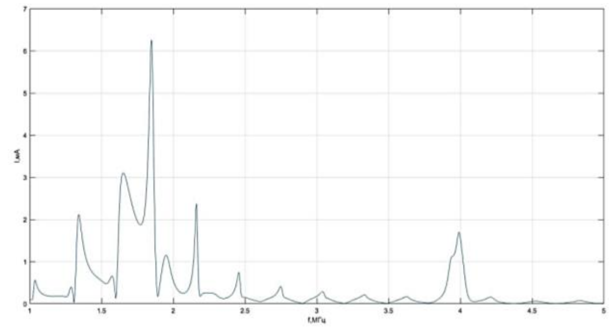

Fig. 5 – Frequency response for the case of a free rear side of the damper when weak absorption is introduced

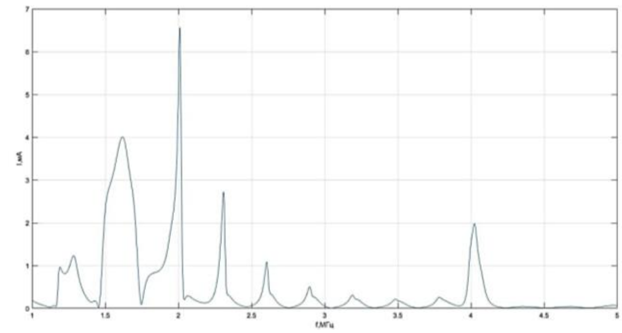

Fig. 6 – Frequency response for the case of contact of the damper with a rigid cover when a weak absorption is introduced

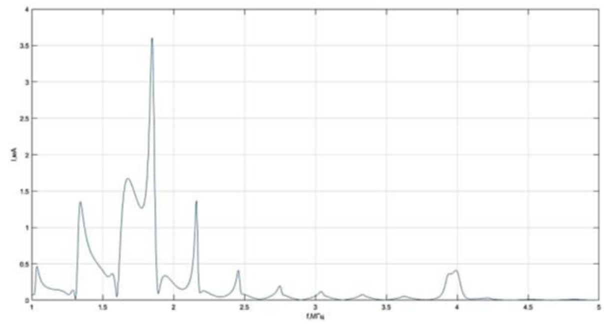

Fig. 7 – Frequency response for the case of a free rear side of the damper when a strong absorption is introduced

Fig. 7, 8 illustrate the same cases, but with strong absorption. An expansion of the bandwidth is noted with a simultaneous significant reduction in the transmission coefficient (of course, the more the greater the additional attenuation). It can also be seen that the use of dampers with rigid covers reduces the line transfer coefficient several times compared to the case when the outer surface of the damper is free. It should also be noted that the maximum current values in the load are below the required value of 1.5 mA.

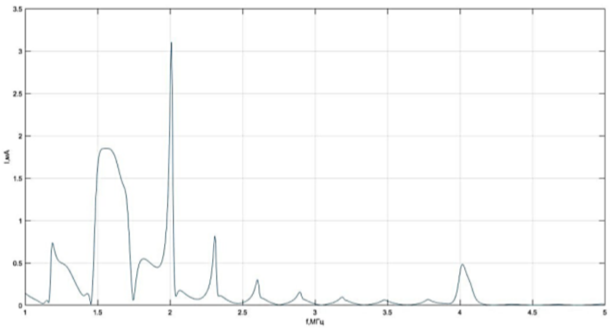

Fig. 8 – Frequency response for the case of contact of the damper with a rigid cover when a strong absorption is introduced

The graph in Fig. 9 shows the effect of changing the thickness of protectors (material – Teflon, "weak damping" of bio tissue, "light dampers" with free outer edges). Adding an imaginary part to the sound speed values considered the absorption in the protector material, which at the same time is much smaller compared to the absorption in the bio tissue and damper. The thickness of the protectors is $h_1 = 0{,}5$ mm, $h_3 = 0{,}5$ mm. Fig. 9 illustrates that an increase in the protector thickness leads to an increase in the resonance properties of the line.

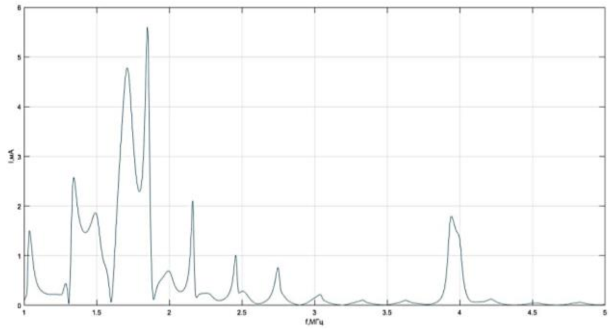

Fig. 9 – Frequency response for the case of a free rear side of the damper when introducing weak absorption, with a protector thickness of 0.5 mm

3. CONCLUSIONS

The results of the calculations can be used in the design of the structure of the ultrasonic communication line. Since in real design the choice of many parameters is determined by the requirements of technological feasibility and biological compatibility, not the absolute values of the parameters are important, but the main trends that can be observed by analyzing the results of the calculations. These trends are:

1) The outer edges of the dampers must be free and not parallel to the piezoelectric plates.

2) The choice of the protector of the receiving element is dictated by the requirements of biological compatibility. The material of the emitter protector is more free, it just shouldn't irritate the outer skin cells. The thickness of the protectors can significantly affect the frequency response of the communication line.

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Імплантат внутрішнього вуха людини з акустичним каналом передачі інформації

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В роботі розглянута електроакустична система – імплантат внутрішнього вуха людини Імплантат слугує для часткового відновлення слуху людини шляхом безпосереднього впливу на нервові закінчення завитки електричними імпульсами. Електричне подразнення слухового нерву, в свою чергу, приводить до слухових відчуттів. Обрана блок-схема імплантата, що використовує ультразвукову лінію зв'язку між зовнішньою та внутрішньою частинами системи. Проведено теоретичне дослідження впливу параметрів на частотні властивості та чутливість ультразвукової лінії передачі акустичної інформації, яке базується на математичному алгоритмі, що реалізує розрахунок чутливості каналу передачі залежно від частоти та параметрів каналу. Для розрахунків вибрано багатошарову модель лінії зв'язку, яка містить випромінюючу та приймаючу п'єзопластини, два протектори для забезпечення безпечного контакту між приймаючою-випромінюючою п'єзопластинами та біотканиною, проміжний шар біотканини, два демпфери, які прилягають до випромінювача та приймача відповідно. При цьому враховано, що демпфери мають обмежені розміри. В результаті, теоретично досліджено вплив окремих конструктивних параметрів лінії передачі на максимальну чутливість лінії та її частотну характеристику.

Ключові слова: Акустика, Електроніка, Імплантат, Електроакустичний перетворювач, Внутрішнє вухо людини, Режим випромінювання, Режим приймання, Частотна характеристика.