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REGULAR ARTICLE

Theoretical and Experimental Study of a Human Inner Ear Implant with an Ultrasonic Communication Line

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The study examines the principles of constructing a cochlear implantation system with a new acoustic communication line between its external and internal components. This system is used to restore human auditory function, which may be impaired due to illness or external factors (such as mine-blast injury). To restore hearing, the system directly stimulates nerve fibers with electrical impulses that generate auditory sensations. The developed system with implanted electrodes consists of three main components: a microphone, a sound processing device, and a series of electrodes. It utilizes the place principle, where a specific sound frequency corresponds to a particular location in the cochlea of the human inner ear. The study defines the requirements for the individual components of the cochlear implant, including the microphone, compressor, bandpass filters, amplifiers, volume regulator, modulator, ultrasonic communication line, and demodulator. The chosen sound signal encoding strategy represents speech signals in digital form, modifying both pulse width and amplitude to achieve the most energy-efficient stimulation. A critical issue discussed is the placement of electrodes in the cochlea, ensuring full coverage of the speech frequency range. The use of a ground electrode placed in the temporal muscle reduced the nerve fiber excitation threshold by half. A prototype of the human inner ear implant with an acoustic information transmission channel has been developed, and its use has confirmed the possibility of generating impulses with the required duration and current amplitude. Additionally, an important advantage of the system is the ability to implant only passive elements, significantly enhancing its reliability and safety.

Keywords: Acoustics, Electronics, Implant, Electroacoustic transducer, Human inner ear.

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1. INTRODUCTION

Modern technologies for restoring human hearing function are associated with cochlear implantation. As a result of diseases or external influences (e.g., mine-blast injuries), the part of the hearing aid that precedes the formation, transmission, and further processing of electrical impulses in the central nervous system loses its functionality. At the same time, the part of the hearing aid, starting with the afferent nerve fibers, remains intact. Then, to restore hearing, direct action on the nerve fibers with electrical impulses is used, which cause hearing sensations. A cochlear implant is used to generate such electrical impulses.

To implement the implant, a system was chosen, the block diagram of which is shown in Fig. 1. [1].

The system with electrode implantation consists of three main parts (Fig. 1): microphone (1); sound processing device (units 2-9); and a series of electrodes (10).

The principle of operation of the system is as follows. Sound vibrations are converted into electrical signals using microphone (1), which is attached by the ear hook. These signals are then sent to a signal processing device worn on the body. They are sent to a microphone amplifier (2) with an adjustable gain.

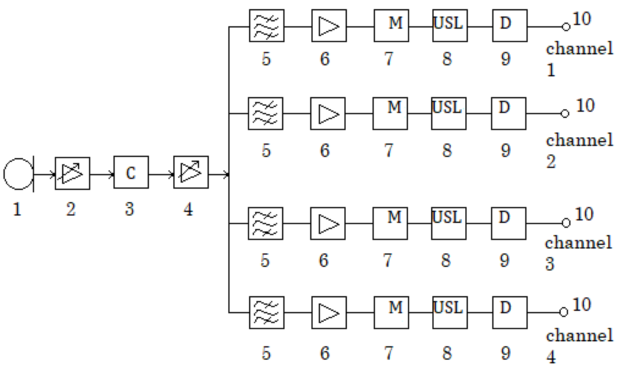

Fig. 1 – Block diagram of a human inner ear implant

Depending on the signal level at its output, the dynamic range of the signals at the output of the compressor (3) changes. The compressor compresses signals into a narrow dynamic range, which is typical for most patients. From the compressor, the signals are sent to amplifier (4), changing the gain, which affects the volume level. Then the signals are separated depending on the frequency using 4-band filters (5). After that, the signals are sent to modulator (7).

There are two versions of the system: without encoding the speech signal and with encoding [2, 3].

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In the first variant, the speech signal is modified by the modulator parameters. The carrier frequency is chosen equal to the operating frequency of the piezoelectric elements used in the acoustic communication line, namely, 2 MHz. Depending on the initial transmitted 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 is applied to the electrodes (pulse coding). Both variants differ only in the device of the modulator unit (7).

The modulated signal is returned to the ear hook and fed to the ultrasonic (US) communication line (8), the detailed calculation of which was carried out in [1]. The function of this element of the system is to transmit the signal in the cochlea, i.e., to the implanted part of the system. In the implant, the signal is sent to demodulator (9) (amplitude detector), which extracts a useful speech signal. After that, the signal is fed to the final element of the system, a series of electrodes (10), which consists of 4 platinum spherical elements. Each electrode uses a common ground located in the parietal area. This is the so-called unipolar excitation, in which the threshold is half that of bipolar excitation, when each electrode has its own ground. As a result, we can excite the electrodes with a lower current.

This system uses the principle of place, which means that a certain frequency of sound corresponds to a certain place in the cochlea of the human inner ear.

In the analog version of the system, the volume is encoded by changing the amplitude of the output current, i.e., increasing or decreasing the excitation. In the second, pulsed version, the volume is encoded by changing the charge density. Charge density is defined as the product of pulse width and amplitude. It should be noted that the dependence of the pitch on the frequency of electrical excitation makes it difficult to encode intensity.

The system uses synchronous excitation of the electrodes.

The purpose of this work is to determine the requirements for individual units of the cochlear implant and to create a layout to test its performance with a new type of communication line between the external and internal parts.

2. Calculation of the Acoustic Path of a Cochlear Implant

2.1 Microphone

Let's choose a portable piezoelectric microphone [4]. Such microphones weigh 10-12 g. The principle of their operation is that a thin duralumin diaphragm is mechanically connected to a bimorphic piezoelectric element. When the diaphragm vibrates, an electric voltage is generated on the electrodes of the piezoelectric element. The parameters of the piezoelectric microphones are as follows: piezoelectric element capacitance is 500÷1500 pF; microphone sensitivity is 50÷100 mV/Pa; frequency range is 100÷5000 Hz. The selected center frequencies of the bandpass filters are 500 Hz, 1000 Hz, 2000 Hz, 3300 Hz, and are within the frequency range of the microphone.

The average sound pressure generated by a human voice at 100 cm is 0.05 Pa. The dynamic range of speech is 25...35 dB. This corresponds to: $P_{\max} / P_{\min} = 17.8 / 56.2$ Pa.

Let's determine the maximum pressure value using the average pressure value and the peak factor value for speech. For $P_{av} = 0{,}05$ Pa, the peak factor: $\Pi = 10$ dB $= 20 \lg (P_{\max}/P_{av})$, де $P_{\max} = 0{,}16$ Pa.

The minimum value is equal to:

$$P_{\min} = \frac{P_{\max}}{56.2} \div \frac{P_{\max}}{17.8} = (2.8 \cdot 10^{-3} \div 9 \cdot 10^{-3})\ \text{Pa}$$

At the output of the microphone, with a sensitivity of 50 mV/Pa, we get an electrical signal that varies in the range: (0.14÷0.45) mV÷8 mV.

2.2 Compressor

The sensitivity is controlled by an amplifier located at the compressor input. The compressor converts the incoming signal so that low-level signals are amplified more than others. By changing the amplifier gain, the dynamic range of the sound at the compressor output can be changed from 10 dB to 20 dB in intensity. This range is typical for most patients.

2.3 Bandpass Filters

The center frequencies of the filter channels are selected for satisfactory speech perception from the range used in hearing aids: 500 Hz, 1000 Hz, 2000 Hz, 3300 Hz. Filter passbands are selected for the following reasons. It is known that based on studies of the auditory perception of acoustic noise, it was concluded that if the width of the noise band does not exceed a certain critical value, the loudness level in this band is determined only by the total noise energy and does not depend on the nature of the noise intensity distribution in this band: the intensity can be distributed in part of the band, or even concentrated in one tone. Such bands are called frequency groups. Within frequency groups, hearing integrates the excitation without considering the subtle spectral structure of the stimulus. The width of a frequency group does not depend on the noise level. However, it does depend on the average frequency as shown in Fig. 2.

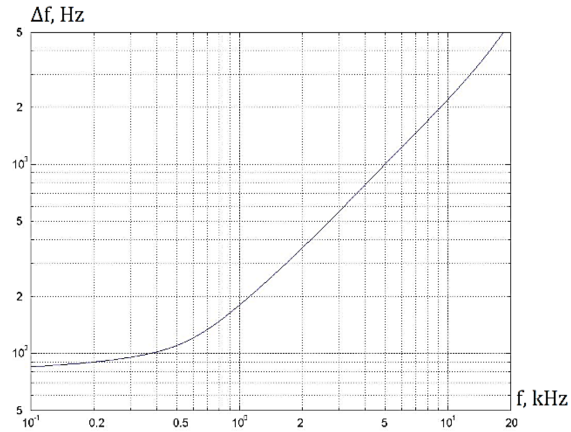

Fig. 2 – Dependence of frequency group width on average frequency

In the frequency range below 500 Hz, the width of frequency groups is almost independent of frequency and is approximately 100 Hz. In the region above 500 Hz, it increases in proportion to the frequency and is 20 % of the average frequency. The division of the sound spectrum into frequency groups is one of the important properties of hearing [5]. In the course of work on hearing implants, it was found that the width of a frequency group approximately corresponds to the effective bandwidth of primary afferent fibers. Based on this, the frequency bands of the filters are respectively equal to 100 Hz; 200 Hz; 400 Hz; 660 Hz.

2.4 Amplifiers

The amplifier gains in the channels were chosen for the following reasons. It is known that the value of the sound intensity level does not give an idea of its loudness, since it does not consider the frequency dependence of auditory perception. Therefore, to compare sounds in terms of loudness, another value is introduced, called the loudness level, which is the intensity level of a reference sound with a frequency of 1000 Hz that has the same loudness as the sound under consideration. To determine the loudness level of sinusoidal sounds, a family of equal loudness curves is used (Fig. 3), where on the right: N is the sound pressure level that has the same loudness as the pressure P (on the left) at a frequency of f = 1000 Hz. Note that the pressure P_{min}= 2×10^{-3} Pa at the lower boundary of the region at this frequency is 2 orders of magnitude greater than the sensitivity threshold defined for a harmonic signal. Since the average sound pressure generated by a person when speaking at 100 cm is 0.05 Pa, Fig. 3 shows that it corresponds to an intensity level at f = 1000 Hz equal to 50 dB.

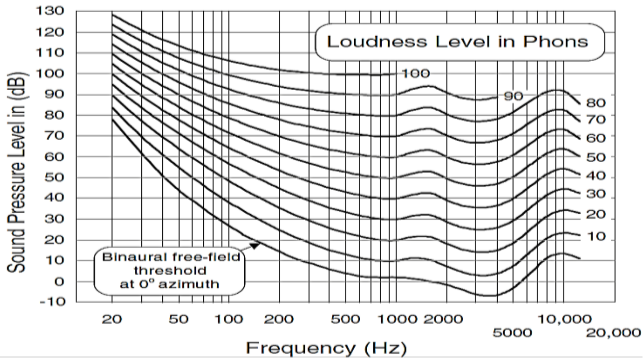

Fig. 3 – Equal volume curves [5]

The fact that both the sensitivity threshold and the pressure values on the equal volume curves are minimal in the frequency range of 3-5 kHz is due to the resonance of the air column in the external auditory canal tube when its length becomes equal to a quarter of a wavelength. The first resonance of a 27 mm long tube corresponds to approximately 3.1 kHz. The pressure on the tympanic membrane increases by 2-3 times compared to the pressure at the entrance to the ear canal. Since there is no auditory canal in the hearing prosthesis, with the same amplification in all channels, the stimulus signal at frequencies of about 3 kHz would be much less than required for undistorted speech perception.

Considering that the minimum in the 3.5 kHz region on the 50 dB curve is due to this effect, the difference value can be used to distribute the gain across the channels: $N(2\text{ kHz}) - N(2 < f < 3.5\text{ kHz})$; the maximum value of which is 5 dB. Considering that:

$$5\text{ dB} = 20\lg\left(\frac{P_1}{P_2}\right) = 20\lg K\,,$$

where $P_1 = P(2\text{ kHz})$; $P_2 = P(3.3\text{ kHz})$, we get $K_{Max} = 1.78$.

The accurate distribution of the gain across the channels could be obtained experimentally by equalizing the channel output signals from the free microphone and the microphone in the artificial ear. To be able to select the gain in each channel individually, we choose a maximum gain value of 10 dB for that channel. These amplifiers are also the decoupling outputs of the filters from the further elements.

2.5 Volume Control

The range of gain control in amplifier (4) (Fig. 1) and, accordingly, the volume is 10 dB, which is significantly less than the speech range of 20 dB. The remaining 10 dB is provided by the compressor.

2.6 Modulator and Communication Line

The developed ultrasonic (US) communication line [1] uses waves of the megahertz range to transmit both energy and information. The operating frequency of the piezoelectric elements is 2.5 MHz. The carrier frequency is equal to this frequency, i.e. $f_n = 2.5$ MHz. This provides for the most efficient transmission of ultrasonic vibrations. The principle of operation of the ultrasonic communication line is as follows: an information signal modulates the voltage of the frequency of 2.5 MHz in each of the four independent channels (Fig. 1). Each of these modulated signals is fed to its own section of the four-section piezoelectric emitter, and, having passed through the biotissue layer, is received by the corresponding section of the four-section piezoelectric receiver. The signal is then detected. The parallelism of the emitter and receiver is ensured by a system of magnets.

2.7 Demodulator

In ultrasonic echoscopy, pulse modulation of ultrasonic vibrations is used. The demodulation process is carried out by well-known methods in radio engineering circles containing active elements.

When using ultrasonic vibrations in a non-contact communication line, the presence of active elements in the hearing prosthesis implant is undesirable, and the size and materials of the passive elements must be acceptable for implantation into the tissue.

To consider possible ways of demodulation, we will use the equivalent circuit of a piezoelectric receiver in the form of a two-pole with a series connection of the EMF source and the active and reactive components of the impedance (Fig. 4, a) [6].

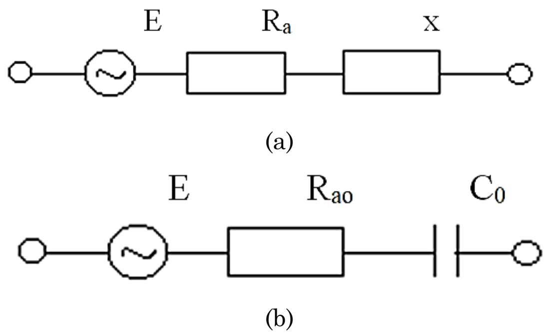

Fig. 4 – Equivalent circuit of a piezoelectric receiver in the form of a two-pole with a series connection of the EMF source, active and reactive components of the impedance (a), and an equivalent circuit at antiresonant frequency (b)

At the frequency of mechanical resonance (antiresonance), it will take the form as shown in Fig. 4, b. Due to the presence of the capacitance C_0 (static capacitance of the piezoelectric element), which does not pass the DC or slow current component in the presence of a detector connected in series with the load, it is necessary to choose a circuit with a parallel diode connection (Fig. 5, a), or connect an active resistance in parallel with the piezoelectric element and use a serial diode connection (Fig. 5, b).

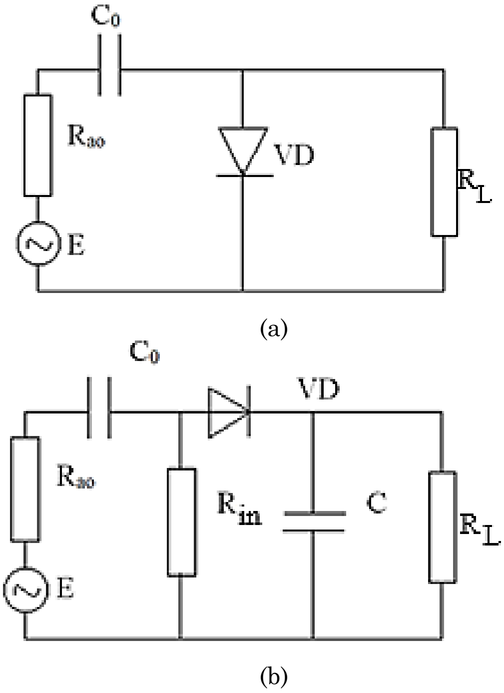

Fig. 5 – Diode connection diagram: a – parallel, b – serial

The condition for choosing a detector is:

$$R_g << R_L. \tag{1}$$

There are two possible cases of the relation between R_{ao} and C_0:

1. $R_{ao} >> 1/(\omega_0 C_0)$;
2. $R_{ao} \approx 1/(\omega_0 C_0)$.

In the first case, in the voltage half-period when the diode passes current, the current through the load resistance is small, and in the reverse half-period the current is large and equal:

$$I_m = \varepsilon / (R_a + R_L). \tag{2}$$

The current versus time dependence is as follows (Fig. 6):

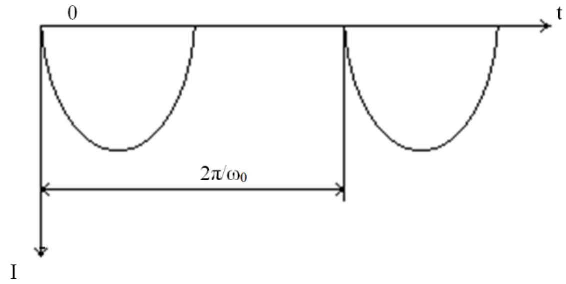

Fig. 6 – Current vs. time dependence

Let's pay attention to the facts that:

1) to obtain the maximum amplitude of the excitation current, it is necessary that:

$$R_a << R_L, \tag{3}$$

however, the value R_{ao} is a function of the piezoelectric element parameters and acoustic load, and for a damped piezoelectric element is equal to:

$$R_{ao} = \frac{4k_t^2}{\pi\omega_0 C_0 \left(k_1 + k_2\right)}, \tag{4}$$

where k_t is the electromechanical coupling coefficient; k_1, k_2 are the relative impedances (relative to the wave impedance of the piezoceramic) of the damper and the medium (bio tissue). In the presence of a matching quarter-wave layer on the receiving surface with a relative impedance k_3:

$$R_{ao} = \frac{4k_t^2}{\pi\omega_0 C_0 \left(k_1 + k_3^2 / k_2\right)}; \tag{5}$$

2) in the absence of modulation, a constant current component will flow through the load, equal to:

$$I_0 = I_m/\pi. \tag{6}$$

If the HF EMF ε and the HF current are modulated with a modulation coefficient:

$$m = \frac{I_{\max} - I_{\min}}{I_{\max} + I_{\min}} = \frac{I_{mLF}}{I_{av}} \le 1, \tag{7}$$

then:

$$I_{mLF} = mI_{av}. \tag{8}$$

The inability to eliminate the DC (average) component of the current raises the questions:

1) whether a pulsating LF current envelope will affect nerve endings in the same way as an alternating LF current with the same amplitude;

2) what effect the constant component has on nerve endings.

A negative answer to the first question and a negative impact on the second can be an insurmountable obstacle to analog information transmission.

In the case of $R_{ao} \approx 1/(\omega_0 C_0)$, the capacitance C_0 can have a filtering effect on the HF component of the detected current if the following conditions are met:

$$1/(\omega_0 C_0) << R_L. \qquad (9)$$

Choosing a diode let's take for the load resistance the value $R_L = 1$ kΩ.

Let's select a diode whose parameters meet the requirements of the model experiment, and which can be used for implantation.

It should be noted that amplitude detectors operate in receivers at low output signal voltages 30...300 mV. At the same time, the transmission coefficient of the diode detector is small: at 50 mV is 0.2; at 300 mV is 0.9.

The load resistance is chosen as low as 2-10 kΩ.

With the selected $R_L = 1$ kΩ and voltage $\varepsilon < 1$ V, the operation mode of the piezoelectric element is close to these conditions. According to the reference data [7], we see that the selected SG211 diode passes a current of 2 mA at $U_g = 0.5$ V, i.e., $R_g = 250\ \Omega << 1$ kΩ, which meets the requirement (1).

2.8 Pulse Modulation of Ultrasonic Vibrations and their Demodulation

The strategy of audio coding is to represent the speech signal in digital form.

In the case of switching to pulse coding of ultrasonic oscillations, the first issue is demodulation, since the detection scheme described below does not allow single-phase pulses to be obtained to stimulate the auditory nerves.

We note only the fundamental possibility of using such pulses. To do this, each of the ultrasonic channels of the communication line consists of two pairs of piezoelectric elements that are excited by RF pulses in such a way that the second piezoelectric element is excited after the first (i.e., pairs of elements work in turn).

The detectors in the receiving piezo plates are switched on in opposite directions, and the excitatory pulses are delivered to the auditory nerve through separate channels to two closely spaced electrodes.

Note that, unlike amplitude modulation, when the modulation coefficient $m << 1$, for pulse modulation $m = 1$, which gives a gain in the amplitude of the excitation current. The most important stimulation parameters:

1) sequential biphasic pulsatile stimulation (Fig. 7),
2) stimulation by a current source,
3) bipolar stimulation between any pair of electrodes or to a common ground, when stimulation between any active electrode and ground,
4) output current between 15 μA and 1.5 mA,
5) 239 gradations of current amplitudes,
6) change of pulse width in the range of 20-408 μs,
7) total pulse frequency is more than 1600 pulses per second.

The localization principle of frequency coding (the place principle) remains valid for pulse stimulation, and the volume is controlled by changing the charge density. Charge density is related to the product of pulse width and amplitude. In this coding strategy, both pulse width and amplitude are varied to produce the most effective energy stimulus.

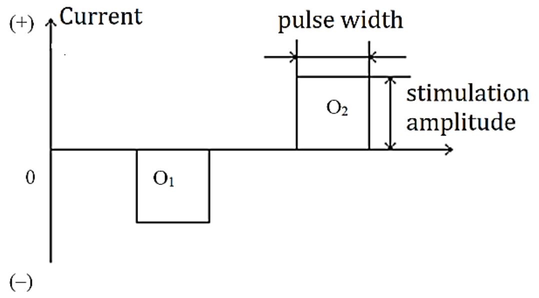

Fig. 7 – Biphasic stimulating pulse

2.9 Electrodes

Excitation electrodes should be as close to the nerve structures as possible to be able to stimulate them with minimal currents and maintain a minimum range of suprathreshold stimulation. The most suitable place for implantation of electrodes is the curl.

In humans, the cochlea is about 32 mm long, and the speech region (250 Hz to 3.5 kHz) is located at 12 mm to 29 mm from the round window.

Fig. 8 shows the scheme of the curl: 1 – partition, 2 – forecourt staircase, 3 – stirrup, 4 – drum staircase, 5 – round window membrane, 6 – helicotremes, and the location of implanted elements.

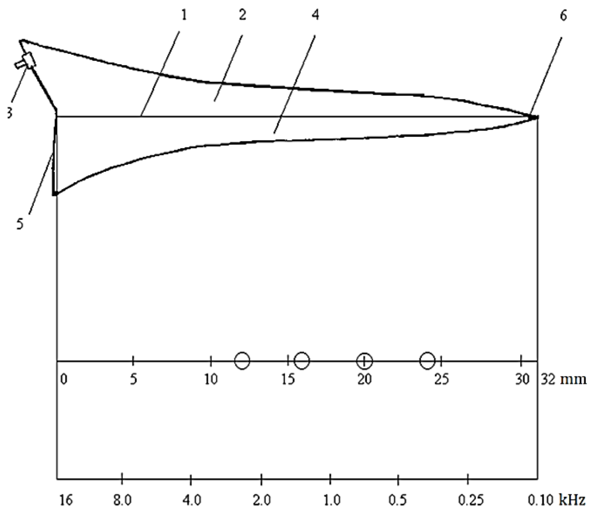

Fig. 8 – Principle of electrode arrangement

In the system under development, 4 electrodes are used, which are platinum spheres with 4 mm between them. The low-frequency electrode is located, as can be seen from Fig. 8, at 20-24 mm from the round window, and the high-frequency electrode is located at 8-12 mm. The electrodes completely cover the human speech frequency range.

Each electrode uses a ground electrode located in the temporalis muscle. As mentioned above, using this type of excitation it is possible to halve the threshold.

In work [8], monopolar excitation is used with a single electrode, and a large opposite electrode is placed in the tissue. In this case, only the local current density on the excitatory membrane of the myelinated axon is a measure of the first action potential during electrical stimulation. The spatial distribution of excitation is mainly determined by the stimulus current, distance from the electrode, and electrode shape. A simple model is usually used to calculate the characteristics of current density as a function of distance. A spherical electric electrode of radius r_k is located in the electrolyte, the opposite electrode is an infinitely distant metal sphere.

The interference from the lead wire is neglected, and the frequency of the stimulus is considered to be so low that a quasi-stationary state can be assumed:

$$\Phi = \int E dr \, , \; E = i/\rho, \qquad (10)$$

where Φ is the potential, E is the field voltage, i *is* the current density, and ρ is the specific electrical conductivity of the electrolyte.

With monopolar stimulation, when a large area of fibers is stimulated synchronously, it makes no sense to stimulate different areas of the curl at successive times, so we used simultaneous stimulation of all relevant areas of the curl.

The system uses the localization principle of frequency coding, and the intensity is coded by increasing irritation.

It should be noted that during implantation the active surfaces of the electrodes should lie as close as possible to the nerve fibers, probably closer to the transition of the basal plate into the spiral bone plate. In most of the experiments described the electrodes were inserted through a round window.

It is extremely important to point out that when inserting the electrodes, bleeding and especially tearing of the baseplate should be avoided. However, the chances of successful implantation are only when the first attempt to insert the electrodes is successful.

3. EXPERIMENTAL STUDY OF ULTRASONIC COMMUNICATION LINE AND AMPLITUDE DETECTOR OPERATION

The developed model of the human inner ear implant with an acoustic information transmission channel is shown in Fig. 9.

Fig. 9 – Appearance of the outer and inner parts of the created model of a cochlear implant with an acoustic information transmission channel

The purpose of the experiment is to confirm the possibility of obtaining pulses with a duration of 20 µs and a current amplitude of 1.5 mA on the load equivalent (at $U_{in} < 9\,V$).

First, the possibility of detecting low voltage was investigated and the value of the resistance R_{in} was chosen. The measurement scheme is shown in Fig. 10.

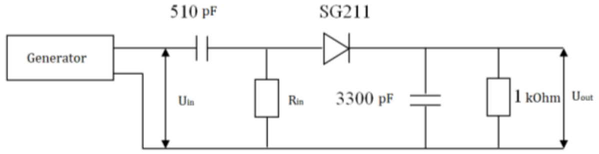

Fig. 10 – Measurement scheme of the ultrasonic communication line and amplitude detector

The voltage U_{in} was kept constant at the value of 0.6 V.

The dependence of $U_{out}\left(R_{in}\right)$ is shown in Table 1:

$$K = \frac{U_{out}}{U_{in}} \; ; \; I_L = \frac{U_{out}}{10^3 \text{Ohm}} \; .$$

Table 1 – Experimental data when connecting the generator

U_{in}, V	0.6	0.6	0.6
R_{in}, Ω	100	1000	10 000
U_{out}, V	0.52	0.48	0.1
K	0.87	0.8	0.17
I_L, µA	520	480	100

Thus, the optimal value of R_{in} is 100 Ω with a detector transmission coefficient of 0.87.

A real single channel of the ultrasonic line is connected instead of the generator. Its parameters are as follows: piezoceramic material is PZT-8; piezoelectric element area is $151{,}71 \cdot 10^{-6}$ m²; load is epoxy resin on the back side; Teflon layer is on the working side. The measurement scheme is shown on Fig. 11. The results are summarized in Table 2, and the dependence of the load current on the input voltage is shown in Fig. 12.

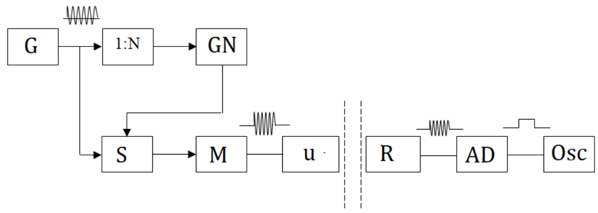

Fig. 11 – Measurement scheme of an ultrasonic communication line and amplitude detector

Table 2 – Experimental data when connecting an ultrasonic line

U_{in}, V	1.7	1.7	1.7	1.7	1.7	1.7
R_{in}, Ω	6800	100	330	1000	1000	1000
U_{out}, V	60	150	250	200	375	600
K	0.035	0.088	0.147	0.120	0.134	0.15
I_L, µA	60	150	250	200	375	600

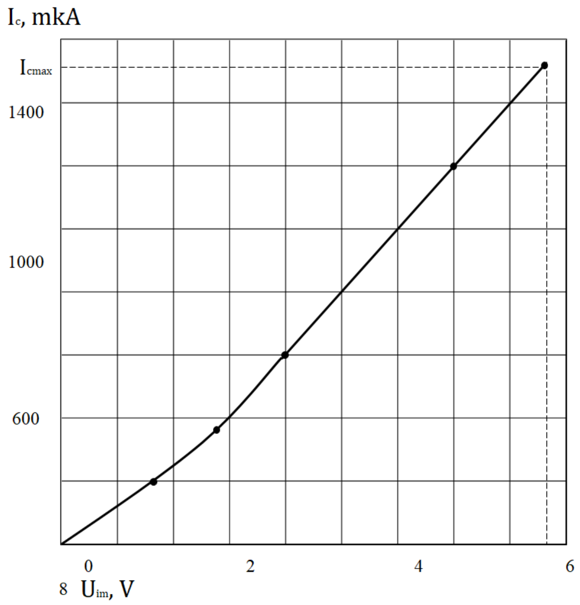

Fig. 12 – Dependence of load current on input voltage

The duration of the pulse detection varied from 20 μs to 200 μs, with a pulse shape close to rectangular.

4. CONCLUSIONS

The studied ultrasonic communication line with an amplitude detector made it possible to obtain current pulses with a lower width limit of 20 μs on the equivalent of a real load. This corresponds to the required minimum width of the current pulse. The maximum permissible current pulse amplitude is 1.5 mA and is achieved at $U_{in} = 8.6$ V The minimum current pulse amplitude of 5 μA corresponds to $U_{in} = 0.3$ V. Thus, the developed communication line allows for providing the required range of current amplitude gradations from 5 to 1500 μA and the range of pulse durations from 20 to 408 μs. Its additional advantage is the ability to implant only passive elements.

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Теоретичне та експериментальне дослідження імплантату внутрішнього вуха людини з ультразвуковою лінією зв'язку

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В роботі розглянуті принципи побудови системи кохлеарної імплантації з новою акустичною лінією зв'язку між її зовнішньою і внутрішньою частинами. Така система застосовується для відновлення слухової функції людини, яка порушена внаслідок захворювання або зовнішнього впливу (мінно-вибухової травми). Для відновлення слуху система безпосередньо діє на нервові волокна електричними імпульсами, які і викликають слухові відчуття. Розроблена система з імплантацією електродів складається з трьох основних частин: мікрофону; пристрою обробки звуку; та ряду електродів, і використовує принцип місця, який полягає в тому, що певній частоті звуку відповідає певне місце в завитці внутрішнього вуха людини. Визначено вимоги до окремих блоків кохлеарного імплантату: мікрофону, компресора, смугових фільтрів, підсилювачів, регулятора гучності, модулятора, ультразвукової лінії зв'язку, демодулятора. Вибрана стратегія кодування звукових сигналів в системі, яка полягає у представленні мовного сигналу у цифровій формі. Вона передбачає зміну як ширини імпульсу, так і його амплітуди, для отримання найефективнішого енергетично стимулу. Обговорено важливе питання розміщення електродів у завитці внутрішнього вуха людини, які в розробленій системі повністю забезпечують мовний діапазон частот. Використання земельного електроду, розташованого в скроневому м'язі, дозволило вдвічі знизити поріг збудження нервових волокон. Створено макет імплантату внутрішнього вуха людини з акустичним каналом передачі інформації, і з його використанням підтверджена можливість отримання імпульсів з необхідною тривалістю та амплітудою струму. Зазначено, що важливою додатковою перевагою системи є можливість імплантації лише пасивних елементів, що значно підвищує її надійність та безпечність.

Ключові слова: Акустика, Електроніка, Імплантат, Електроакустичний перетворювач, Внутрішнє вухо людини.