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Theoretical model of intracranial media ultrasonic attenuation measurement method

V. Petkus[1], A. Ragauskas[1], R. Jurkonis[2]

[1]*Telematics Scientific Lab., Kaunas University of Technology.*

[2]*Ultrasound Biomedical Engineering Lab., Kaunas University of Technology.*

Introduction

The state of intracranial media (IM) and an intracranial pressure (ICP) as its characteristic depends on the relations of the intracranial components' volumes and their changes. [1,2]. Intracranial components (brain tissue, blood, cerebrospinal fluid (CSF)) are characterized by different acoustic parameters, such as ultrasound speed and frequency dependent attenuation. We have shown in our previous work that the volume changes of these components can be measured by ultrasonic time-of-flight method [3]. In this work [3] we have shown, that by applying simultaneous measurement of the signal's time-of-flight through IM and the ultrasound attenuation in IM, it is possible to obtain information about the character of the physiological phenomena that occur in IM. For example, in the cases of cerebral vasodilatation and swelling phenomena that cause the increase of ICP, the decrease of signal's time-of-flight through IM is observed. Therefore, the changes of attenuation in IM are of the opposite signs, i. e. in the case of the swelling phenomenon, the attenuation increases, and in the case of the vasodilatation phenomenon the attenuation decreases [3]. Besides, various reactions of time-of-flight and attenuation can be observed during the occurrence of physiological phenomena, by choosing different acoustic paths through IM. Additional information about the ultrasound attenuation in IM is supposed to allow choosing more optimal measurement acoustic path and give more exact interpretation of the physiological phenomena measured.

The aim of this study is to create a mathematical model of ultrasonic attenuation measurement in the intracranial media that could be realized by the non-invasive ultrasonic time-of-flight methods and to perform analysis of possible components of the uncertainty of measurement.

Theoretical model

Many of the ultrasound attenuation measurement techniques are based on the fact that the attenuation coefficient in biological tissues is frequency dependent and can be expressed by power low function: [4,5,6]:

$$\alpha(f)=\alpha_0 f^{\,n}, \qquad (1)$$

where α_0 and n are tissue dependent attenuation parameters (α_0 is attenuation in *Np*/(cmMHz^n)), f is frequency in MHz. It can be seen from the equation, that ultrasound attenuation in higher frequencies is higher than in lower frequencies. It results in the decrease of signal spectrum bandwidth, when it travels through the lossy medium. Ophir and Jaeger (1982) show that the downshifted center frequency and decreased variance of the spectrum for a Gaussian pulse can be expressed in the following equations [5,6,7]:

$$\Delta f = f_0 - f_c = 2nd\sigma^2\alpha_0 f_c^{\,n-1}, \qquad (2)$$

$$\sigma_c^2 = \frac{\sigma_0^2}{1+2n(n-1)f_0^{\,n-2}\sigma_0^2\alpha_0 d}, \qquad (3)$$

where f_0 and σ^2_0 are center frequency and spectrum variance of the input signal, f_c ir σ^2_c are center frequency and spectrum variance after the signal passes through the lossy medium, *d* is the total propagation distance. The spectrum variance characterizes a bandwidth of Gaussian pulse and can be obtained from the relation B=2.36σ [8], were B is the 6-dB bandwidth calculated from the signal power spectrum.

One of the most important factors, which influence the possibilities of the non-invasive measurement of IM attenuation, is a skull bone that is characterized by a high ultrasound attenuation. It results in a high downshift of ultrasonic pulse signal center frequency and high losses of signal energy, when it travels through the skull bones. Therefore, additional measurement errors can be caused by hemodynamics and swelling of human head external tissues, which aren't related to the attenuation and the state of IM. The proposed method for the compensation of the influence of the external tissues and skull bones for non-invasive measurements of the ultrasound attenuation in IM

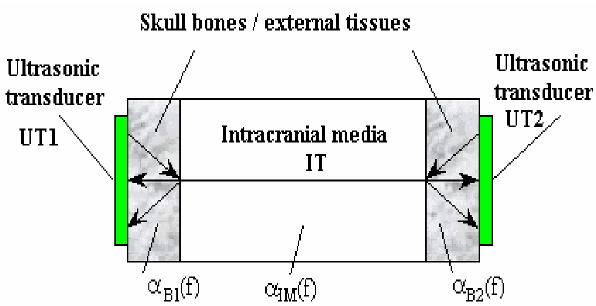


Fig.1. **The diagram of ultrasound attenuation measurement in intracranial media**

is shown in Fig.1.

The total ultrasound attenuation in IM, skull bones and external tissues is defined from the changes of ultrasound signal's parameters when it passes from one ultrasonic transducer to the other. The attenuation in skull bones and external tissues is determined from the parameters of the signals, which reflect from inner wall of the left and right side of the skull bones.

Two methods, differential and direct, were proposed by solving a problem, to determine the ultrasound attenuation in IM and compensate the influence of the skull bones and external tissues (SB/ET) according to the measured parameters of these signals.

1. Differential method. The total attenuation in IM and SB/ET measured from the left $\alpha_{\Sigma1}$ and right $\alpha_{\Sigma2}$ sides can be expressed:

$$\alpha_{\Sigma1}(f)=\alpha_{\Sigma2}(f)=\alpha_{B1}(f)+\alpha_{IM}(f)+\alpha_{B2}(f), \text{[dB]} \quad (4)$$

where α_{B1} and α_{B2} are attenuations in SB/ET measured in the left and right sides, α_{IM} is attenuation in IM. The attenuation in IM α_{IM} after the compensation of the influence of SB/ET, is defined as the difference between the total attenuation in IM and SB/ET and the attenuation in SB/E:

$$\alpha_{IM}(f)=\frac{\alpha_{\Sigma1}(f)+\alpha_{\Sigma2}(f)}{2}-\left(\alpha_{B1}(f)+\alpha_{B2}(f)\right), \quad (5)$$

A attenuation in IM can be expressed from (1) and (5) as a function of thicknesses of medium layers and frequency:

$$\alpha_{IM1}(f)=\alpha_{0\_IM1}d_{IM}f^{n_{IM}}=\alpha_{0\_\Sigma1}d_{\Sigma}f^{n_{\Sigma}}-\alpha_{0\_B1}2d_{B1}f^{n_{B1}},$$

$$\alpha_{IM2}(f)=\alpha_{0\_IM2}d_{IM}f^{n_{IM}}=\alpha_{0\_\Sigma2}d_{\Sigma}f^{n_{\Sigma}}-\alpha_{0\_B2}2d_{B2}f^{n_{B2}},$$

$$\alpha_{IM}(f)=\frac{\alpha_{IM1}(f)+\alpha_{IM2}(f)}{2}, \quad (6\text{-}8)$$

where the parameters with indices 1 and 2 mean that they are calculated according to the measured results obtained by performing head radiation from the opposite sides, α_{0_IM}, n_{IM}; α_{0_Σ}, n_{Σ}; α_{0_B1}, n_{B1} and α_{0_B2}, n_{B2} are attenuation parameters of IM, total media of IM and SB/ET and the media of SB/ET, $d_{\Sigma} = d_{IM} + d_{B1} + d_{B2}$ is the total propagation path of the ultrasonic signal that is equal to the sum of acoustic paths' lengths in intracranial media and in skull bones on the left and right sides of the human head.

By simplifying the measurement method it was proposed to calculate the attenuation at the frequency f=1MHz, and to measure the signals' periods instead of the center frequencies $f_0=1/T_0$. Then, performing these changes and from eqn (2), (6), (7), (8), final equations for calculating attenuation in IM can be expressed:

$$\alpha_{IM1}(1MHz)=\frac{\left(1/T_{01}-1/T_{\Sigma1}\right)}{2n_{\Sigma}\sigma_{01}^2T_{\Sigma1}^{1-n_{\Sigma}}}-\frac{\left(1/T_{01}-1/T_{B1}\right)}{2n_B\sigma_{01}^2T_{B1}^{1-n_B}}, \quad (9)$$

$$\alpha_{IM2}(1MHz)=\frac{\left(1/T_{02}-1/T_{\Sigma2}\right)}{2n_{\Sigma}\sigma_{02}^2T_{\Sigma2}^{1-n_{\Sigma}}}-\frac{\left(1/T_{02}-1/T_{B2}\right)}{2n_B\sigma_{02}^2T_{B2}^{1-n_B}}. \quad (10)$$

Where T_{01}, T_{02}, $T_{\Sigma1}$, $T_{\Sigma2}$, T_{B1} and T_{B2} are the periods of input signals, the signals passed through a human head and echo signals performing the measurements from the left and right sides, σ_{01}^2, σ_{02}^2 are spectrum variances of the left and right sides of the input signals. The attenuation parameters n for the total media and for SB/ET were assumed be equal among themselves $n_B=n_{B1}=n_{B2}$ and $n_{\Sigma}=n_{\Sigma1}=n_{\Sigma2}$.

2. Direct method. The attenuation in IM is calculated by referring to eqn (2) from the downshift of the output signal center frequency with respect to the center frequency of the echo signal:

$$\alpha_{IM1}(f)=\alpha_{0\_IM1}d_{IM}f^{n_{IM}}=\frac{\left(f_{B1}-f_{\Sigma1}\right)f^{n_{IM}}}{2n_{IM}\sigma_{B1}^2f_{B1}^{n_{IM}-1}}, \quad (11)$$

$$\alpha_{IM2}(f)=\alpha_{0\_IM2}d_{IM}f^{n_{IM}}=\frac{\left(f_{B2}-f_{\Sigma2}\right)f^{n_{IM}}}{2n_{IM}\sigma_{B2}^2f_{B2}^{n_{IM}-1}}. \quad (12)$$

The spectrum variances of the echo signals can be expressed from the spectrum variances of input signals and the downshifts of the center frequencies of the echo signals Δf from eqn (1) and (2):

$$\sigma_B^2=\frac{\sigma_0^2}{1+(n_B-1)\Delta f\,\dfrac{f_0^{\,n_B-2}}{f_B^{\,n_B-1}}}. \quad (13)$$

Then, rewriting the equations respectively to the frequency f=1MHz and performing the changes from the center frequency into the period, from eqn (11-13) we obtain:

$$\alpha_{IM1}(1MHz)=\frac{\left(1/T_{B1}-1/T_{\Sigma1}\right)}{2n_{IM}\sigma_{01}^2T_{B1}^{1-n_{IM}}}\times$$

$$\times\left(1+\left(n_B-1\right)\left(\frac{1}{T_{01}}-\frac{1}{T_{B1}}\right)\frac{T_{01}^{\,2-n_B}}{T_{B1}^{\,1-n_B}}\right), \quad (14)$$

$$\alpha_{IM2}(1MHz)=\frac{\left(1/T_{B2}-1/T_{\Sigma2}\right)}{2n_{IM}\sigma_{02}^2T_{B2}^{1-n_{IM}}}\times$$

$$\times\left(1+\left(n_B-1\right)\left(\frac{1}{T_{02}}-\frac{1}{T_{B2}}\right)\frac{T_{02}^{\,2-n_B}}{T_{B2}^{\,1-n_B}}\right). \quad (15)$$

These equations for the calculation of attenuation in IM were derived by taking into account the possibility of the method realization using the ultrasonic time-of-flight measurement technique that is capable of measuring small deviations of signals' periods. This method also could be realized by using spectral methods, the implementation of which could allow to define attenuation dependence in a wide range of frequency. However, the problems can be encountered by performing the measurement of the echo signal spectrum. The spectrum of the echo signal can be distorted by superposition of a few existing echo signals (echoes from bone, dura and brain surface).

The main factors that determine the errors of the proposed mathematical attenuation measurement models are:

1) The model is derived assuming that the ultrasonic signal is an ideal Gaussian signal. Meanwhile, the spectrum of the real signals isn't ideal Gaussian and is distorted by transducer reverberations, and radial oscillations. Also, additional distortion of a signal spectrum can be caused by filters, amplifiers and ultrasonic transducers that can limit the bandwidth of the signal.

2) Diffraction errors. A diffraction effect is caused by the finite aperture of the transducer, and results in undesirable changes of the ultrasonic signal spectrum. Especially, it must be taken into account while performing

measurements in a near field of transducer radiation zone (in our case, by measuring the echo signal period). The diffraction correction method has been defined for reducing these errors and it has been used in our study.

3) The distortion of a signal waveform caused by the frequency dependent dispersion phenomenon. The anomalous dispersion phenomenon is observed for the ultrasonic signal when it travels through the lossy medium ($1 \le n < 2$), i.e. higher frequency components of ultrasound pulse spectrum travel at higher phase speeds than lower frequency components and it causes the modulation of a signal period. The question is which period or half-period must be measured for the calculation of attenuation in IM?

4) While performing the calculation, the attenuation parameters n of the media were taken from references [9] ($n_B \approx 2$, $n_{IM} \approx 1.1$) and it is clear that for real measurements these coefficients can be different. Besides, performing the calculation of attenuation in IM by a differential method, the attenuation parameter n_Σ of the total IM and SB/ET media must be evaluated a priori, which is a function dependent on the attenuations in intracranial media and skull bones:

$$n_\Sigma = \frac{\lg\left(\dfrac{\alpha_{0\_IM} d_{IM} f^{n_{IM}} + (\alpha_{0\_B1} d_{B1} + \alpha_{0\_B2} d_{B2}) f^{n_B}}{\alpha_{0\_IM} d_{IM} + \alpha_{0\_B1} d_{B1} + \alpha_{0\_B2} d_{B2}}\right)}{\lg f}. \quad (16)$$

Mathematical simulation and results

Mathematical calculations were performed for the determination of the influence of the factors mentioned above on the accuracy of the attenuation measurement in IM. The simulation of ultrasound pulse propagation through the human head was performed and the parameters of the input, output and echo signals, that must be used for attenuation in IM calculation, were determined. For the evaluation of the influence of the skull bone attenuation, the attenuation parameter α_{0_B} was changed in the range 1…11dB/(cmMHz^n) (the values of α_{0_B}, presented in [8,9], are in the ranges 5…11 dB/(cmMHz^n).The attenuation parameters of IM were determined assuming, that IM consists of 1150ml of brain tissue, 75ml of CSF, 75ml of blood and the length of the acoustic path is 15cm. The acoustic parameters of the media used for cranial simulation are obtained from the references [8,9,10] and from the results obtained by performing the sound speed in blood and CSF measurements for various groups of patients. These measurements were performed and the results were kindly given by Dr. R. Šliteris and Dr. B.Voleišiene (Ultrasound Institute, KTU).

Table 1. **Acoustic parameters and thicknesses of the media used for cranial simulation**

Medium	Attenuation parameters		Speed of sound *c*, m/s	Layer thickness L,cm
	α_0, dB/(cmMHz^n)	n		
Skull bone	1…11	1.89	2652	0.8 (each side)
IM	0.7813	1.0747	1562	15
Brain tissue	0.8692	1.078	1563	13.27
CSF	0.0023	1.9937	1533	0.865
Blood	0.212	1.2662	1583	0.865

While defining the influence of attenuation, sound dispersion and diffraction on the results of attenuation calculation in IM, the simulation of ultrasound pulse propagation through the human head was performed by 3 different methods:

1. One-dimensional (1D) simulation method that includes only the signal attenuation and the center frequency downshift effects. The output signal is obtained multiplying an input signal spectrum by the medium attenuation function (1). According to these simplified calculations, the equations (2,3) were derived by J.Ophir (1982) and have been used in attenuation measurement [5,6,7].

2. One-dimensional (1D) simulation method that includes signal attenuation, dispersion, and center frequency downshift effects. The method is based on the signal spectrum decomposition principle i.e. the input signal spectrum is decomposed into narrowband components, and for each component a group delay, phase shift and attenuation are calculated separately. This method was derived by Ping He [4] and was used in our previous work [3].

3. Our three-dimensional simulation method (3D) that includes all basic effects, which influence the simulation results, i.e. signal attenuation, downshift of center frequency, dispersion and diffraction of ultrasound beam. This is a combined method that consists of dispersion (according to Ping He [4]) and diffraction (according to Khimunin [11]) calculation methods. Diffraction effects were calculated for both output and echo signals, under condition that the transducer plane surface is parallel to the skull bones' plane, and the edge wave passes the boundary bone-intracranial media corresponding to the Snell's law.

The attenuation in IM was calculated by differential and direct models, according to the parameters of echo and output signals that were simulated by different methods, described above. The calculation results and the influence of the attenuation in skull bones are shown in Fig 2, where the curves presenting the attenuation in IM are shown as a function dependent on the attenuation in skull bones. The true value of the ultrasound attenuation in IM (α_{IM}(1MHz)=11.72dB) was determined from the parameters presented in Table1 and wasn't changed during the simulation (a straight line in Fig 2.).The calculations of attenuation in IM using a an ideal Gaussian input signal are shown in Fig 2a and 2b, and those using the real Gaussian input signal are shown in Fig. 2c and 2d. The real Gaussian input signal was obtained *in vivo* using PVDF piezofilms, and the waveform and spectrum of the signal are shown in Fig.3. The ideal Gaussian signal, used in simulations, was calculated according to the parameters of the real Gaussian signal: center frequency f_0=1.86MHz and spectrum variance σ_0^2=0.604 (MHz)^2 (Fig.3.).

It can be seen from Fig.2, that the best results of the calculation of attenuation in IM were obtained by performing a simulation of the ideal Gaussian signal's propagation without the dispersion and diffraction effects (Fig. 2a and 2b curves with asterisk). However, a large disagreement between the calculated curves and the true attenuation in IM is observed, by including the ultrasonic signal's distortion caused by the diffraction effects (Fig. 2

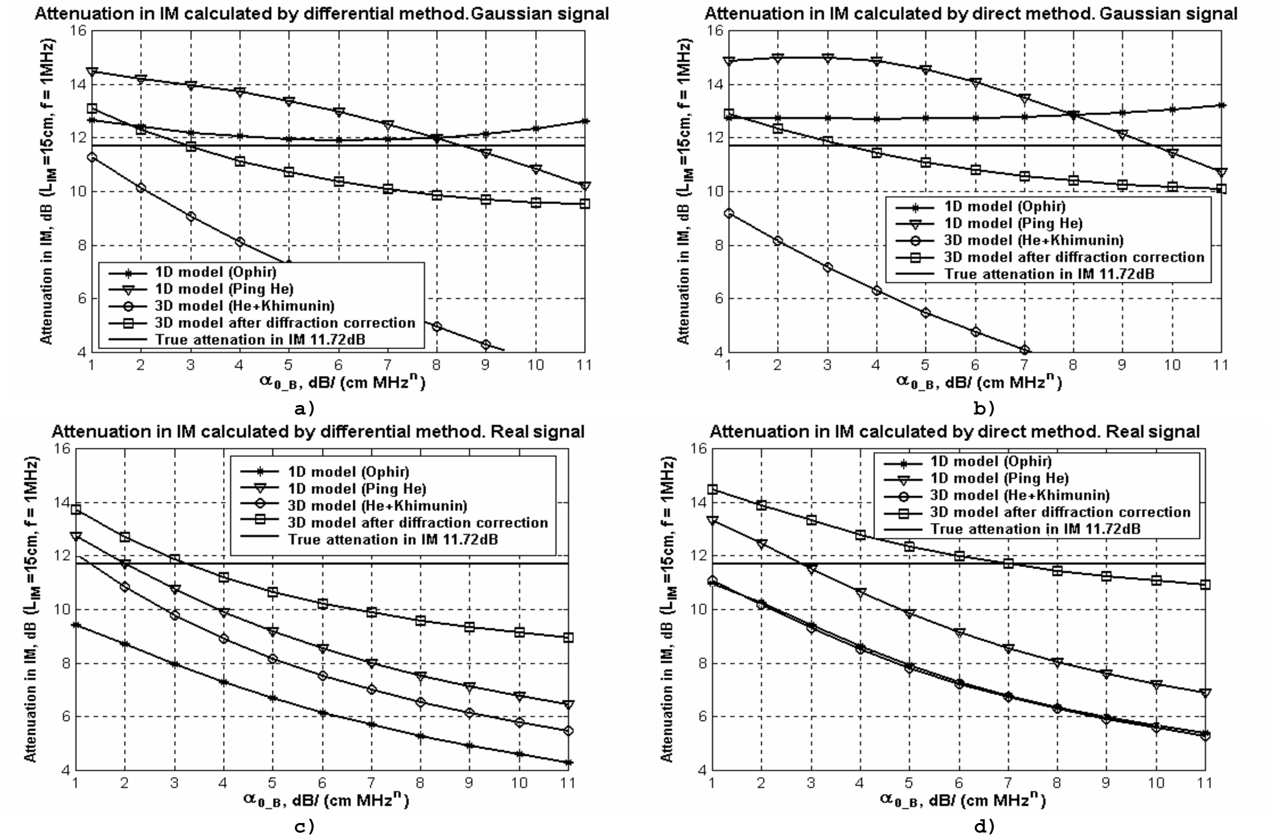


Fig.2. The influence of the skull bones attenuation on the calculation of the attenuation in IM. (a,c) - attenuation calculated by differential method, (b,d) - attenuation calculated by direct method. The calculation performed using an ideal Gaussian signal (a,b) and using a real Gaussian signal (c,d). The true value of the ultrasound attenuation in IM is 11.72dB and wasn't changed during the simulation).

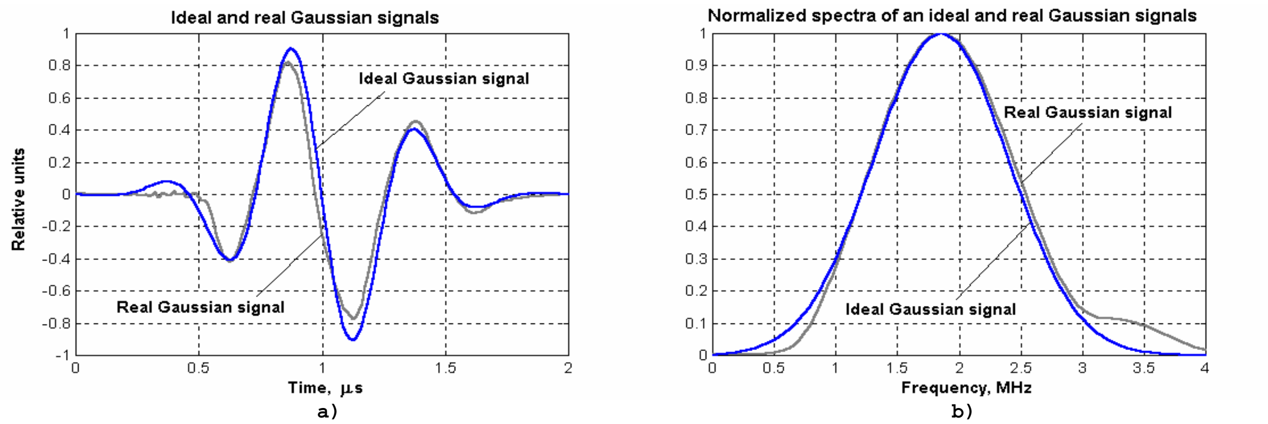


Fig.3. The ultrasonic ideal and real Gaussian signals used for simulation (a) and their spectra (b). The ideal Gaussian signal was calculated according to the parameters of the real signal: center frequency f_0=1.86MHz and spectrum variance σ_0^2=0.604(MHz)^2.

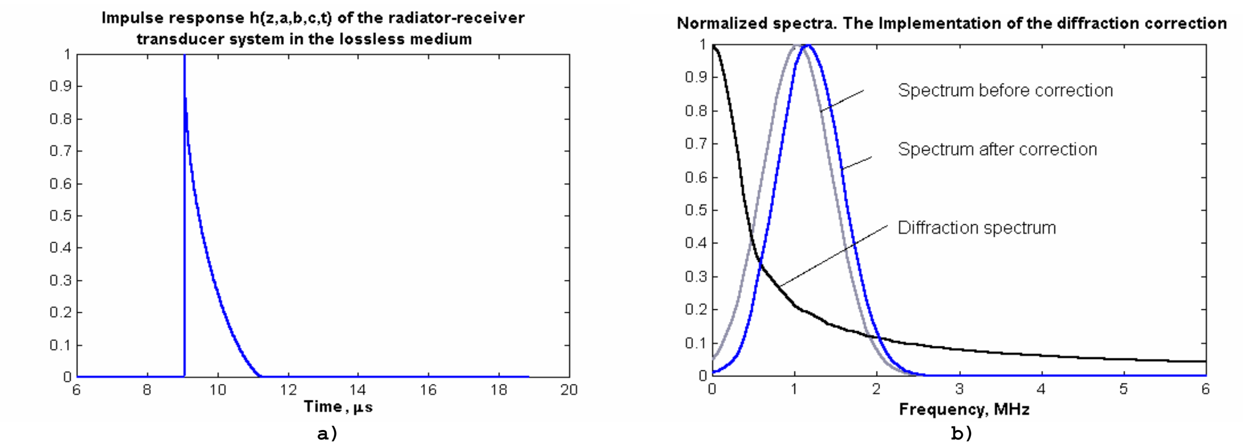


Fig.4. The impulse response h(z,a,b,c,t) of the radiator-receiver transducer system in the lossless medium (a), and spectrum of this response (diffraction spectrum) in (b). The signal spectrum before and after the diffraction correction in (b).

curves with circles). For the correction of these errors, the Inverse Diffraction Filtering (IDF) method was applied, described by J. Ophir (1990) [12]. The method is based on the assumption that the transducer diffraction and tissue attenuation affect the ultrasonic signal spectra independently and then diffraction-free estimates of the signal spectrum can be obtained be eliminating the diffraction component in the signal spectrum using inverse diffraction filtering [12]:

$$S_{out}(j\omega) = \frac{S'_{out}(j\omega)}{X(j\omega)} = S_{in}(j\omega) * H(j\omega), \qquad (17)$$

where $S'_{out}(j\omega)$ is the spectrum of the signal which passed through the lossy medium and is distorted by diffraction effects, $S_{in}(j\omega)$ is the spectrum of the input signal, $H(j\omega)$ is a transfer function of the medium that includes only the attenuation effects, $X(j\omega)$ is the diffraction spectrum, obtained by performing measurements in lossless medium, $S_{out}(j\omega)$ is the spectrum of the signal which passed through the lossy medium, after diffraction correction.

This IDF method was modified and applied for the correction of echo signals diffraction effects caused by the finite aperture of the ultrasonic transducers. This correction is implemented in the following steps:

1. Measurement of the echo signal period T_B, input signal period T_0 and input signal spectral variance σ_0^2.

2. Calculation of the echo signal spectrum variance σ_B^2, according to the changes of the echo signal's period with respect to the input signal's period:

$$\sigma_B^2 = \frac{\sigma_0^2}{1 + (n_B - 1)\left(\frac{1}{T_0} - \frac{1}{T_B}\right)\frac{T_0^{\,2-n_B}}{T_B^{\,1-n_B}}}. \qquad (18)$$

3. Calculation of the Gaussian spectrum of the echo signal, according to the obtained values of the σ_B^2, T_B:

$$S(f) = \exp\left(-\frac{(f - 1/T_B)^2}{2\sigma_B^2}\right). \qquad (19)$$

4. Calculation of the impulse response $h(z,a,b,c,t)$ of the radiator-receiver transducer system in the lossless medium according to the method derived by Khimunin (1984) [11]. The data that must be known for calculation are: z is a total propagation distance in skull bone of the echo signal, a is the radiator radius, b is the receiver radius (in our cases this is the same transducer working as a radiator-receiver $a=b$), c is a sound speed in the skull bone.

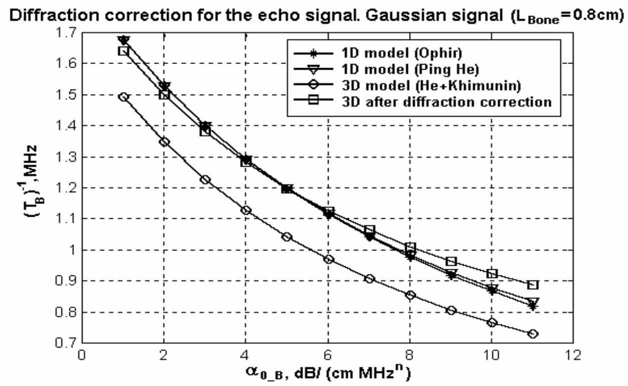
5. Calculation of the signal spectrum after diffraction correction:

$$S_{corrected}(f) = \frac{S(f)}{FFT\left[h(z,a,b,c,t)\right]}. \qquad (20)$$

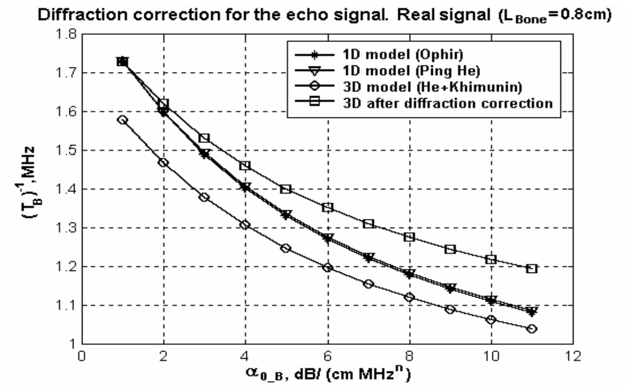
6. Determination of corrected signal period or center frequency.

The calculated impulse response that defines the diffraction effects is shown in Fig. 4a. It was calculated for the echo signal under the condition that the thickness of the skull bone is 0.8cm, the transducer radius is 0.75cm, the speed of sound in a skull bone is c_B=2652m/s and medium attenuation is α=0. The spectrum of this impulse response (diffraction spectrum) is shown in Fig. 4b. The echo signal spectra before and after diffraction correction are also shown in Fig. 4b (the presented echo signal spectra were calculated under condition α_{0_B}=11dB/(cmMHz^n)).

The inverse period ($1/T_K$) values of the echo signal that were obtained during the simulation are shown in Fig.5. The curves with the circles show what deviation of the echo signal inverse period can be caused by diffraction effects. The curves with the squares show the values of the echo signals inverse periods after diffraction correction. The values of inverse periods presented in the figures are obtained using the ideal Gaussian input signal in Fig.5a and using the real Gaussian input signal in Fig.5b. The calculation results of the ultrasound attenuation in IM after performing the diffraction correction for the echo signal are shown in Fig. 2 (curves with squares).



a)



b)

Fig.5. **The values of inverse periods of the echo signals used for attenuation calculation before and after diffraction correction. The calculation performed using an ideal Gaussian signal (a) and a real Gaussian signal (b) .**

This method corrects only the diffraction distortions caused by finite aperture of the transducers [12]. The calculations were performed under the assumption that the echo signal reflects from the skull bones surface plane that is parallel to the transducer surface plane. However, the diffraction distortions that can be caused by a "curved" plane of skull bones and the distortions caused by inhomogeneous of the medium aren't corrected [12].

Discussion

It was found that the main factors that restrict the accuracy of the presented attenuation measurement in IM methods are:

a) The diffraction errors caused by the finite aperture of the transducer. This effect mainly affects the echo signal that is in the near field of the transducer radiation zone. While creating the diffraction correction algorithm it was assumed that signal diffraction and medium attenuation independently affect the spectrum of the ultrasonic signal and that is why the diffraction-free estimates of the signal spectrum can be obtained by eliminating the diffraction component in the frequency domain [12,13].

b) The influence of the frequency dependent dispersion effect. This effect distorts the ultrasonic signal when it travels through the lossy medium ($1 \le n < 2$), i.e. the modulation of the inner period for the Gaussian ultrasonic signal is observed. By implementing the presented attenuation methods with the time-of-flight measurement technique, it is important to know which period or half-period of the ultrasonic signal must be measured. During the presented simulations, the first period of the ultrasonic signal was used for the attenuation in IM calculation, but the right choice of the period or half-period is expected to depend on the waveform of the ultrasonic signals used and on the attenuation characteristics of the media as well.

c) The influence of the non-ideal waveform of the ultrasonic signal. The ultrasonic attenuation in IM models are derived making an assumption that the ultrasonic signal is an ideal Gaussian signal. Meanwhile, the spectra of the real signals are distorted by real transducer's transfer function. For example, in the case of highly attenuating medium a high downshift of signal spectrum will be observed, therefore, all filters (including ultrasonic transducers) that limit signal spectrum in low frequencies will impede the shift of a signal spectrum into low frequencies.

It is seen from the attenuation in IM calculation results (Fig. 2 and Fig. 6) that the errors caused by the influence of all investigated factors (the waveform of the ultrasonic signal, diffraction and dispersion effects) have a systematical character, therefore, they can be corrected by performing a computer simulation of the ultrasound signal propagation through the human head.

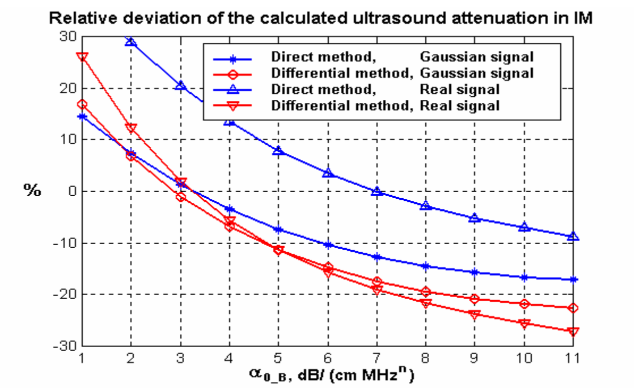


Fig.6. **Relative deviation of the attenuation in IM obtained by calculating attenuation from the parameters of ultrasonic signals that were simulated in attenuating, dispersive media with the diffraction losses. The diffraction correction was performed for echo signal before calculating the attenuation in IM**

Conclusions:

1. The ultrasound attenuation in human intracranial media can be determined from the periods of the output (passed through the human head) and echo (reflected from inner surface of skull bones) Gaussian ultrasonic signals, measured by the non-invasive time-of-flight technique. The applied theoretical model and computer simulation performed show that including the ultrasonic signal's attenuation, dispersion and diffraction effects, and applying the compensation of the skull bones' influence, the ultrasound attenuation in the human intracranial media can be estimated with the deviation of ±30%, when the attenuation coefficient of the skull bones varies in the wide range of 1.0…11.0dB/(cmMHz^n). (Fig.6).

2. The realization of the simultaneous measurement of the ultrasound signal time-of-flight and attenuation in intracranial media, would allow to increase the informative capabilities of the non-invasive neurosurgical measurement systems, facilitate the non-invasive calibration procedures of such a system, and give the possibility to perform ICP or intracranial blood flow autoregulation monitoring carrying out the measurements in various parenchymal acoustic paths of the intracranial media.

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V. Petkus, A. Ragauskas, R. Jurkonis

Teorinis ultragarso slopinimo intrakranialinėje terpėje matavimo modelis

Reziumė

Straipsnyje pateiktas teorinis ultragarso slopinimo intrakranialinėje terpėje matavimų modelis, kuriame yra numatytas kaukolės kaulų ir išorinių audinių įtakos matavimų rezultatams kompensavimas. Pateikti slopinimo skaičiavimo metodai gali būti panaudoti neinvaziniuose žmogaus intrakranialinio slėgio matavimuose. Išnagrinėtos teikiamų metodų paklaidos, atsirandančios dėl ultragarso signalo slopinimo, dispersijos, difrakcijos reiškinių, pasiūlyti būdai, kaip šias paklaidas mažinti.

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