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Use of Pulse-Echo Ultrasound Imaging in Transcranial Diagnostics of Brain Structures

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Abstract—The results are presented from computer simulations of acoustic pulse propagation in heterogenous media mimicking the human head in two-dimensional and three-dimensional geometries. In the three-dimensional experiment, the cranial bone is presented as a liquid layer with a speed of sound corresponding to that of longitudinal waves in the bone. In the two-dimensional experiment, both longitudinal and transverse waves are considered. Based on data obtained in the numerical experiments, the possibility of obtaining ultrasound images of point scatterers by compensating for aberrations introduced by cranial bones is studied. It is shown that even a simple time delay correction along straight rays greatly improves the quality of an ultrasound image obtained through a nonuniform-thickness solid layer.

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INTRODUCTION

For many years, ultrasonic diagnostics has been the most affordable way of investigating the inside of the human body, and one of the safest for a patient. Ultrasonic (US) examinations make it possible us to detect different pathologies, tumors, and foreign bodies. An important problem in this area is developing means for the ultrasonic diagnostics of inner brain structures through intact cranial bones. The capabilities of US brain scanning are strongly limited because of strong aberrations introduced by cranial bones. Physically, this is due primarily to the high speed of sound in bones and the fact that the skull has substantial variations in thickness. As a result of refractive aberrations, an ultrasound beam structure is greatly distorted, leading to both reduced spatial resolution and lower signal levels. There is also considerable weakening of the ultrasonic signal, because of high absorption in the bone at megahertz frequencies and the great difference between the acoustic impedances of bone and soft tissues.

Ultrasonic examinations today must therefore be done only through so-called acoustic transparency windows, reducing the accuracy of examinations and leading to the emergence of dead zones inaccessible to ultrasound. The traditional US diagnostics devices used only for soft tissue visualization are thus unsuitable for such investigations. To improve the quality of US brain examinations, we must develop new techniques suitable for transcranial diagnostics, with account for the abovementioned effects.

Note that in addition to US technologies, modern medicine uses such highly informative means of brain diagnostics as magnetic resonance imaging and computed tomography. Despite their indisputable virtues, however, all of these have a number of substantial shortcomings, e.g., limited temporal resolution or even the inability to acquire a dynamic picture. There is thus a need for finer diagnostic tools. US transcranial brain examination could become one of these in the future.

NUMERICAL SIMULATION

To study the possibility of US transcranial brain examination, we performed a numerical experiment often used when problems of US wave scattering in heterogenous media are concerned. For the physical model to correspond best to reality, we must consider the problem in 3D geometry, and present the cranial bone as a solid layer of heterogenous thickness while allowing for the probability of shear wave propagation. Collectively, these factors make the problem of studying US pulse propagation and scattering in such a model fairly difficult, since high computing power is required. The problem was therefore considered in this work using two different models that include some simplifying assumptions.

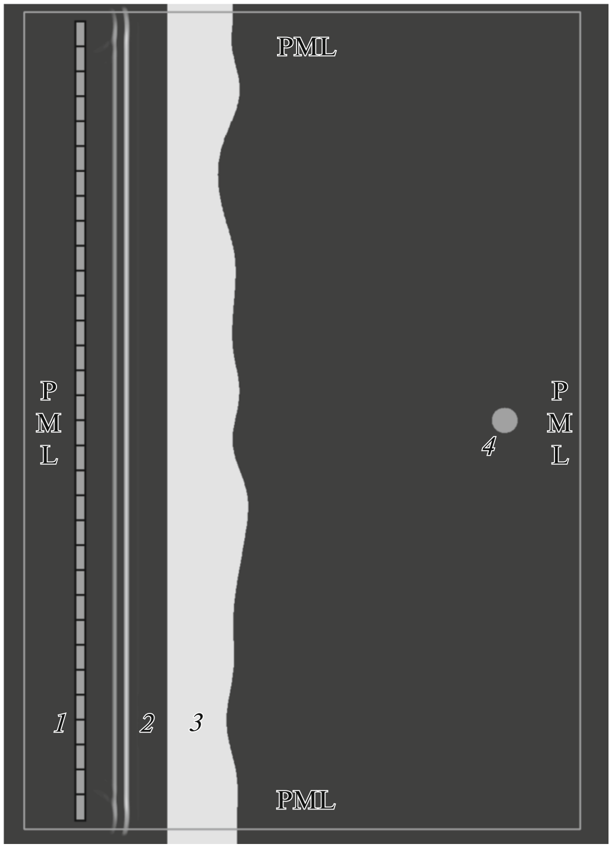

Fig. 1. Scheme of our two-dimensional numerical experiment: (*1*) multi-unit flat US sensor, (*2*) incident wave, (*3*) cranial bone, (*4*) scatterer in the form of a homogenous round object with high acoustic contrast.

TWO-DIMENSIONAL NUMERICAL EXPERIMENT

In the first case, our numerical experiment on acoustic pulse propagation from a multi-element ultrasound sensor in heterogenous media consisting of cranial bone and soft brain tissue was conducted in two-dimensional approximation. Such simplification allows us to analyze the ultrasound imaging process with allowance for the excitation of longitudinal and transverse waves in the cranial bone even on a PC.

In the simulation, the bone was considered as a nonuniform-thickness solid layer of heterogenous isotropic material, and its environment was considered to be a liquid (Fig. 1). The acoustic parameters of the bone model were chosen according their tabular values: density ρ = 1900 m s^{-1}, longitudinal wave speed c_l = 2400 m s^{-1}, shear wave speed c_t = 1500 m s^{-1} [1, 2]. The external surface of the bone (i.e., the surface directed toward the emitter) was taken to be flat, and the internal surface to contain irregularities. Mean bone thickness was l = 7.9 mm, and variations in thickness approached Δl = 3.5 mm.

The visualized scatterer was chosen in the form of heterogenous round object 3 mm in diameter, located 40 mm from the external surface of the cranial bone. The values of acoustic parameters of the scatterer in the numerical model were 1300 kg m^{-3} for density and 1950 m s^{-1} for the speed of sound. Note that the density and speed of sound of real heterogeneities in brain tissues differ by only several percent from those of water. The acoustic contrast of the scatterer in the numerical model was thus deliberately made stronger in order to increase the amplitude of the reflected signal, and to ensure its detection against the background of multiple re-reflections from the bone layer.

The incident field was presented as a plane wave. Its profile was presented as a tone burst with band center f = 1 MHz and a hypergaussian envelope. The pulse duration according to signal level e^{-1} of the maximum was 0.6 μs. This pulse corresponds to the band of frequencies used in medical ultrasound diagnostics [3]. In plane wave form, the incident field corresponds to when a multi-element US sensor produces simultaneous emission with all its elements.

The set of linear acoustics equations for a heterogenous elastic medium was solved using the finite difference method on staggered grids [4, 5] with a numerical scheme of second order precision in terms of time and spatial coordinates. To effectively remove parasitical reflections of a wave from the boundaries of the area of calculation, we used an perfectly matched layer (CFS MZT-PML) [6] whose width was set at 20 steps (1.48 mm) along its horizontal sides, and by 40 steps (2.96 mm) along its vertical sides.

As a result of numerical simulation, we acquired echo signals from the described area, registered by a virtual 128-element US sensor. The signals were used to build US images of the studied area, based on the standard Delay-and-Sum (DAS) algorithm and its modifications [7]. In the simplest variant of the algorithm, it was assumed the area of propagation was filled with water, and the time delay of the signal from the scatterer to the receiver elements was calculated along straight rays. In two other variants, the cranial bone in the studied area was allowed for. In one, the delay was calculated along straight rays; i.e., refraction was not taken into account. In the other, the ray path was calculated on the basis of Fermat's principle. This requires a large number of numerical operations, but is most effective in compensating for distortion. The two ways of calculating the delay described above assume the position, form, and thickness of the cranial bone are known. This assumption is logical, since it is sufficient to scan the human cranium and brain once (by, e.g., means of computed tomography) to know its parameters.

Let us consider US images of the studied area, acquired in three different ways. In the first image (Fig. 2a), we can see the spatial resolution of the first approach allows us to determine the heterogeneity's

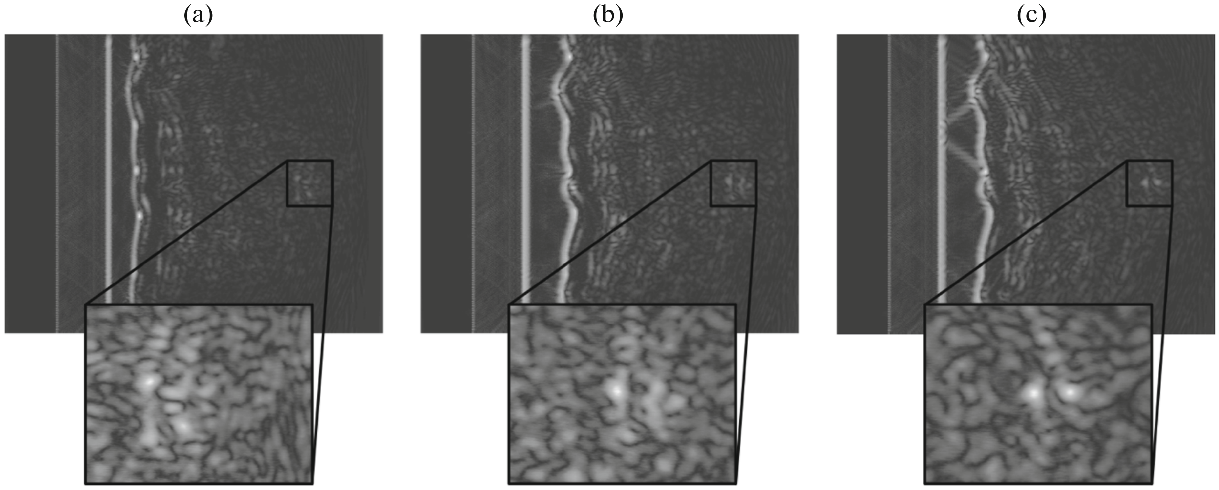

Fig. 2. Brightness pictures of the studied area, acquired (a) with no bone layer, and with a bone layer in calculating the delay along (b) direct rays, and (c) along rays traced with allowance for refraction.

location only approximately. Using the second approach enables us to improve the US image of the studied area (Fig. 2b) so we can accurately determine the location of the heterogeneity's front edge. With the third image, we can determine the locations of the front and back edges with high precision, which was not possible using the second approach. The ability to determine the exact location of the heterogeneity testifies to the increase in spatial resolution when using the above algorithms to calculate the time of US pulse wave propagation in heterogenous media of the skull-and-water type.

THREE-DIMENSIONAL NUMERICAL EXPERIMENT

The two-dimensional numerical experiment described above yielded good results on the possibility of localizing scatterers through the cranial bone in US imaging. However, a two-dimensional approximation could be too rough an approximation for the problem, since refraction effects are stronger in real three-dimensional situations. In the statement of a three-dimensional problem, the signal can be received by two-dimensional antenna arrays, offering the possibility of more effective compensation for aberration than when using one-dimensional arrays. To verify these assumptions, a three-dimensional numerical experiment was conducted (Fig. 3), but with the simplification that shear waves propagating in real bone were not considered in the model of the brainpan.

The propagation of a plane acoustic pulse with the same parameters as in the two-dimensional experiment was simulated in an area of 100 × 100 × 72 mm using the *K*-space approach to solving a system of equations for linear acoustics in a heterogenous medium [8].

The acoustic parameters of our bone model were chosen according the tabular values described in the first numerical experiment. As in the first experiment, the external bone surface was assumed to be flat, and the internal surface contained irregularities. The mean bone thickness was $l = 8$ mm, and variations in thickness $\Delta l = 4$ mm, which corresponded to the similar parameters of the two-dimensional numerical experiment.

The visualized scatterer was a homogenous sphere 3 mm in diameter, located 40 mm from the external

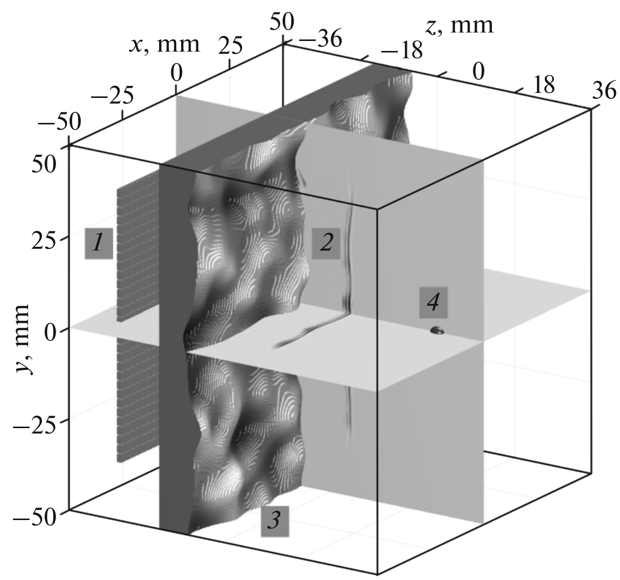

Fig. 3. Scheme of our three-dimensional numerical experiment: (*1*) multi-unit flat two-dimensional US sensor, (*2*) wave traveling through bone, (*3*) cranial bone, (*4*) scatterer in the form of a homogenous spherical object with high acoustic contrast.

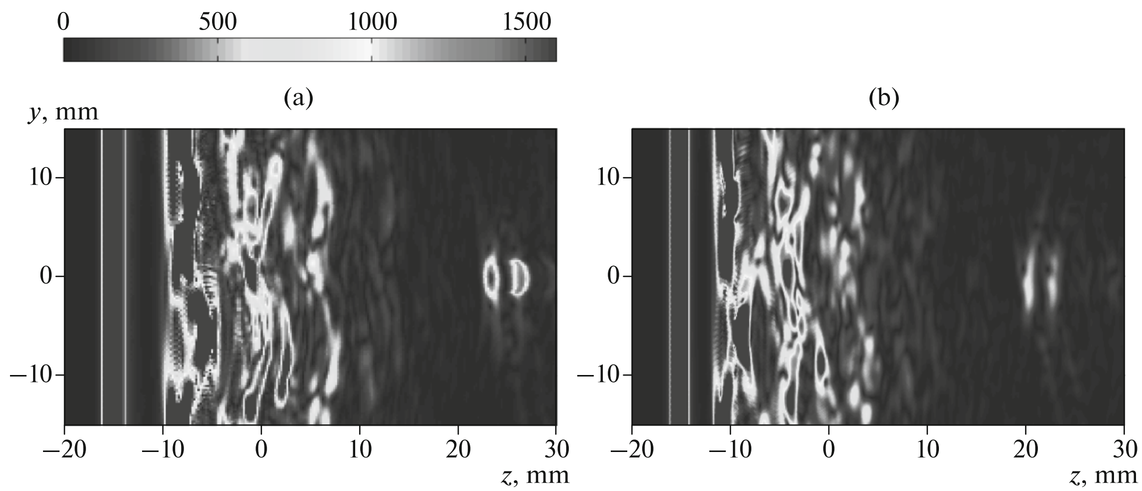

Fig. 4. Brightness picture of the studied area, acquired (a) with a bone layer in calculating the delay along direct rays and (b) with no bone layer.

surface of the cranial bone, the acoustic parameters of which were also set equal to those from the numerical experiment in the two-dimensional approximation.

Echo signals from the studied area were recorded during our simulation, using a square two-dimensional (112 × 112 elements) US array. As in the first experiment, the Delay-and-Sum (DAS) algorithm [7] was used to obtain US images, with the delay time calculated in two different ways. In the first case, it was assumed the area was homogenous and filled with water; i.e. the image was rendered without correcting for the cranial bone in the area of diagnostic pulse propagation. In the second, the cranial bone in the studied area was considered, and rectilinear pulse propagation in the media was assumed. In contrast to the two-dimensional experiment, the Delay-and-Sum algorithm was not modified to consider refraction on the bone layer's imperfections, due to the great difficulty of doing so for a three-dimensional case. To reduce the noise signal in the US image, the receiving aperture on the surface of the two-dimensional array was limited. This was done by summing the echo pulses only for the elements of the receiving antenna that fell into the base of a circular cone with radius $a = \frac{z_s - z_r}{2F_\#}$, where z_r is the coordinate of receiving antenna plane and $F_\#$ is a specific tangent of the full angle of the cone's flare, to build the image of a scatterer at the point with coordinates (x_s, y_s, z_s).

As the results, we present two US images of the studied area, obtained with (Fid. 4a) and without (Fig. 4b) allowance for the bone, and the isosurfaces of the three-dimensional image of a heterogeneity, according to the half-height of the maximum signal level for these cases (Fig. 5).

In comparing the US images of the studied areas obtained with and without correction, we can see that introducing the correction doubled the spatial resolution and the level of the signal from the heterogeneity. In examining the isosurface of the heterogeneity's three-dimensional image, we can see that the image of the heterogeneity obtained without correction has a blurred structure with several local maxima, while the image of the irregularity obtained with correction has a more regular and rounded shape. Altogether, these factors show that the proposed algorithm for considering the presence of the cranial bone in the medium makes it much easier to determine the location of the studied heterogeneity.

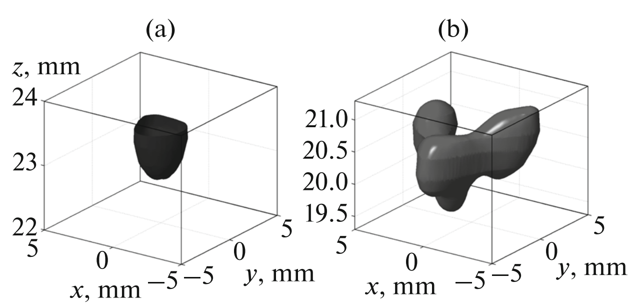

Fig. 5. Volumetric isosurfaces of a brightness picture at half the maximum level near the scatterer, acquired (a) with a bone layer in calculating the delay along direct rays and (b) with no bone layer.

CONCLUSIONS

The initial steps were taken in numerically studying the US imaging of brain structures through cranial bones. We first studied the possibility of compensating for aberrations in an image, introduced by refraction on imperfections on the internal cranial surface.

Two similar numerical experiments were performed that used the same characteristics of the studied area and diagnostic pulse, with the difference that a two-dimensional approximation was used in the first experiment, and the bone was represented by a homogenous isotropic solid layer. Three-dimensional geometry was used in the second experiment, but the cranial bone was represented by a liquid layer; i.e., the shear waves intrinsic to a solid body were not considered. In both experiments, the image of a heterogeneity with acoustic contrast was obtained, testifying to the possibility of conducting US transcranial brain examinations in the future. It was shown that the algorithm for considering the presence of the bone considerably improves the spatial resolution even if we assume rectilinear diagnostic pulse propagation.

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REFERENCES

1. White, P.J., Clement, G.T., and Hynynen, K., *Ultrasound Med. Biol.*, 2006, vol. 32, no. 7, p. 1085.
2. Fry, F.J. and Barger, J.E., *J. Acoust. Soc. Am.*, 1978, vol. 63, no. 5, p. 1576.
3. Osipov, L.V., *Ul'trazvukovye diagnosticheskie pribory. Rukovodstvo dlya pol'zovatelei* (Ultrasound Diagnostic Tools. User Manual), Moscow: Vidar, 1999.
4. Sapozhnikov, O.A., Maxwell, A.D., MacConaghy, B., and Bailey, M.R., *J. Acoust. Soc. Am.*, 2007, vol. 112, no. 2, p. 1190.
5. Vireux, J., *Geophysics*, 1986, vol. 51, p. 889.
6. Shi, R., Wang, S., and Zhao, J., *J. Geophys. Eng.*, 2012, vol. 9, no. 2, p. 218.
7. Thomenius, K.E., in *Proc. IEEE Ultrasonics Symp.*, San Antonio, TX, 1996, vol. 2, p. 1615.
8. Mast, T.D., Souriau, L.P., Liu, D.-L.D., et al., *IEEE Trans. Ultrason., Ferroelectr., Freq. Control*, 2001, vol. 48, no. 2, p. 341.

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