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TWO-DIMENSIONAL OPTOACOUSTICAL TOMOGRAPHY OF BIOLOGICAL TISSUES

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The results of theoretical investigations of process of optoacoustic (OA) pulse in flat-layered medium, its distortions during propagation in a medium with frequency-dependent attenuation and OA signal transformation in receiving/amplifying circuits are presented in this paper. Possible methods of preliminary processing of obtained oscillograms intended for compensation of distortion influence and improvement of longitudinal resolution are considered. Processed oscillograms are used then to solve problem of OA sources localization. Various methods of spatial scanning and algorithms of tomographic reconstruction are used for this purpose. The perspective of using scanning of an object by a small number of weakly directed receivers has been demonstrated.

Recently an interest to optoacoustic (OA) method increased that is caused by perspectives of its application in tomography of biological tissues [1-3]. OA tomography allows to diagnose inhomogeneities differing from surrounding tissues by optical absorption coefficient and to provide resolution about some millimeters and less in depth range of some centimeters. In contrast to non-biological objects of investigation biological tissues have several specific features related with both their optical and acoustical characteristics.

Measurements of optical properties of soft biological tissues, investigation of light propagatioon in scattering and absorbing media are of great importance for medical applications. In particular OA tomogram contrast depends on optical absorption coefficient of inhomogeneity and surrounding medium as well as on penetration depth of light inside the tissue.

It is evident that it is necessary to provide reliable receiving of OA signals on the surface of investigated object to obtain highest contrast. However form and duration of acoustical wave change under its propagation in biological tissue due to frequency-dependent ultrasound attenuation, dispersion and diffraction of acoustical wave and nonlinear effects as well [4]. Acoustical receiving system also distorts the profile of OA signal. In most experimental works ultrasonic attenuation and receiver distorsions are the main distorting factors, so let us consider them in detail.

Real inhomogeneities have irregular geometry, but for the purpose of studying distortions of OA pulse in medium and recording system we can consider one-dimensional case of flat-layered medium.

Consider flat optically transparent layer of thickness *h*, placed in the medium with optical absorption coefficient á. Let beam of heating pulsed laser radiation with uniform transversal distribution of energy density propagates normally to the layer boundaries. If laser pulse duration is lower than characteristic time of thermodiffusion and time of sound transition of light absorption length 1/ á, then straight after its termination in time moment $t = 0$ an initial distribution of pressure disturbance is formed in media [3]:

$$p(z,t=0)=\begin{cases} a\,W_0\Gamma\exp[-az]\,, & 0\le z\le L\ ;\\ 0\ , & L<z<L+h\ ;\\ a\,W_0\Gamma\exp[-a\,(z-h)]\,, & z\ge L+h\ . \end{cases} \qquad (1)$$

Here $\Gamma=\dfrac{c^2 b}{C_V}$ is thermoacoustical parameter, *c* is sound speed in the medium, *â* is thermal expansion coefficient, C_V is specific heat at constant volume, *L* is distance from plane $z = 0$ to the nearest boundary of transparent layer, W_0 is radiation energy density in laser pulse at $z = 0$.

Thus profile of initial pressure disturbance straight after heating pulse termination has a form of piecewise smooth curve exponentially falling outside transparent layer and identically equal to zero inside. In any subsequent time moment $p(z,t)$ is the superposition of two travelling waves satisfying wave equation and initial condition (1). Considering pressure disturbance to be small, wave equation for continuous medium can be linearized and its solution can be found in the form of superposition of harmonic plane waves.

For soft biological tissues the dependence of ultrasound damping on frequency is close to linear [4,5], and corresponding disperion equation can be written as: $w_{1,2} = \pm ck - im^{*}k$, where *ì** characterizes the ultrasound attenuation. In this case one can obtain the expression for pressure wave propagating in positive *z*-axis direction from flat optically transparent layer:

$$
\begin{aligned}
p(z,t) = \frac{p_0}{4} e^{-a(z-ct)} \Bigg\{ & \frac{2}{p} arctg\left(\frac{z-ct}{m^{*}t}\right) - \frac{2}{p} arctg\left(\frac{z-L-ct}{m^{*}t}\right) + \\
& + e^{ah}\left(1 + \frac{2}{p} arctg\left(\frac{z-(L+h)-ct}{m^{*}t}\right)\right)\Bigg\}
\end{aligned} \tag{2}
$$

In the case of equal acoustical impedances of adjacent layers no reflection of pressure waves occurs at their boundaries and only one wave propagating along *z* axis will reach ultrasonic probe placed in area with $z > (L + h)$.

Accomplished by authors numerical simulation of expression (2) demonstrates that as wave propagates the initially sharp edges broaden and their duration increases directly proportional to $t = \frac{m^{*}z}{c^{2}}$. At some distance the duration of initially short OA pulse becomes close to $2t$ and the magnitude of pulse begins on the decrease. Thus information about initial form and duration of OA pulse can be lost.

Usually in biotissue diagnostics it is enough to determine positions of defect boundaries. In OA tomography localization of front boundary is performed by measurement of time delay between laser pulse and rising edge of acoustical pulse on the receiver. Position of rear boundary can be determined only if resolution of registrator is high. In general spatial resolution Δr is determined by integral duration *ô** of receiver response characteristics: Δr~*cô**. Limit spatial resolution is obviously reached in a case of sharp edges of OA signal. Shift and delay of pulse edges due to its propagation in medium causes decreasing resolution and positioning precision thus decreasing quality of tomographic reconstruction.

Qualitative analysis of OA signal distortions due to frequency response of registration system can be performed studying variation of OA pulse waveform at its passage through frequency-selective filter corresponding to the characteristic passband of quasiresonant probes. Numerical simulation of such filtering shows that signal integral duration increases with distance between OA source and receiver thus decreasing system spatial resolution. Moreover the edge of recorded signal is shifted and its magnitude decreases causing lower localization precision and sensitivity of OA measurements. To improve localization precision one can apply oscillogram processing allowing partial reconstruction of initial profile of OA wave or specification of edges positions. For example, to select information about positions of OA signal edges in flat-layered medium (see Fig. 1) one can use inverse filtering procedure. Estimations demonstrate possible 2-3 times compression of diffused edge.

One-dimension images (oscillograms) of OA signals are low informative themselves for purposes of clinical diagnostics. Two-dimensional image of mutual positions of characteristic internal structures (layer boundaries, local inhomogeneities etc) gives more complete information on the morphology of investigated object. To obtain such images one must apply object scanning by means of mechanical or electronic rotation of receiver or by means of using multi-channel receiving schemes with further processing.

We suggested to use a ring of weakly directed receivers inside the object with possible access from all directions. The receivers are positioned around an object in immersion liquid and side illumination of an object is used (see Fig. 2). In this case back-projection i.e. direct algebraic algorithm similar to those useful in radar for spatial localization of multiple targets can be used for reconstruction. Position of object edges is determined by interception of arcs with radii corresponding to the delays of OA pulses propagation in medium.

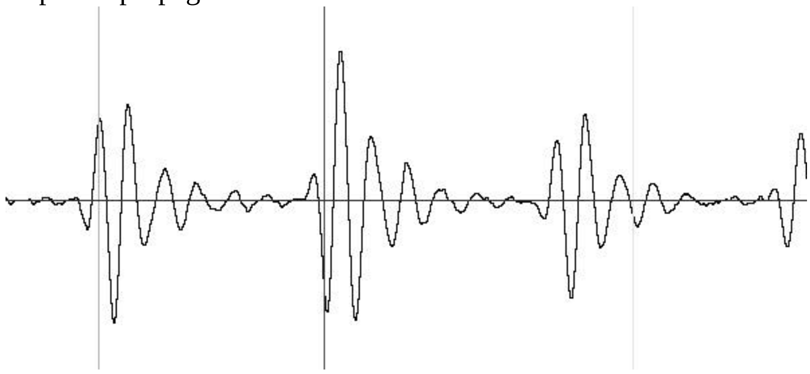

Fig. 1. The experimental oscillogram of OA signal from plane-parallel glass plate, placed in the ink-water solution.

This method gives good results for weakly heterogeneous (in acoustical parameters) medium because quality of image 'sewing' depends on accuracy of sound speed measurements. For such measurement one can use reference signals, e.g. those caused by scattered light hit on receivers and OA signals from sample boundaries. Reconstruction quality for small inhomogeneities significantly depends on number of receivers but numerical simulation demonstrates that it is enough to use sixteen receivers with 90° diagram width. Further increasing of receiver number causes complication of electronic circuits without drastic improvement of reconstruction quality.

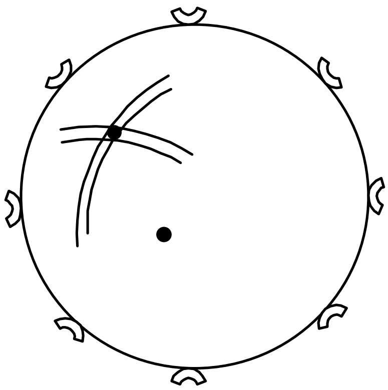

Fig. 2. Source localization by a ring of weakly directed receivers

More complicated situation occurs when access to the object is possible only in limited angle range, e.g. during OA investigation of internal organs from body surface. To solve this problem synthetic aperture system can be used allowing to obtain angular steering of acoustical beam. In this kind of system the synchronized receiving of signals by a small number N of probes localized in limited spatial area takes place followed by composing these signals with pre-calculated time delays. However such system is characterized by presence of sidelobes with level about $1/N$. For N=16 (typical value for diagnostic ultrasonic systems with electronic scanning) we obtain signal dynamic range about 24 dB. This range can be insufficient because strong signals from medium boundaries interfere localization of weak objects inside and point source image will have the form of an arc with diffuse edges. Increasing number of elements seems to be inexpedient because receiver becomes complicated and element size decreases making their matching with amplifier difficult and finally has an influence on sensitivity. Preliminary suppression of strong interfering signals before beam formation can be an outlet from this situation.

Thus application of ring system of weakly directed receivers with sufficient number of elements, using of algebraic reconstruction algorithm as well as taking account of OA signal distortion admit to get quality OA tomogramms.

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