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A PROBLEM OF A DISPLACEMENT CALCULATION OF TISSUE SURFACE IN NON-CONTACT PHOTOACOUSTIC TOMOGRAPHY**

Background. Photoacoustic tomography (PAT) is a relatively new imaging modality,which allows e.g. visualizing the vascular network in biological tissue noninvasively. This tomographic method has an advantage in comparison to pure optical/acoustical methods due to high optical contrast and low acoustic scattering in deep tissue. The common PAT methodology, based on measurements of the acoustic pressure by piezoelectric sensors placed on the tissue surface, limits its practical versatility. A novel, completely non-contact and full-field PAT system is described. In noncontact PAT the measurement of surface displacement induced by the acoustic pressure at the tissue/air border is researched.
Objective. To solve a simulation problem of the displacement calculation based on the medium pressure, which consists in deriving a formula for recalculating the pressure in the surface displacement based on the momentum conservation law, developing a simulation technique, and comparing the error of the proposed technique with the earlier used one.
Methods. Comparing the experimental data with simulated pressure data in the k-Wave toolbox. The criterion of comparison is the relative quadratic error.
Results. The simulation results of the displacement based on a new approach are more consistent with the experimental data than previous. The quadratic error numerical value of the new approach is 18% and the previous is 71%.
Conclusions. The theoretical features of the surface displacement simulation are investigated and the solution of this problem is proposed based on momentum conservation law. The implementation of the proposed methodology has a four times smaller simulation error compared to the previous technique, so it can be implemented in the non-contact PAT. The residual error can be caused by the properties of the tissue, which are not taken into account in the model, which requires further research.

Keywords: non-contact photoacoustic tomography; displacement; calculation error.

Introduction

Medical tomography has numerous methods of imaging, such as optical or electrical impedance tomography, ultrasound imaging, single-photon emission, positron emission and magnetic resonance [1, 2]. Each of them is developed to increase the image resolution and contrast or to reduce the costs and negative health effects of these techniques.

In the last decade, photoacoustic tomography (PAT) has attracted attention of various researchers as a diagnostic tool for imaging tissue vascular structures in depth up to few millimeters. The main advantage against other imaging modalities is simultaneously high spatial resolution, good contrast and a high penetration depth [3].

The physical basis of the PAT is the generation of acoustic waves by absorption of electromagnetic energy, called photoacoustic effect [4].

Non-ionizing nanosecond laser pulses illuminate biological tissue, wherein chromophores, e.g. Hemoglobin (blood vessels) absorb the optical energy and then convert it into heat. The following thermoelastic expansion results in transient acoustic waves, which can be measured at the tissue border.

In comparison to optical scattering, the ultrasound scattering in a scattering medium is two to three orders of magnitude weaker. Additionally, PAT can provide a better resolution than the pure optical tomography in depths greater than 1 mm. In the case of contact PAT, well known ultrasonic transducers are used to detect the acoustic waves. The obtained data is used to reconstruct the initial pressure.

Problem statement

The contact PAT method limits the application versatility. To overcome this disadvantage a new non-contact PAT method is developed. All reconstruction algorithms of the contact PAT may be implemented for the non-contact PAT. In the paper the approach of recalculation the acoustic wave pressure to the surface displacement and vice versa is developed.

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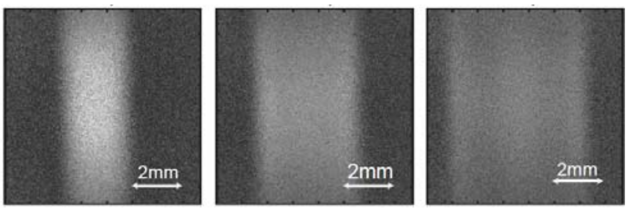
Setup for non-contact PAT

In the following, a brief introduction in a non-contact PAT setup by Medical Laser Center Lübeck GmbH and the Institute of Biomedical Optics is given, a detailed description can be found in [5].

A detection laser in conjunction with a high speed camera for scanning the surface of the object to be measured is used, without direct contact to the object. The detection laser beam is split into an object and reference beam in a Mach-Zehnder-Inferferometer. When the object beam illuminates the surface of the object, the reflected part from the surface is recombined with the reference beam. The superposition of the object and reference beam on the high speed camera allows obtaining a series of interference images. The equipment for the contact PAT and for the non-contact PAT are shown in Fig. 1, *a* and Fig. 1, *b* respectively.

The photoacoustic signals generated by the excitation laser induce a surface displacement change and a change in the pathlength of the object beam at the same time. The following interference pattern change can then be calculated back to a series of surface displacement images, Fig. 2. The axial resolution during detection is ~1 nm and the sample rate is up to 40 MHz.

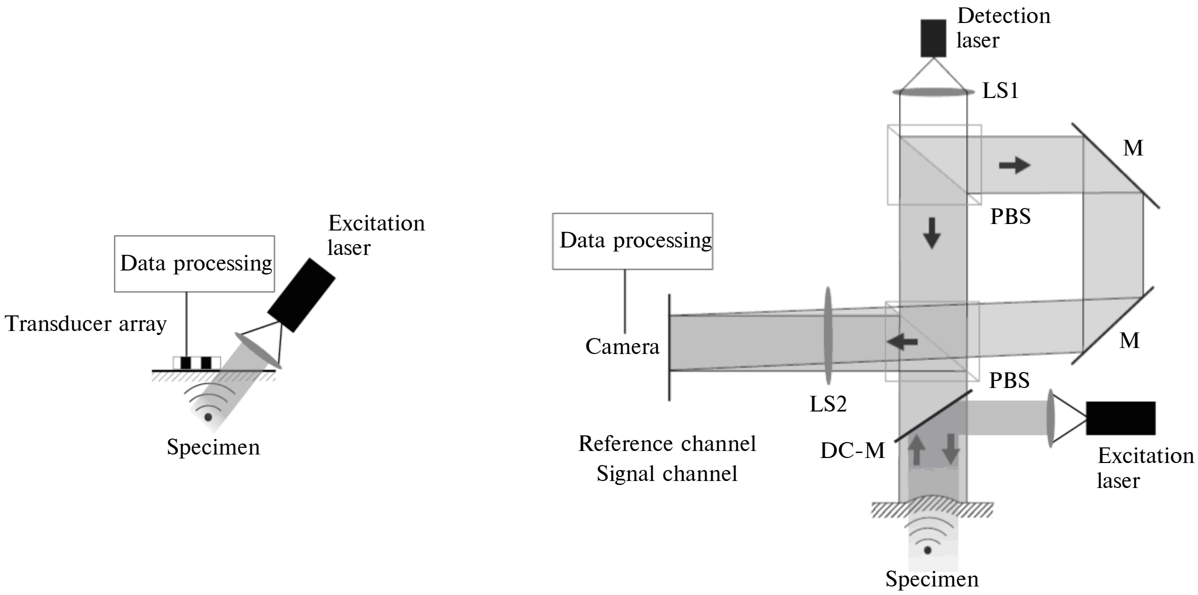
The gray value of each image pixel shows the displacement amplitude relatively to the initial state and a movement on the specimen surface to the camera appears white (Δ 1 nm ~ Δ 1 gray value). Thus, obtained images may be used for subsequent phantom reconstruction.



a *b* *c*

Fig. 2. Selected images of the surface deformation of a rod absorber at different time steps after excitation: *a* — 1.9 μs; *b* — 2.5 μs; *c* — 3.1 μs

According to the same physical fundamentals, all algorithms of the contact PAT may be implemented for the non-contact PAT [7]. It is only necessary to develop the approach of recalculation the acoustic wave pressure to the surface displacement and vice versa. The problem of calculating the displacement ξ based on the acoustic wave derivation of pressure *p* [8]. The practical implementation of this approach is influenced by the distortions in the initial pressure reconstruction. In this paper a new approach of displacement determination is proposed and compared to its error with the previous one.



a *b*

Fig. 1. Acoustic pressure detection in the contact PAT (*a*) and surface displacement detection in the non-contact PAT; the specimen is illuminated by a collimated beam of the detection laser. The backscattered light is imaged onto a camera where it interferes with reference light (LS1, lens system for beam focusing; LS2, lens system for phase extraction; M, plane mirror; PBS, polarised beam splitter; DC-M: dichroic mirror) (*b*) taken and adapted from [9]

Mathematical relations between the acoustic wave pressure and surface displacement

The previous approach of the displacement ξ calculation on basis of the acoustic wave pressure p is described by the formula:

$$\xi(t) = \frac{2}{Z}\int_0^t p(t^*)dt^*, \quad (1)$$

where Z is the acoustic impedance.

According to formula (1) the displacement ξ is defined by the pressure p on the surface, whereby neighboring points don't have any impact, which is a very rough consideration. Therefore calculations of the displacement by the formula (1) may have a methodic error that is shown in [5].

A more general approach for defining the displacement ξ is based on the momentum conservation law, which is given by [8, 9]:

$$\rho_0 \frac{\partial^2 \xi}{\partial t^2} + \nabla p = 0, \quad (2)$$

where ρ_0 is the density of the tissue, the operator Δ is a gradient operator, $\nabla = e_x \frac{\partial}{\partial x} + e_y \frac{\partial}{\partial y} + e_z \frac{\partial}{\partial z}$, where e_x, e_y, e_z are the Cartesian basis vectors.

Comparing formula (2) with formula (1), the gradient operator shows a relation between the displacement ξ and pressure for all neighboring points of the tissue. In our case, only the z component of the displacement ξ_z is known. Thus, we can rewrite formula (2) to the z direction:

$$\rho_0 \frac{\partial^2 \xi_z}{\partial t^2} + \frac{\partial p}{\partial z} = 0. \quad (3)$$

Let us express the displacement ξ in formula (3) as:

$$\xi_z(t) = \frac{1}{\rho_0}\int_0^t dt^* \int_0^{t^*} \frac{\partial p(t^{**})}{\partial z} dt^{**}. \quad (4)$$

Formulas (1) and (4) are basic for subsequent comparing the simulations by different approaches with the experimental data. In the following, the relative quadratic error δ as a criteria for comparing is proposed:

$$\delta = \sqrt{\frac{\int_{y_{\min}}^{y_{\max}}\int_{x_{\min}}^{x_{\max}}\int_{t_{\min}}^{t_{\max}} (Y_0(x,y,t) - Y(x,y,t))^2 dtdxdy}{\int_{y_{\min}}^{y_{\max}}\int_{x_{\min}}^{x_{\max}}\int_{t_{\min}}^{t_{\max}} Y_0(x,y,t)^2 dtdxdy}} 100\ \%, \quad (5)$$

where Y_0 is a reference signal and Y is the comparison signal.

To perform the comparison following procedures are necessary:

– processing of the experimental data;

– determine the initial pressure distribution for the experimental data;

– calculating the acoustic wave pressure distribution caused by the initial pressure;

– determine the surface displacement with the formulas (1) and (4) on basis of the acoustic wave pressure distribution;

– estimate the error of the surface displacement on basis formulas (1) and (4) with experimental data.

Let us consider these steps in details.

Experimental data processing

The experimental data of the surface displacement is shown as a selected image series from a full record with 200 images in Fig. 2. The surface displacement defines the gray value of the pixel. The images contain noise defined by the camera sensor and speckles, which are an intensity pattern error on the camera chip produced by coherent illumination of optical rough surface by the detection laser, see Fig. 1, *b*.

The speckle noise may be detected as incorrect data. For this purpose mean value M and standard deviation σ are calculated. The incorrect data of the displacement $\xi_z(x,y,t)$ fit to one of the following inequalities:

$$\xi_z(x,y,t) > M + 3\sigma,$$

or

$$\xi_z(x,y,t) < M - 3\sigma.$$

After speckles eliminating was performed, an averaging in time and space was applied to the data. A 3D model of the surface displacement after data processing is shown in Fig. 3. The 3D model is smooth enough and can be used for comparing the simulated data on the basis of formulas (1) and (4).

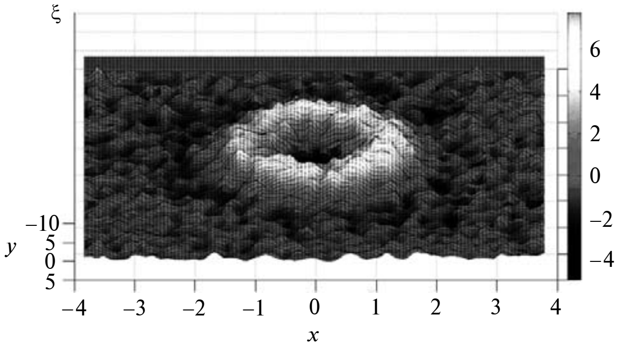


Fig. 3. The experimental displacement data after processing

The simulation procedure

In the experiment a black silicone sphere as a light absorber was used. The diameter d is 1 mm and the sphere was placed in transparent silicone (ELASTOSIL RT 604 A/B), shown schematically in Fig. 4. We consider that the medium is ideal without any absorption, except in the black sphere. We assume that the electric and magnetic permittivities of the black and transparent silicones are similar.

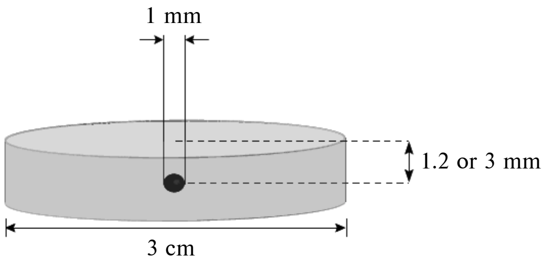


Fig. 4. Colored silicone sphere absorber in the silicone matrix

Thus the laser beam does not change the direction of the propagation. The absorption coefficient of the phantom μa is 29 cm^{-1}. It is assumed, that the pressure difference between the nearest and farthest points of the light absorber is neglectable and we can suppose that the initial pressure distribution is constant.

The initial data is sufficient for the simulation of the acoustic wave propagation. Thus, we can use already developed software, for instance the k-Wave toolbox [11] and implement the obtained data for the displacement calculation. The surface displacement based on the experiment and the formulas (1), (4) for the different time is shown in Fig. 5.

According to the simulation results, the displacement based on formula (4) corresponds to the experimental data. Also, it proves the numerical value of the error calculated by the formula (5). The error of the formula (1) is 71 % and for the formula (4) 18 %. Thus, we can conclude that formula (4) is more effective for calculating the displacement on basis of the acoustic wave pressure. The residual error may be caused by individual tissue properties. Thus, this problem requires subsequent development.

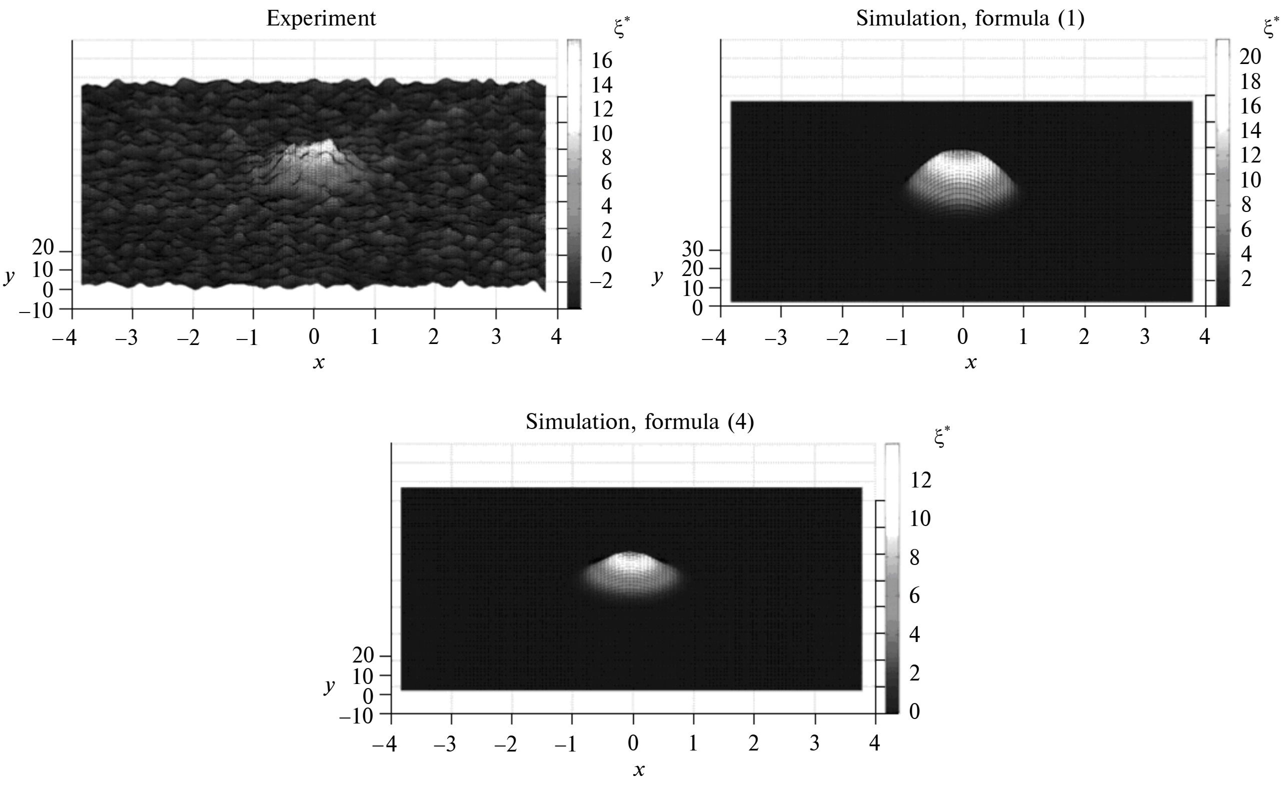
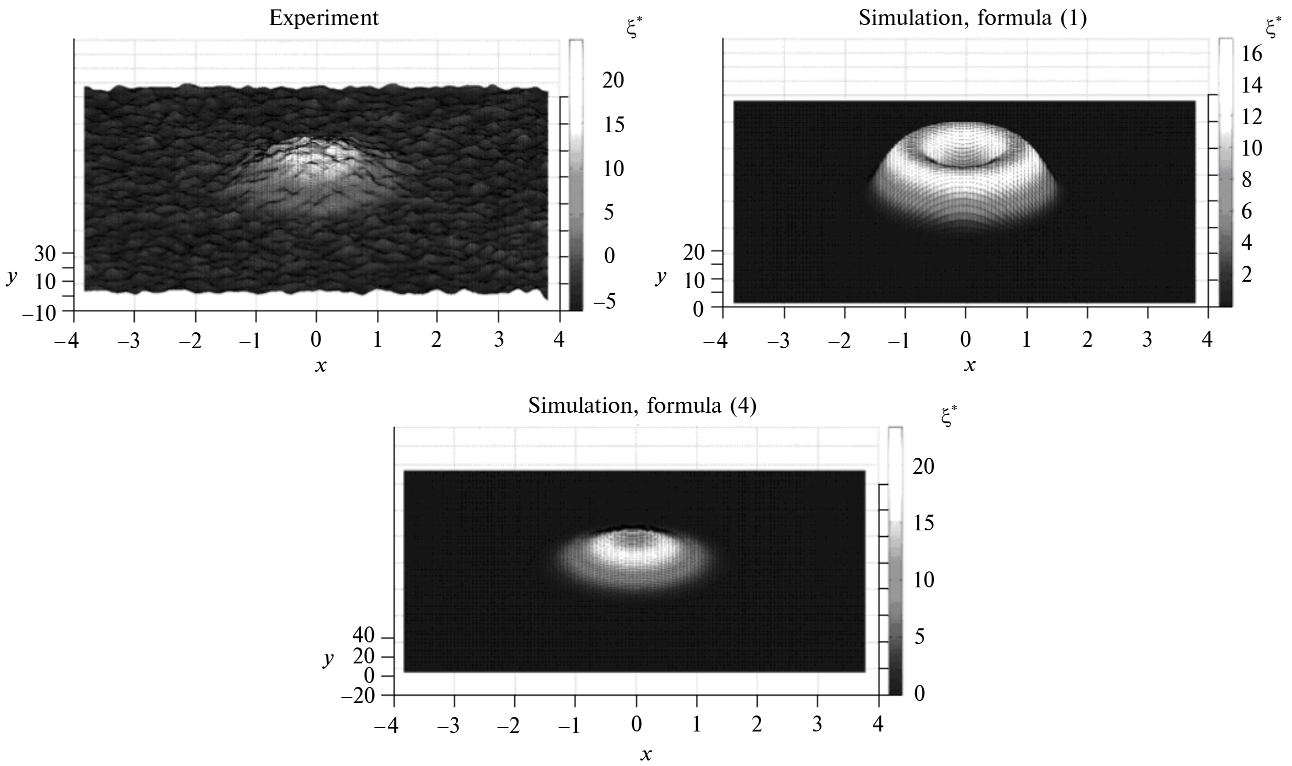
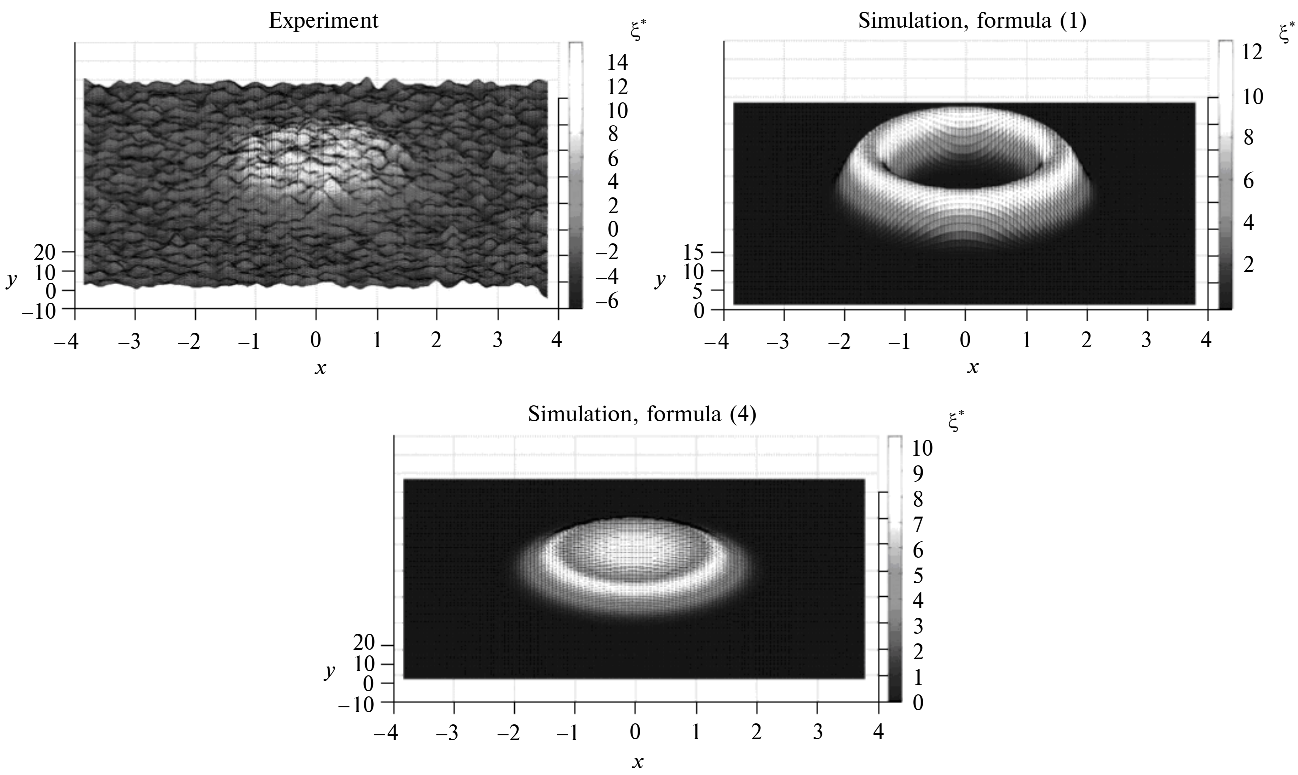


Fig. 5 (figure continued on next page). The surface displacement based on the experiment and the formulas (1), (4): a – the surface displacement at $t = 1$ μs; b – the surface displacement at $t = 1.5$ μs; c – the surface displacement at $t = 2$ μs



b



c

Fig. 5 (continuation). The surface displacement based on the experiment and the formulas (1), (4): *a* – the surface displacement at $t = 1$ μs; *b* – the surface displacement at $t = 1.5$ μs; *c* – the surface displacement at $t = 2$ μs

Conclusions

In the paper, physical fundamentals of the photoacoustic tomography are given. Differences between contact and non-contact photoacoustic tomography are identified. The problem of a surface displaycement calculation based on the acoustic pressure wave is introduced. The features of the displacement calculation on basis of a known approach are analyzed. A new approach based on the momentum conservation law is proposed and compared with the the existing. The effectiveness of the new approach is proved.

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ЗАДАЧА РОЗРАХУНКУ ВІДХИЛЕННЯ ПОВЕРХНІ ТКАНИНИ ДЛЯ БЕЗКОНТАКТНОЇ ФОТОАКУСТИЧНОЇ ТОМОГРАФІЇ

Проблематика. Фотоакустична томографія (ФАТ) є відносно новим методом діагностики, який дає змогу отримати зображення мережі судин біологічної тканини неінвазивно. Цей томографічний метод має перевагу над іншими суто оптичними/акустичними методами завдяки великому оптичному контрасту і низьким втратам енергії в тканинах. Загальновідома методика ФАТ, що базується на вимірюваннях акустичного тиску п’єзоелектричними датчиками, розміщеними на поверхні тканини, має обмежене практичне використання. У статті описана нова, повністю безконтактна ФАТ-система з повним збудженням. Досліджено основну відмінність безконтактної ФАТ, що полягає у вимірюванні відхилень поверхні тканини, спричинених акустичним тиском на межі тканина–повітря.

Мета дослідження. Розв’язати задачу моделювання відхилення поверхні на основі тиску всередині середовища, що полягає у виведенні формули для перерахунку тиску в відхилення поверхні на основі закону збереження імпульсу, розробці методики моделювання і порівняння помилки запропонованої методики з раніше використовуваною.

Методика реалізації. Експериментальні дані порівнюються з промодельованим відхиленням поверхні в програмі k-Wave toolbox. Критерій порівняння – відносна квадратична помилка.

Результати дослідження. Промодельовані дані відхилення поверхні на основі нової методики більше відповідають експерименту порівняно з попередньою методикою. Квадратична помилка нової методики становить 18 %, попередньої – 71 %.

Висновки. У роботі досліджено теоретичні особливості моделювання відхилення поверхні та запропоновано розв’язання цієї задачі на основі закону збереження імпульсу. Реалізація запропонованої методики має в чотири рази меншу помилку моделювання відхилення порівняно з попередньою методикою, тому вона може бути реалізована у безконтактній ФАТ. Залишкова помилка може бути спричинена властивостями тканини, які не враховані в моделі, що потребує подальших досліджень.

Ключові слова: безконтактна фотоакустична томографія; відхилення; помилка обчислень.

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ЗАДАЧА РАСЧЕТА ОТКЛОНЕНИЯ ПОВЕРХНОСТИ ТКАНИ ДЛЯ БЕЗКОНТАКТНОЙ ФОТОАКУСТИЧЕСКОЙ ТОМОГРАФИИ

Проблематика. Фотоакустическая томография (ФАТ) является относительно новым методом диагностики, который позволяет получить изображение сетки сосудов биологической ткани неинвазивно. Этот томографический метод имеет преимущество над другими сугубо оптическими/акустическими методами благодаря большому оптическому контрасту и низким потерям энергии в тканях. Общеизвестная методика ФАТ, которая основывается на измерениях акустического давления пьезоэлектрическими датчиками, размещенными на поверхности ткани, имеет ограниченное практическое применение. В статье описана новая, полностью бесконтактная ФАТ-система с полным возбуждением. Исследовано основное отличие бесконтактной ФАТ, которое состоит в измерении отклонений поверхности ткани, вызванных акустическим давлением на границе ткань–воздух.

Цель исследования. Решить задачу моделирования отклонения поверхности на основе давления внутри среды, которая состоит в выводе формулы для пересчета давления в отклонение поверхности на основе закона сохранения импульса, разработке методики моделирования и сравнения ошибки предложенной методики с раннее использованной.

Методика реализации. Экспериментальные данные сравниваются с промоделированным отклонением поверхности в программе k-Wave toolbox. Критерий сравнения – относительная квадратическая ошибка.

Результаты исследования. Промоделированные данные отклонения поверхности на основе новой методики больше соответствуют эксперименту в сравнении с предыдущей методикой. Квадратическая ошибка новой методики составляет 18 %, предыдущей – 71 %.

Выводы. В работе исследованы теоретические особенности моделирования отклонения поверхности и предложено решение этой задачи на основании закона сохранения импульса. Реализация предложенной методики имеет в четыре раза меньшую ошибку моделирования отклонения в сравнении с предыдущей методикой, поэтому она может быть реализована в бесконтактной ФАТ. Остаточная ошибка может быть обусловлена свойствами ткани, которые не учтены в модели, что требует дальнейших исследований.

Ключевые слова: бесконтактная фотоакустическая томография; отклонение; ошибка вычислений.

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