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Principles and Methods of Biological Objects Internal Structures Identification in Multifrequency Electrical ImpedanceTomography Based on Natural-Model Approach

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Abstract: A complex nonlinear dependence of the electrical parameters on the injecting current frequency that differs for different tissues, leads to the need of use several mathematical models: three, five and six component. Since, it's impossible to reliably determine exact location of the tissue types boundary in the body before measurement, there is a problem of suitable mathematical models selection in the multi-frequency electrical impedance tomography. In order to overcome mentioned difficulties method of biological tissues identification based on natural-model approach, applicable for two-dimensional and three-dimensional multi-frequency electrical impedance tomography is proposed. In this case injecting current of different frequencies can be supplied either simultaneously or at different times. It is assumed decomposition mathematical model of a biological object into two levels of simulation: distribution of conductivity, dielectric constant and currents in the area occupied by the object and model of object electrical parameters depending on injecting current frequency. The proposed method allows to select the most adequate model of dependence biological tissues local electrical parameters on injecting current frequency, it would be useful for the automatic identification of tissue located in considerated area. Model selection algorithm is based on the principle of building hypotheses and selection of them as a result of subsequent measurements that most accurately predict the electrical parameters dependence on frequency. The technique involves a measurement for the pre-selected injecting current frequencies, followed by automatic detection of an additional set of frequencies and additional measures for them. Thus, the scheme of cyclic multi-frequency electrical impedance tomography is proposed. At each iteration of the loop are executed sequentially: the injection of current and potential value measurements, reconstruction of the distribution of the electrical parameters, the analysis of parameters dependence on the frequency and the selection of the most adequate model. Reconstruction of the electrical parameters distribution can be performed using known numerical techniques for solving inverse problems of identification. An analysis of the applicability of the stationary and symbolic models currents spreading for the implementation of the proposed approach shows that even in the region of stationary model applicability, it's desirable to use a symbolic model due to the fact that it allows to get information about dielectric constant distribution.

Key words: Full-scale modeling experiment, inverse problem, diagnostics, simulation, regression, forecasting

INTRODUCTION

Electrical Impedance Tomography (EIT) is one of the promising methods for diagnosing a person's functional status. Its advantages are: safety, non-invasiveness, ease of hardware implementation, the economic efficiency of use in clinical practice (Aleksanyan *et al*., 2014). EIT is a method for imaging of electrical conductivity distribution in a biological object by measuring the potential values on contact electrodes excited by injecting a high frequency alternating electric current. EIT is a multidisciplinary method and requires complex numerical simulation of the electromagnetic field (Holder, 2005). Scanning can be done either on a single fixed frequency (single frequency EIT) or several (multi-frequency EIT). In the first case, the EIT possibilities limited to obtain visual reconstructed picture of tissues electric parameters distribution. Multi-frequency EIT based on nonlinear dependence of biological tissues conductivity and permittivity on the frequency that allows to receive more information in addition to a visual picture, for example, to estimate the ratio of intracellular and extracellular fluids (Nikolaev *et al*., 2009). Multi-frequency EIT consists in current supplying at different frequencies at different moments or

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simultaneously. In the first case multi-frequency EIT is a sequence of different single frequencies. The second case may also be reduced to a set of single-frequency measurements with separation of signals from different sources (Aleksanyan *et al*., 2016). Thus, the overall scheme of operation of multifrequency EIT can be represented as follows (Fig. 1):

- A series of measurements of potentials values the at the contact electrodes for multiple injection current frequency (Aleksanyan *et al*., 2015a-c)
- Solving the inverse problem of distribution of electrical parameters for each frequency
- Analysis of the frequency dependencies of the electrical parameters at each field (Aleksanyan *et al*., 2015c)

MATERIALS AND METHODS

The technique of multi-frequency EIT based on natural-model approach: Electrical parameters of biological object σ and ε have complex nonlinear dependence on the frequency of the injection current and the nature of depending may vary for different tissues (Aleksanyan *et al*., 2014). This situation leads to simultaneous use of different models approximating electrical parameters of frequency dependencies for different sub-areas of the object. As it is impossible to determine the exact position of different tissue types boundary in the body a priori before the measurement, there is a problem in selecting appropriate mathematical models in a multifrequency EIT procedure.

To overcome the difficulties identified are encouraged to use natural-model approach suggested by Gorbatenko (2001) to monitor and control the quality of ferromagnetic materials products.

The use of natural-model approach leads to a changes in the scheme of multi-frequency EIT (Fig. 1) when unit analyzing electrical parameters frequency dependences transforms to unit choosing the most suitable mathematical model describing frequency dependencies which, if necessary, will be able to request additional measurements at frequencies of a predetermined range $[f_{min}, f_{max}]$ to refine the experimental results. Thus, linear algorithm is converted into cyclic (Fig. 1). When implementing the method, the choice of initial set of allowable frequency range of r frequencies f^k for which the measurement will be performed in a first iteration is required:

$$F = \{f^k\}, f^k \in [f_{min}, f_{max}], k = 1, ..., r$$

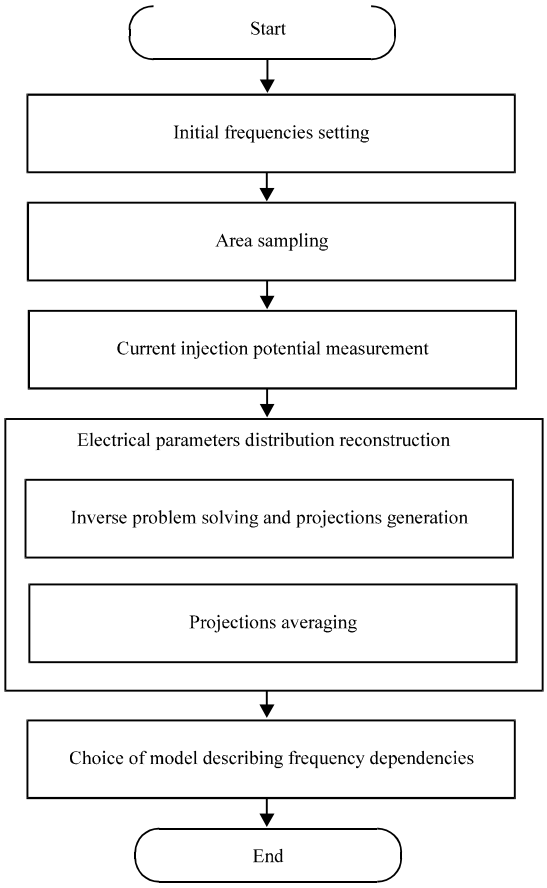

Fig. 1: The methodology of the multi-frequency EIT performing

The following describes in detail the blocks constituting the proposed technique.

Measuring system and injection strategies: Consider a on an example a measuring system multifrequency EIT of human chest. When the two-dimensional EIT performed, cross-section of the chest is considered (Fig. 2) with boundaries formed by measuring belt with fixed electrodes E1-16. Different configurations of two current electrodes are chosen to supply current and the remaining electrodes chosen to measure potentials relative to the common point.

When implementing three-dimensional EIT, volume model of chest is considered, and the electrodes are arranged in multiple zones (electrode belts). During the measurement, as in the case of two-dimensional EIT, various configurations of two current electrodes are choosed and data sets obtained for each configuration.

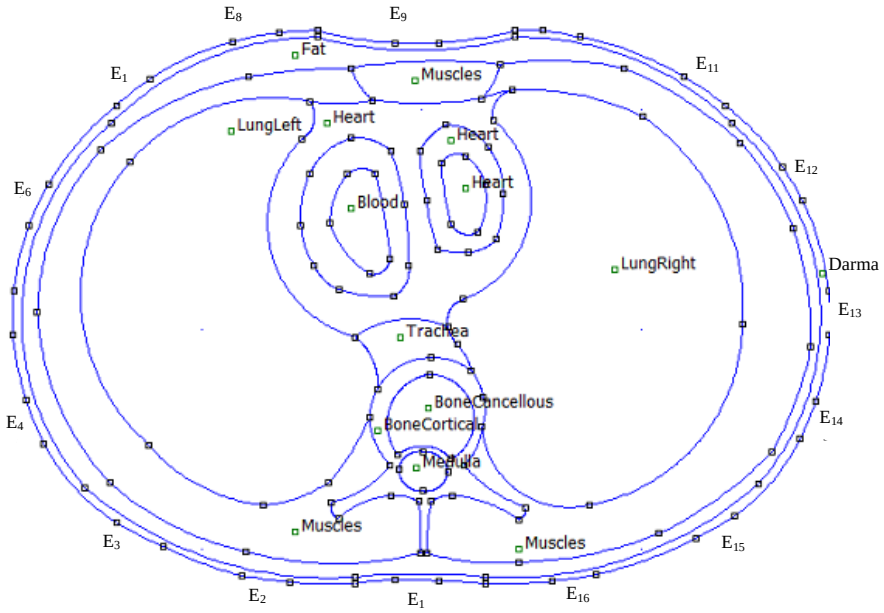

Fig. 2: Area of research when performing two-dimensional EIT (built in Femm 4.2)

We enumerate all the measuring electrodes and designate them as a plurality of electrodes $\{E_l\}$, l = 1, ..., N, where N the total number of electrodes. When the first performance of the measurement, it is advisable to perform actions for multiple originally selected frequency f^k, k = 1, ..., r. It is possible as consecutive measurements consisting in supplying current at different frequencies at different times and simultaneous when different pairs of current electrodes simultaneously connected to current sources at different frequencies. Last injection strategy reduces the time of the EIT but imposes additional requirements on the equipment. These requirements are associated with the need to divide measuring electrode signals of different frequencies.

When choosing any strategy as a result of each frequency f^k is obtained from the set of K configurations of the current electrodes with its own array of measured potential values $\Phi^{k,p} = \{\varphi_l^{k,p}\}_{l=1,..,N}$ for each configuration p = 1, ..., K.

Reconstruction of the distribution of the electrical parameters of a biological object: The major component of the method of the EIT is the solution of inverse problem of the reconstruction of the distribution of the specific electrical characteristics of the test in a volume of a biological object. Significant impact on the result of this renovation has a choice of a mathematical model of the electromagnetic field. In general, it is described by Maxwell's equations (in differential form) (Holder, 2005):

$$\nabla\times\bar{E} = -\frac{\partial\bar{B}}{\partial t},\ \nabla\times\bar{H} = -\frac{\partial\bar{D}}{\partial t}+\bar{J}_C$$
$$\nabla\cdot\bar{D} = 0,\ \nabla\cdot\bar{B} = 0,\ \bar{J}_C = \sigma\bar{E} \quad (1)$$
$$\bar{B} = \mu\bar{H},\ \bar{D} = \varepsilon\bar{E}$$

Where:

$\bar{E}$ = Electric field intensity
$\bar{B}$ = Magnetic induction
$\bar{H}$ = Magnetic field intensity
$\bar{D}$ = Electric Displacement vector
$\bar{J}_c$ = Density of the conduction currents induced by field
μ = Magnetic permeability, t-time
∇ = Vector differential operator of Hamilton

In the Cartesian coordinate system $\nabla = (\partial/\partial x, \partial/\partial y, \partial/\partial z)$ for the three-dimensional problem and $\nabla = (\partial/\partial x, \partial/\partial y)$ for two-dimensional.

The system of Eq. 1 takes into account that in the area of the object there are no free charges. Equations are given in an invariant form, valid for both two-dimensional and three-dimensional problem.

When injected harmonic current that varies sinusoidally $i(t) = i_A \cdot \sin(\omega t+\varnothing)$ where i_A amplitude of current, $\omega = 2\pi f$-angular frequency, ϕ-phase which can be a zero to a source of current as well as Maxwell's equations can be used with the measurement after the transients in symbolic form:

$$\nabla\times\dot{\bar{E}} = -j\omega\dot{\bar{B}},\ \nabla\times\dot{\bar{H}} = -j\omega\dot{\bar{D}}+\dot{\bar{J}}_C,\ \nabla\cdot\dot{\bar{D}} = 0$$
$$\nabla\cdot\dot{\bar{B}} = 0,\ \dot{\bar{J}}_C = \sigma\dot{\bar{E}},\ \dot{\bar{B}} = \mu\dot{\bar{H}},\ \dot{\bar{D}} = \varepsilon\dot{\bar{E}}$$

where the dot symbol indicates that the number is complex and contains information about the amplitude magnitude and phase shift relative to the injected current source the imaginary unit. In this case, the complex scalar electric potential can be entered:

$$\dot{\bar{E}} = -\nabla\dot{\varphi}$$

and Maxwell's equations can be reduced to a single Eq. 2:

$$\nabla\cdot\left(\dot{\gamma}\nabla\dot{\varphi}\right) = 0 \quad (2)$$

where, $\dot{\gamma} = \sigma+j\omega\varepsilon$ specific complex conductivity. Based on a mathematical model of the field Eq. 2 for each configuration injecting electrodes can be formulated inverse problem of the reconstruction of the biological object electrical characteristics for two-dimensional and three-dimensional electrical impedance tomography:

$$\left\|\dot{\varphi}-\dot{\varphi}_{\text{изм}}\right\| \to \min,\ \nabla\cdot\left(\dot{\gamma}\nabla\dot{\varphi}\right) = 0\text{-in area D} \quad (3)$$

$$\int_{E_l}\dot{\gamma}\frac{\partial\dot{\varphi}}{\partial n}d\Gamma = \dot{I}_l\text{-on electrodes } E_l,\ l=1,...,N$$
$$\frac{\partial\dot{\varphi}}{\partial n} = 0\text{-on otherwise areas of boundary}\ \Gamma/\bigcup_{l=1,...,N}E_l\ l=1,...,N$$
$$\dot{\varphi}^+ = \dot{\varphi}^-\ \frac{\partial\dot{\varphi}^+}{\partial n} = \frac{\partial\dot{\varphi}^-}{\partial n}\text{-on homogeneous are internal boundaries}$$

Where:

$\dot{\varphi}$	= Calculated potential distribution
$\dot{\varphi}_{\text{изм}} = \{\dot{\varphi}_l^{kp}\}_{l=1,..,N}$	= potential values measured on the electrodes for configuration p 1, ..., K on injection current frequency f^k, k = 1, ..., r
$\dot{I}_l$	= Current, injected through electrode
E_l, $\bar{n}$	= The normal to the boundary of the areas
Γ	= Boundary of area
D	= Where studied biological object is located
$\dot{\varphi}^-$	= Value on the boundary from the normal side
$\dot{\varphi}^+$	= value on the boundary from opposite side

Methods for solving such problems are known (Holder, 2005) and in this paper is not considered. Since, the computational domain has a complex geometry, then to solve the inverse problem it is necessary to use numerical methods requiring sampling area of the studied biological object on the elements $\{D_d\}$, d = 1, ..., m in each of which electrical parameters σ and ε are considered to be constant, independent of the spatial coordinates. Thus, in step the conductivity distribution reconstruction of the permittivity and are solved n.m inverse problems, resulting in a set of values of the conductivities $\sigma_d^{k,p}$ and dielectric constants $\varepsilon_d^{k,p}$ for each element D_d, d = 1, ..., m, current configuration p = 1, ..., K of current electrodes and frequency f^k, k = 1, ..., r.

Then, for each element D_d at each frequency f^k values of electrical characteristics are averaged on configurations:

$$\sigma_d^k = \frac{1}{K}\sum_{p=1}^{K}\sigma_d^{k,p},\ \ \varepsilon_d^k = \frac{1}{K}\sum_{p=1}^{K}\varepsilon_d^{k,p}$$

Thus, for each element ?? of the sample sets of electrical parameters will be obtained for the used frequencies $\{\sigma_d^k\}$, $\{\varepsilon_d^k\}$, k = 1, ..., r , d = 1, ..., m which can be used for the construction of the visual pattern and to determine the nature of the frequency dependencies.

Frequency model of biological tissue and the analysis of the frequency dependences of electrical parameters: The dependence of the parameters σ and ε on the frequency of injection current is caused by tissue consisting of cells complex heterogeneous structure. Frequency dependence of σ and ε conveniently described by the equivalent electrical equivalent circuits.

The most common model of a living cell as a part of the electrical circuit of the three elements (Nikolaev *et al.*, 2009): the internal cell resistance R_i, resistance of intercellular fluid R_e and membrane capacitance C_m (Fig. 3a). Here, the C_m is meant capacity of the system of two membranes through which current flows in series, i.e., $C_m = C_{member}/2$ where C_{member}-cell membrane characteristics. For such circuit impedance can be determined (Nikolaev *et al.*, 2009):

$$Z = R_{EQ}+jX_{CEQ}$$

Where:

$$R_{EQ} = \frac{R_e+\omega^2C_m^2\cdot R_e\cdot R_i\cdot\left(R_e+R_i\right)}{1+\omega^2C_m^2\cdot\left(R_e+R_i\right)^2}$$

$$X_{CEQ} = \frac{\omega\cdot C_m^2\cdot R_e^2}{1+\omega^2C_m^2\cdot\left(R_e+R_i\right)^2}$$

This approximation of equivalent electrical integrated parameters of biological object is used to estimate the ratio of the volume of intracellular and extracellular fluids. Similar formulas can be obtained for

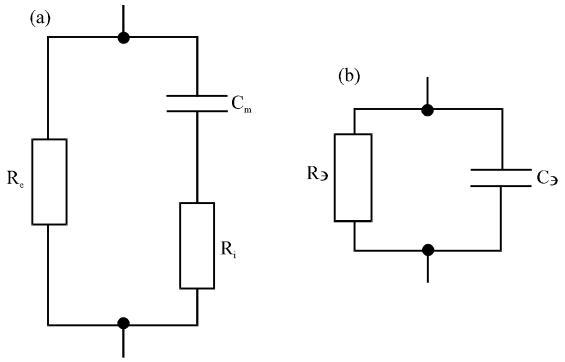

Fig. 3a, b: Equivalent circuit diagram of the living cell

local parameters- conductivity σ and permittivity ε. Equivalent circuit corresponding to formula for the complex conductivity $\dot{\gamma}$, having real and imaginary parts, shown in Fig. 3b. Here two elements, resistance R_E and capacitance C_E, connected in parallel. They correspond to the conductivity σ and permittivity ε, provided that the above electric circuit is for unit of length along the current line, then R_E and C_E are specific parameters. In this case the dependances are valid:

$$s(\omega) = \frac{1}{R_e} + \frac{\omega^2 C_m^2 R_i}{1+\omega^2 C_m^2 R_i^2} \quad (4)$$

$$e(\omega) = \frac{C_m}{1+\omega^2 C_m^2 R_i^2} \quad (5)$$

Here, the values R_i, R_e and C_m should also, be considered as averaged specific parameters the tissue.When using the model Eq. 4 and 5 the problem of analyzing the frequency dependencies of biological tissues electrical characteristics will be reduced to approximation by solving a system of nonlinear Eq. 4 and 5 relative to the cell parameters, such as resistance to intracellular fluid R_i, extracellular R_e, the membrane capacitance C_m.

When using a piecewise-constant approximation of conductivity σ and permeability ε in the volume of the object, inverse problem solution of the electrical characteristics reconstruction based on model Eq. 2 for each element of the sampling value of the electrical conductivity σ and permittivity ε, one for each frequency are received. Thus, when describing the dependencies of characteristics of the medium by three parameters, R_i, R_e and C_m, to determine it correctly at least three equations needed, making it necessary to perform measurements on at least two frequencies. Model Eq. 4, 5 will not always adequately describe the dependence of the electrical parameters of the biological object on the frequency of

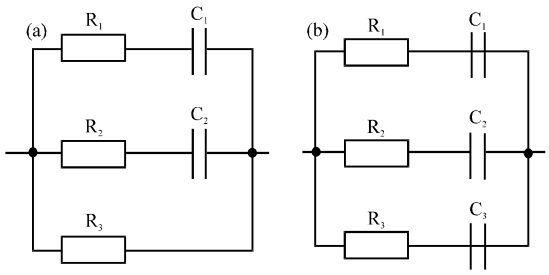

Fig. 4: a) Equivalent circuit diagram of the living cell for the five component and b) Six component models

the injected current. There are other models in the form of equivalent circuits, reflecting a heterogeneous internal structure of cells with varying degrees of accuracy (Lankin *et al.*,, 2015; Zuev *et al.*, 2012). In the present model Eq. 4 and 5, for example is not reflected resistive conductivity of cell membrane, the presence of internal membranes of animal cells having a large area is not considered. These and other features can be manifested in varying degrees for different tissues. For example, the three-component model Eq. 4 and 5 does not describes accurately the dependence of the electrical characteristics of the lungs and blood. For lung modeling is preferableto use a five-component model (Fig. 4a) and the blood-a six-component (Fig. 4b) (Holder, 2005). According to the five-component model of the electrical characteristics depending on the angular frequency approximating functions:

$$s(\omega) = \frac{\omega^2 C_1^2 R_1}{1+\omega^2 C_1^2 R_1^2} + \frac{\omega^2 C_2^2 R_2}{1+\omega^2 C_2^2 R_2^2} + \frac{1}{R_3} \quad (6)$$

$$e(\omega) = \frac{C_1}{1+\omega^2 C_1^2 R_1^2} + \frac{C_2}{1+\omega^2 C_2^2 R_2^2} \quad (7)$$

Six component model describes the frequency dependence as follows:

$$s(\omega) = \frac{\omega^2 C_1^2 R_1}{1+\omega^2 C_1^2 R_1^2} + \frac{\omega^2 C_2^2 R_2}{1+\omega^2 C_2^2 R_2^2} + \frac{\omega^2 C_3^2 R_3}{1+\omega^2 C_3^2 R_3^2} \quad (8)$$

$$e(\omega) = \frac{C_1}{1+\omega^2 C_1^2 R_1^2} + \frac{C_2}{1+\omega^2 C_2^2 R_2^2} + \frac{C_3}{1+\omega^2 C_3^2 R_3^2} \quad (9)$$

Accordingly, for the different sub-areas of the object will be the most appropriate different frequency models q = 1, ..., Q dependencies σ(ω) and ε(ω). The problem of selecting the most appropriate model depending on the electrical characteristics of biological tissue on the frequency proposed to solve during the process of the EIT (Fig. 1). Implementation of the corresponding block is offered as follows:

The approximation to each cell sample D_d d = 1, ..., m, obtained values of electric conductivity $\{\sigma_d^k\}$ and permittivity $\{\varepsilon_d^k\}$, k = 1, ..., r by all Q models. This problem consists of solving systems of 2r nonlinear Eq. 4 and 5 ternary Eq. 6 and 7 five-component and Eq. 8 and 9 for the six-component model and can be solved using the least squares method. As a result in each sample D_d will be obtained an element equivalent circuit element values with which is possible to construct sdependences $\sigma_{d,q}(\omega) = \sigma_{d,q}(2\pi f)$ and $\varepsilon_{d,q}(\omega) = \varepsilon_{d,q}(2\pi f)$ described by each model q = 1, ..., Q.

Comparison of the obtained dependences of the electrical characteristics of the injection current $\sigma_{d,q}(\omega) = \sigma_{d,q}(2\pi f)$ and $\varepsilon_{d,q}(\omega) = \varepsilon_{d,q}(2\pi f)$, q = 1, ..., Q, d = 1..m on the frequency in the frequency range $[f_{min}, f_{max}]$ to detect most of divergence intervals dependency graphs obtained using different models q = 1, ..., Q.

Selection most divergence of several frequencies $F^* = \{f^k\}$ on the obtained intervals and sending a request for additional measures for the injected current at frequencies of the set F*.

Upon receipt of additional sets of electrical conductivity $\{\sigma_d^k\}$ and $\{\varepsilon_d^k\}$ permittivity values from the inverse problem solution module, model selection is performed that most accurately predict the position of these points.

As an illustration, consider the approximation problem dependences of σ and ε on the frequency f of the injection current in the sample D_d by measurements made at two frequencies $F = \{f^1, f^2\}$. In this case as the original data the following average results of inversion problem solution will act (Eq. 3): $\{f^k, \sigma_d^k, \varepsilon_d^k\}$, k = 1, 2. Substituting them in a three-component model Eq. 4 and 5 (set q = 1 for it) we obtain a system of four nonlinear equations for the parametersR_e, R_i, C_m:

$$\frac{1}{R_e}+\frac{\left(\omega^1\right)^2 C_m^2 R_i}{1+\left(\omega^1\right)^2 C_m^2 R_i^2}=\sigma_d^1,\ \frac{C_m}{1+\left(\omega^1\right)^2 C_m^2 R_i^2}=\varepsilon_d^1$$

$$\frac{1}{R_e}+\frac{\left(\omega^2\right)^2 C_m^2 R_i}{1+\left(\omega^2\right)^2 C_m^2 R_i^2}=\sigma_d^1,\ \frac{C_m}{1+\left(\omega^2\right)^2 C_m^2 R_i^2}=\varepsilon_d^2$$

where, $\omega^k = 2\pi f^k$, k = 1, 2. Solving the system of the method of least squares we can find the values of the parameters R_e, R_i, C_m, minimizing the disparity. Substituting these values into the formula Eq. 4 and 5 will have for the sample D_d the dependences σ and ε on frequency f approximated by three-component model: $\sigma_{d,1}(\omega) = \sigma_{d,1}(2\pi f)$ and $\varepsilon_{d,1}(\omega) = \varepsilon_{d,1}(2\pi f)$.

Similarly, obtain the values of the parameters R_1, R_2, R_3, C_1, C_2 and dependences $\sigma_{d,1}(\omega) = \sigma_{d,1}(2\pi f)$ and $\varepsilon_{d,1}(\omega) = \varepsilon_{d,1}(2\pi f)$ for five-component model Eq. 6 and 7 also, parameters values: $R_1, R_2, R_3, C_1, C_2, C_3$ and $\sigma_{d,3}(\omega) = \sigma_{d,3}(2\pi f)$ dependences and $\varepsilon_{d,3}(\omega) = \varepsilon_{d,3}(2\pi f)$ for six-component model 8 and 9. Sketches of dependences are shown in Fig. 5.

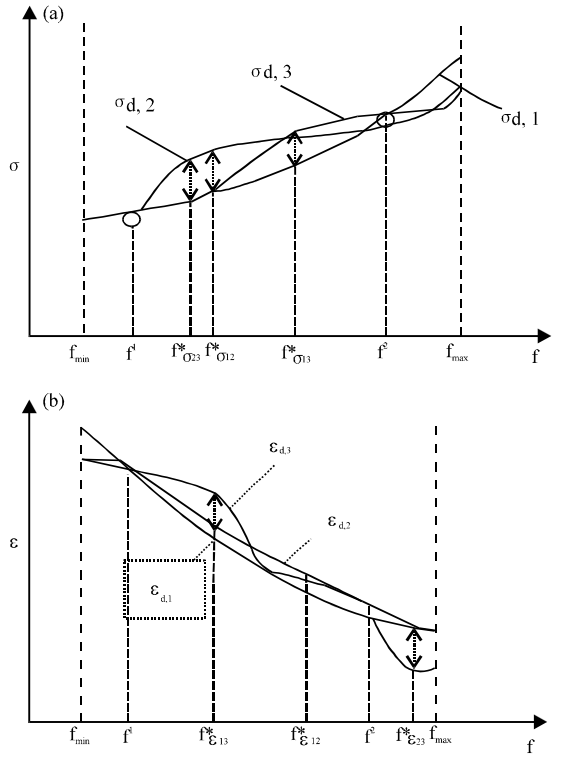

Fig. 5: a) Sketches of approximated three models the dependences of electrical characteristics σ and b) ε on frequency f in the range $[f_{min}, f_{max}]$. The dots indicate the results of inverse problem solution at frequencies f^1 and f^2.

Obtained with using of models q = 1, ..., Q dependences $\sigma_{d,q}(2\pi f)$ and $\varepsilon_{d,q}(2\pi f)$ for samples D_d, d = 1, ..., m in original data points $\{f^k, \sigma_d^k, \varepsilon_d^k\}$ will provide similar results due to a method for determining its parameters by minimizing the disparity in these points. However in the frequency range $[f_{min}, f_{max}]$ can be intervals in which there will be a significant divergence the predicted behavior of σ and ε as shown in Fig. 5. These intervals of most differences can be used to select the most appropriate model hypothesis. To do this for them additional measurements on the studied biological object and get points $\{f^k, \sigma_d^k, \varepsilon_d^k\}$, closeness to which will be model-hypothesis selection criterion that most accurately reflects the frequency dependence of the electrical characteristics.

A specific algorithm for determining the most divergence intervals is a subject of further research. It should allow to select the minimum number of additional frequencies F* = {f*}, needed for model-hypothesis selection. The difficulty here is that for the determination of these frequencies need to perform a sufficiently large number of pairwise comparisons of dependences of $\sigma_{d,q}$ ($2\pi f$) as well as $\varepsilon_{d,q}$ ($2\pi f$) for each model and for each sample and for each pair of the greatest divergence can be observed at different intervals of range $[f_{min}, f_{max}]$. For example, for sketches presented on Fig. 5, maximum divergence of dependences graphs, defined by the formulas:

$$\Delta\sigma_{d,q_1,q_2} = \max_{f\in[f_{min},\,f_{max}]}\left|\sigma_{d,q_1}(f)-\sigma_{d,q_2}(f)\right|$$

$$\Delta\varepsilon_{d,q_1,q_2} = \max_{f\in[f_{min},\,f_{max}]}\left|\varepsilon_{d,q_1}(f)-\varepsilon_{d,q_2}(f)\right|$$

will be achieved on six different $f^*_{\sigma 12}$, $f^*_{\sigma 13}$, $f^*_{\sigma 23}$, $f^*_{\varepsilon 12}$, $f^*_{\varepsilon 13}$, $f^*_{\varepsilon 23}$ for one sample D_d of m. This example illustrates the need for the introduction of integrated evaluation criteria divergence, analyzing multiple curves.

About application of a stationary model of the electromagnetic field: In the simplest approximation, the process of currents spreading in the volume of biological object is considered stationary by setting all time derivatives to zero. In this case, the system of Maxwell's equations is reduced to a single equation for the scalar electric potential φ:

$$\nabla\cdot(\sigma\nabla\dot{\varphi}) = 0 \qquad (10)$$

Such a model has the following advantages over the model (Eq. 2). The system of linear algebraic equations of inverse problem formulated on the basis (Eq. 10) has substantially twice the smaller dimension than (Eq. 3) because the latter has to deal with complex numbers. Application of the model (Eq. 10) makes weaker hardware requirements: the use of stationary model (Eq. 10) as raw data sufficient to measure the potential amplitude at the electrodes and by using the model (Eq. 2) it is necessary to know also the voltage shift relative to the injecting current, i.e., use phase-sensitive voltmeter (Brazovskii, 2015). The model (Eq. 10) has its drawbacks. Firstly, it is characterized by a lower range of applicability due to the assumption of steady-state process. Furthermore, even in the region where the model (Eq. 10) when it is selected by solving the inverse problem is only possible to obtain information on the real part of the complex conductivity, i.e., the actual picture to obtain conductivity where this information about the dielectric constant will be lost. Therefore, in proposed method is better to use a symbolic model of the field (Eq. 2) in spite of its great demands on computing resources compared with the model (Eq. 10), because of the loss of information about the dielectric constant to provide the same accuracy of the determination of the frequency patterns of parameters (Eq. 4-9), measurements at more frequencies required.

RESULTS AND DISCUSSION

The method of identification of biological tissues based on natural-model approach, applicable for multi-frequency two-dimensional and three-dimensional EIT is proposed in the article. It is assumed decomposition of mathematical model of a biological object at two levels:

- Distribution model of conduction, dielectric constant and the currents in the area occupied by the object
- Model of dependence of object electrical parameters on the injected current frequency

The analysis of the applicability of the stationary and symbolic models for the implementation of the proposed approach was performed in which found that even in the area of stationary model applicability it is desirable to use of a symbolic model.

The proposed method allows to perform measurements in the selection of the most adequate models of dependence of local electrical parameters of biological tissues on the injected current frequency, it would be useful for the automatic identification of tissue located in the studied area. Model selection algorithm is based on the principle of building hypotheses and selection of them as a result of subsequent measurements most accurately predict the dependence of the electrical parameters of frequency.

The technique involves a measurement for the pre-selected values of the injected current frequency, followed by automatic detection of an additional set of frequencies and additional measures for them. It should be noted that as the initial frequency must be selected at least three frequencies, the minimum amount required to adequately determine the parameters of six-component model. For a more accurate approximation is desirable to have a bit more of the initial frequency but an excessive

increase their lead to increased time-consuming due to the need to solve the inverse problem repeatedly and by slowing down the process of approximation of the frequency dependency of the large number of non-linear equations. The selection algorithm the values of additional frequencies in the proposed method is the subject of further studies.

CONCLUSION

Analysis of the frequency dependencies implies an approximations of electrical parameters in each sample with all models, comparison of the dependencies in the operating frequency range in order to identify the largest discrepancy intervals, requesting additional measurements for the injecting current at frequencies of these intervals. After getting additional sets of electrical parameters values, model selection is performed, that predict the position of these points most accurately. The selection algorithm of additional frequencies in the proposed method is the subject of further studies.

IMPLEMENTATIONS

Implementation of the proposed method imposes the following requirements and restrictions. Injected current should vary harmonically. Hardware with voltage amplitude measurement on the electrodes and the phase shift relative to the injected current (phase-sensitive voltmeter) capabilities needed. Potential measurements for each configuration injecting electrodes must be performed after the completion of the transitional process in the system when it goes into a well-established harmonic mode. The use of the method imposes high demands on computing power of measuring system as needed to solve a number of inverse problem solutions of identification for the reconstruction of the distribution of the electrical parameters of the tissues, the complexity of which is big. Method requires a series of measurements on a biological object which should be carried out together with the numerical processing fast enough to minimize the impact of changes in the state of the object: the breath and heartbeat when stretched in the course of research time will introduce errors in the result of the study, since, the change of characteristics of the object in time is not considered in the models. To accelerate the process of obtaining information on the potentials at the electrodes is recommended to use several power sources at different frequencies simultaneously which in turn imposes additional requirements for hardware in the form of the need to allocate signals from different sources (for example, frequency demodulation). In order to accelerate the processing of data, reducing the number of used injecting electrode configurations and use only a portion of them, instead of trying all possible is subject of further studies.

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