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Blood Glucose Level Estimation Using Interdigital Electrodes

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***Abstract*—This article explores a possibility to determine blood glucose level in a non-invasive way. Environment similar to human body was simulated in software "Comsol Multiphysics 4.3a" and most suitable topology of interdigital electrodes was chosen. These electrodes were used to carry out investigation of blood glucose levels in several test subjects. Non-invasive measurement method decreases possibility of infection and physical injuries.**

***Index Terms*—Impedance, conductivity, modelling, sensor, blood glucose level.**

I. INTRODUCTION

The incidence of diabetes is increasing worldwide every year [1]. Treating and controlling diabetes, the most important task is to determine the blood glucose level and by the need dose insulin injections.

There are several non-invasive blood glucose measurement methods. Glucometer based on radio waves transmission with consistently located transmitting and receiving antennas. The transmitting antenna sends a signal of frequency in a range from 5GHz to 12GHz while receiving antenna monitors signal attenuation, which determines the blood sugar level. Radio wave transmission requires high frequencies in order to minimize influence of the skin to the accuracy of measurement results, which is the main drawback of this sensor type [4].

Glucometer based on photoplethysmography method, in which infrared absorption measurement is used. The method is based on the fact that the blood with increased sugar level has higher absorption rate of infrared radiation than human skin. The main drawback of such measurements is that an additional sensor is required to detect heart rhythm, which determines the accuracy of the obtained results [5].

According to reference [6] the glucose level change can be detected using interdigital electrode sensor by measuring changes in conductivity and permittivity. In order to estimate applicability of such method statistical data of blood glucose level measurement using invasive and non-invasive methods should be obtained.

In order to optimize the topology of electrodes and frequency of measurement modelling was performed.

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II. MODELLING

The interdigital electrodes sensor and the surrounding environment were modelled using „COMSOL Multiphysics 4.3a" software. The most important components in the model are skin and blood. The dimensions of these components are set as follows: skin thickness – 1.5 mm, diameter of vessel – 3.2 mm [7] (Fig. 1). The modelling was performed according to algorithm presented in Fig. 2.

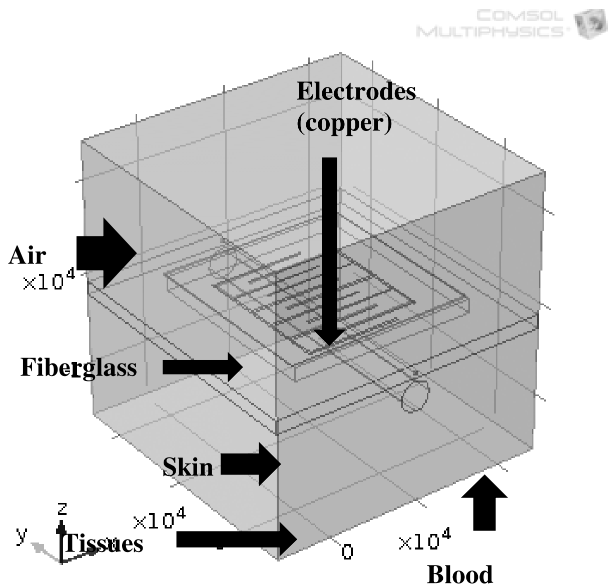

Fig. 1. The model of outer part of the hand.

After geometry selection electrical parameters of tissues, blood and skin (conductivity and permittivity) were chosen (Table I) [8], [9]. The finite element mesh composed of 321648 elements was created for further calculations.

TABLE I. ELECTRICAL PARAMETERS OF COMPONENTS USED IN MODEL.

	Permittivity (ε)	Conductivity (σ) [S/m]
Air	1	0
Copper	1	6.00E+07
Fiberglass	4.5	0.004
Skin	33 - 44	0.00002 - 0.0002
Blood	64 - 70	0.43 - 0.7
Tissues	16	0.03

The modelling was performed in frequency range from 10 kHz to 100 kHz. It was investigated how the modulus of electrical impedance varies depending on the electrical parameters of blood.

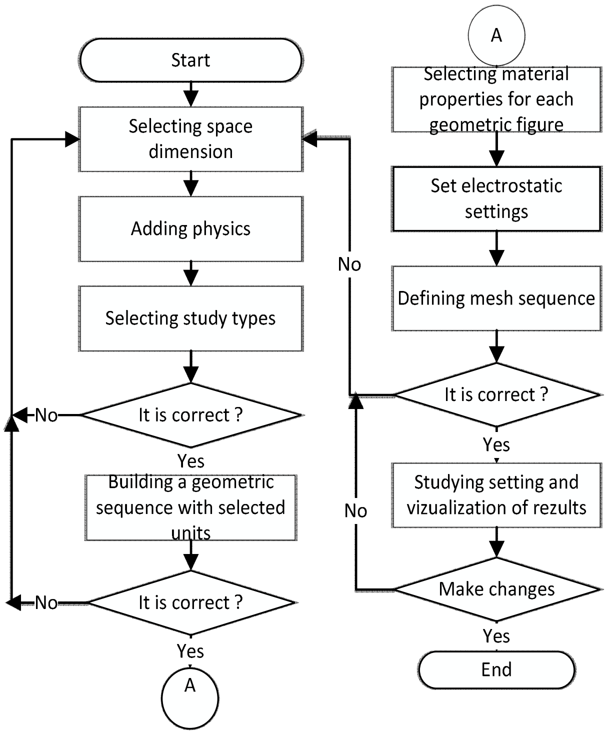

Fig. 2. The algorithm of the modelling.

Modelled electrodes were constructed so that their length does not exceed 20 mm and a width – 15 mm. Such size of electrodes allows placing them on the surface of skin avoiding air gap formation which can cause distortion of measurement results.

In the first stage of modelling, the distribution of electric field in human tissues and its dependency on frequency were calculated. As can be seen in Fig. 3. electrical field penetrates deep enough into tissues and reaches vessels. Varying electrical parameters of blood and skin resulted in highest capacitance value of 19.8 pF.

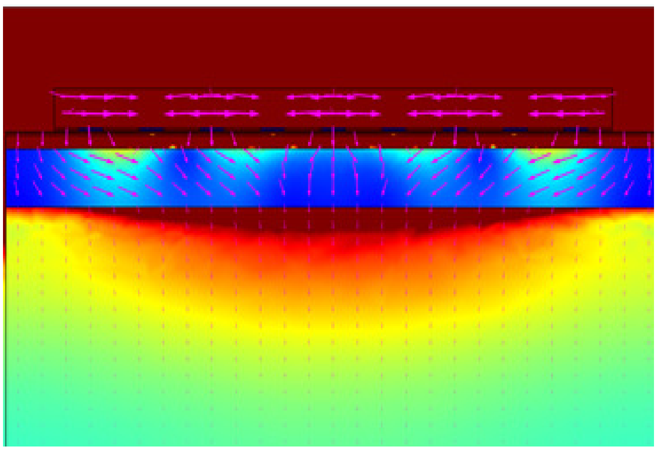

Fig. 3. Electric field distribution.

The influence of electrical parameters of blood on sensor characteristics was modelled. Conductivity of blood was incremented from 0.43 to 0.7 S/m in steps of 0.045 S/m and permittivity was set to 68.5 (Fig. 4).

For more informative representation of results coefficient K (1) was introduced:

$$K = \frac{Z_a - Zz}{Z_a}, \tag{1}$$

where Z_a – highest recorded value of capacitance or impedance modulus; Z_z – lowest recorded value of capacitance or impedance modulus.

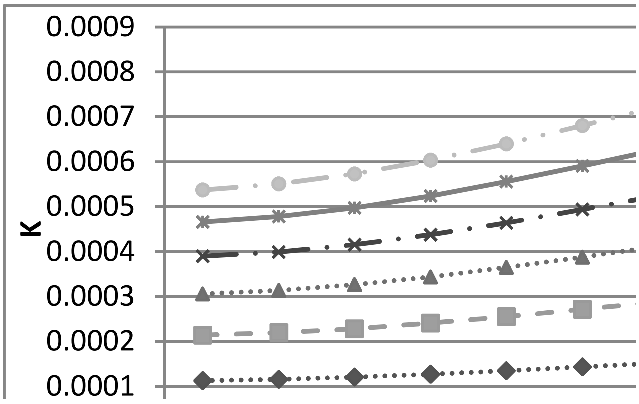

Fig. 4. Sensor impedance module dependence on conductivity of blood.

While conductivity of blood increases, coefficient K increases too in whole frequency range. Variation of sensor impedance is not very considerable, but we can assume that sensor impedance change caused by change of blood conductivity can be registered.

III. MEASUREMENTS

Using optimal parameters obtained from the model four sensors were designed and manufactured. Topology of sensors is presented in Fig. 5.

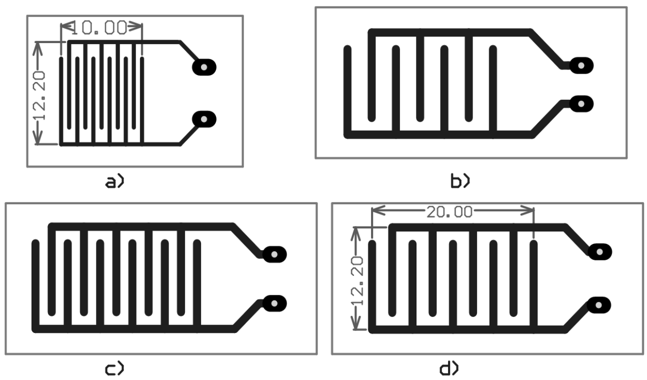

Fig. 5. Modelled and tested sensors.

Measurements have been carried out using high precision chip AD5933 from Analog Devices, which allows impedance measurement in a range from 100Ω to 10MΩ [10]. The structure of equipment used is shown in Fig. 6.

The sensor was placed on an outer side of a hand above thickest vessel named „accessory cephalic vein" (Fig. 7). This part of the hand is sufficiently flat to avoid formation of air gap which can corrupt measurement results.

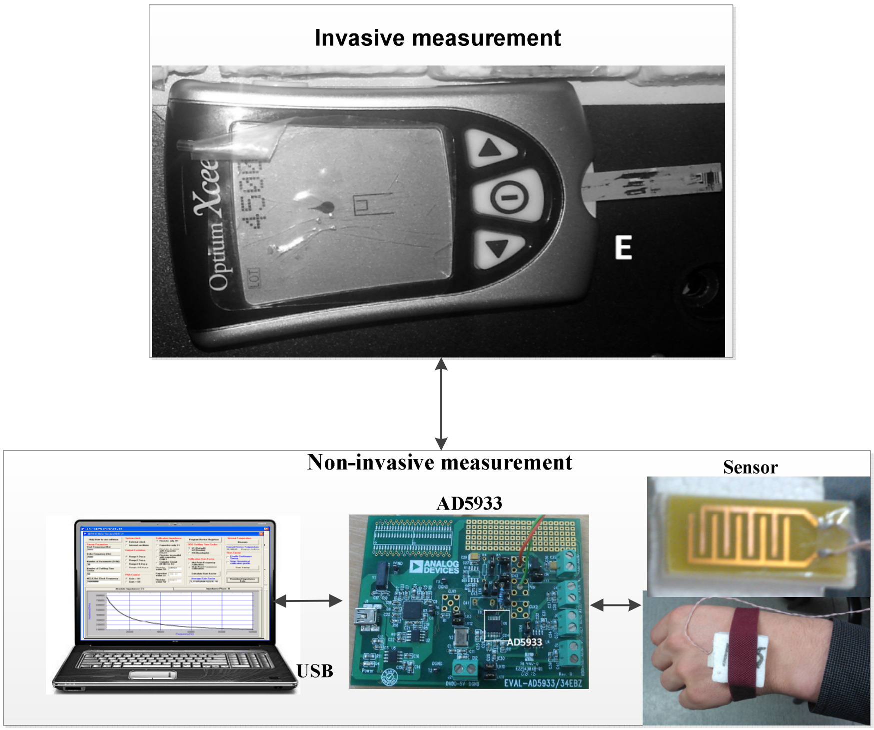

Fig. 6. Structure of measurements.

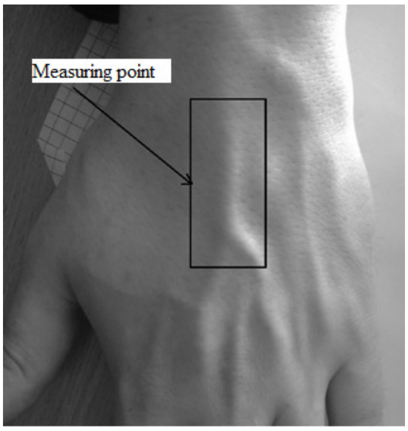

Fig. 7. The position of sensor placement.

IV. Measurement Results

The sensor capacitance and impedance dependence on blood glucose level was determined using data obtained from invasive measurements of glucose level (glucometer Abbott Diabetes Care Optium Xceed was used).

Primarily measurements of impedance at various frequencies were carried out. Investigation results of several subjects suggested possibility to determine change of blood glucose level. Impedance measurement results at different blood glucose levels are presented in Fig. 8. It can be seen that impedance module decreases when blood glucose level increases. The highest scatter of measurement results was observed at 10 kHz and 100 kHz frequencies therefore in further experiments these frequencies were no longer used.

The sensor impedance depends on frequency so it is not convenient to use this parameter for blood glucose level estimation. It was decided to calculate sensor capacitance from obtained measurement results, because the capacitance dependence on frequency is negligible in selected frequency range (Fig. 9).

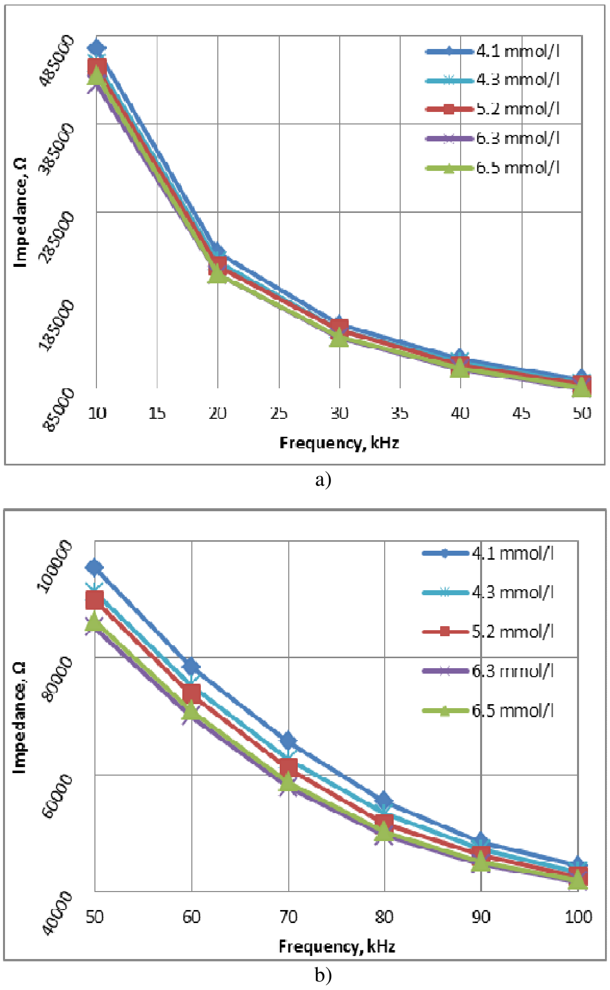

Fig. 8. Impedance dependence on frequency and blood glucose level. a) in a range from 10kHz to 50kHz b) in a range from 50kHz to 100kHz.

It can be observed from Fig. 9, that there is capacitance increase starting from 60 kHz frequency, therefore during calibration the initial sensor capacitance in the air C_0 (not placed on the patient hand) was calculated in order to eliminate parasitic capacitances. This value was subtracted from total calculated capacitance.

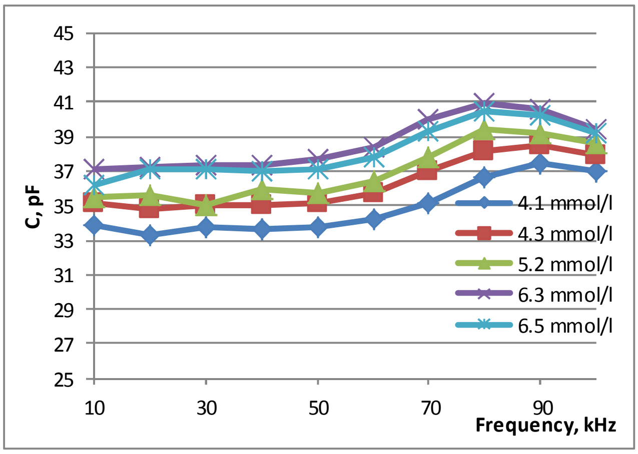

Fig. 9. Sensor capacitance dependence on frequency and blood glucose level.

In Fig. 10 sensor capacitance dependence on blood glucose level of one subject at different frequencies is presented. Scatter of measurement results at different frequencies can be seen.

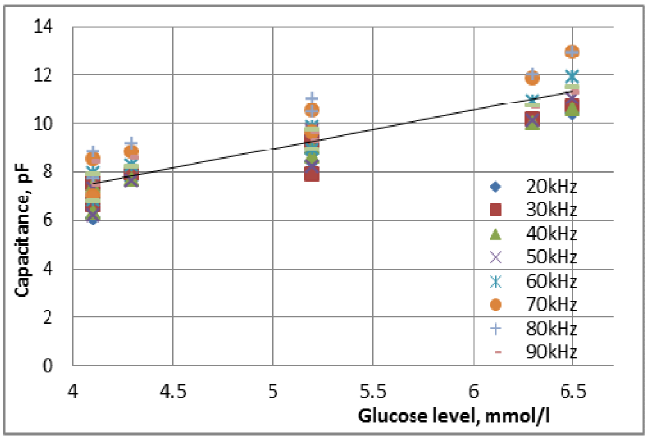

Fig. 10. The sensor capacitance dependence on blood glucose level

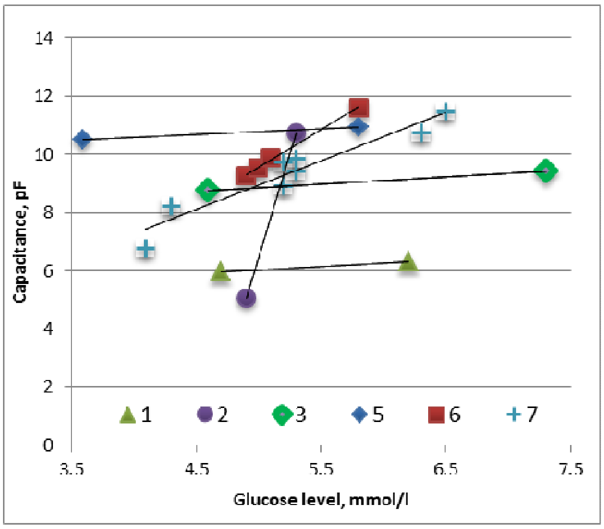

Fig. 11. Sensor capacitance dependence on blood glucose level for five subjects.

In order to reduce calibration and measurement errors, capacitance values at different frequencies were averaged. Blood glucose level increase by 2 mmol/l corresponds to capacitance increase by 3.5 pF. Dependence is linear with 0.9245 determination coefficient.

In Fig. 11 sensor capacitance change dependence on blood glucose level of six subjects are presented (actual blood glucose level was measured using invasive method). It can be observed, that for all five subjects sensor capacitance increases while glucose level increasing, but there are differences in steepness of curves. This means that for blood glucose level estimation using proposed sensor initial calibration for every individual is necessary.

V. CONCLUSIONS

Assumption that blood glucose level can be estimated using a non-invasive method was confirmed by results of the modelling. For reliable measurement the topology of electrodes was designed so that the size of electrode should not exceed 15x20 mm.

Using such method for blood glucose level estimation the continual finger pricking and potential risk of infection could be eliminated.

It was determined that impedance and capacitance of sensor depend on blood glucose level.

Experimental measurements showed that device calibration for every person increases the accuracy of blood glucose level estimation. The error of measurements does not exceed 10%.

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