Analysis of Mass loading effect on SH-SAW Biosensor
SH-SAW バイオセンサの質量解析

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1. Introduction

Recently, an innovation platform for the detection and indemnification of the infectious agents that can be used by the patients has been required. There are various important requirements for the sensor platform, such as portable, cost per test, maximum achievable sensitivity and specificity, ease of use and so on.

Many systems have been realized as easy-to-use membrane based test strips, often enclosed by a plastic cassette. However, membrane-based test strips have some disadvantages due to their visual read-out. In this study we have investigated immunoassay-based biosensor systems using shear horizontal surface acoustic wave (SH-SAW). This system is composed of SH-SAW sensor devices which can be disposable, low cost and suitable to mass-production, and an electric reader which connects to the sensor device to detect antigen-antibody reactions.

In our experimental results from a model immunoassay based upon human serum albumin (HSA) detection [1], we have obtained a limit of detection of 100pg/ml. However, a theoretical analysis of a sensitivity of SH-SAW biosensors has not been investigated.

The aim of this paper is to establish the response mechanism of the mass loading effect on SH-SAW biosensor by numerical calculation method.

2. SH-SAW biosensor

A side structure of our SH-SAW biosensor is shown in Fig.1. There is delay-line on 36-degree Y-cut Quartz substrate. The delay-line has a transmitting floating electrode unidirectional transducer (FEUDT), receiving FEUDT and a propagation area between them. The propagation area is covered with a gold film. The transmitting and receiving FEUDTs are from each end and the center-to-center distance between them is 9mm. A center frequency of about 250MHz was designed using a wavelength of 20 micron meters and a gold metallized thickness of 92nm. The FEUDTs have an aperture of 0.72mm and the number of finger pairs of 80.

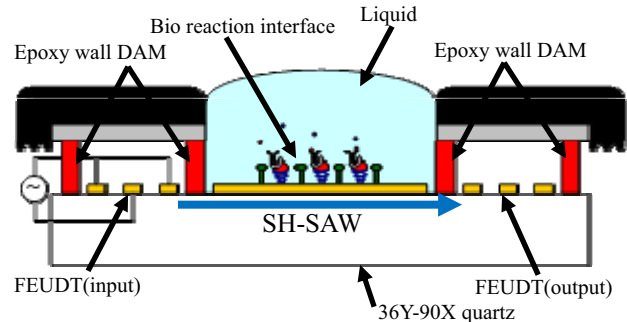
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Fig.1 Side structure of SH-SAW biosensor

The sensor has air-cavities above the FEUDTs that are composed of epoxy walls that surround the FEUDTs and glass lids. Since the FEUDTs are protected from liquids, the sensor devices can be directly dipped into a liquid and a liquid can be directly applied onto the surface of the device.

Also our SH-SAW biosensor device has a target protein recognition component which provides the specificity and selectivity of the measurement. For the detection of the specific target antigens, a modification of the sensor surface with the probe proteins that binds to the antibodies is necessary. To realize a highly defined capture protein layer, a commercialized linker layer was adopted in which self-assembled monolayer 'antibody capture' proteins were fused. The antibodies for the detection of the specific target proteins were immobilized onto this layer.

3. Numerical calculation model

Numerical calculation method was proposed by Campbell and Jones [2]. In this paper, for theoretical approach of the SH-SAW propagation characteristics, we use an improved numerical calculation method which was similar to that of Campbell and Jones but the influence of liquid viscosity was taken into account is described [3].
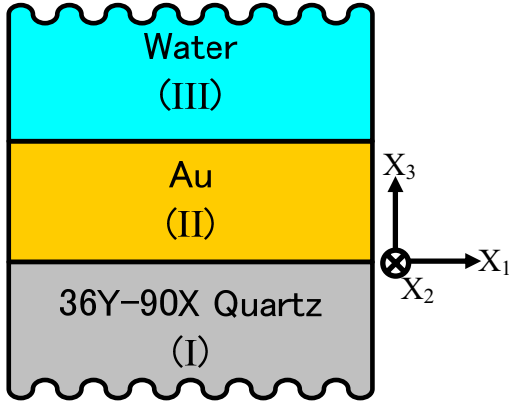
Fig.2 Calculation model

Fig.2 shows the model structure of calculation used in the present study. There are three layers: substrate (I), metal (II) and liquid (III). Substrate layer (I) is consist of 36Y-90X quartz, and metal layer (II) is consist of 92nm gold, and liquid layer (III) is consist of water. We assume only X_2 displacement vector for the SH-SAW propagating along the 36Y-90X quartz with gold electrode.

In the case of layer (I) and (II), we can use a numerical calculation method which was proposed by Campbell and Jones [2]. In the case of layer (III), we can use an improved numerical calculation method which was reported by Moriizumi, Unno and Shiokawa [3]. The boundary condition between (I) and (II), (II) and (III) are continue displacement and stress. A mass-sensitivity of the calculation model shown in Fig.2 can be obtained by changing the condition of the metal layer (II).

3. Calculation results and discussion

The calculated velocity changes of SH-SAWs for the gold film thickness changes are shown in Fig.3, as function of the normalized film thickness, h/λ.

The phase changes of our SH-SAW biosensor for the mass changes of gold are shown in Fig. 4. The phase change is shown by the following equation.

$$\Delta Ph = 360\, fl\left(\frac{1}{V_1} - \frac{1}{V_0}\right)$$

where ΔPh is the phase change, f is frequency, l is propagation length, V_0 is velocity of before loading, V_1 is velocity of after loading.

The theoretical result shows a mass-sensitivity of our SH-SAW biosensor should be 0.0006deg/pg. This value is too small to prove our HSA experimental results [1]. We obtain a phase shift of around 1 degree for a 100 pg/ml HSA

sample. Since the sample volume applied to our device is only 20 μl, the total mass of antigen in the 100 pg/ml HSA sample should be 2pg.
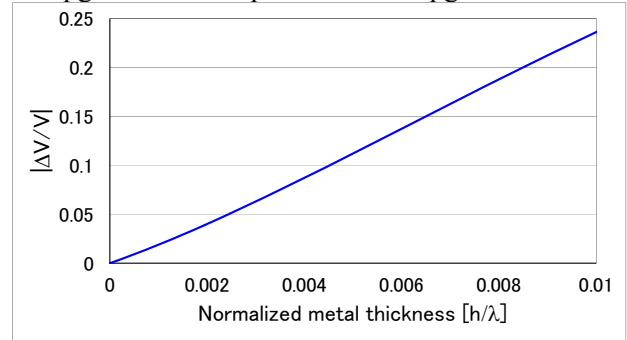

Fig. 3 Numerical calculation results of velocity change as a function of film thickness.
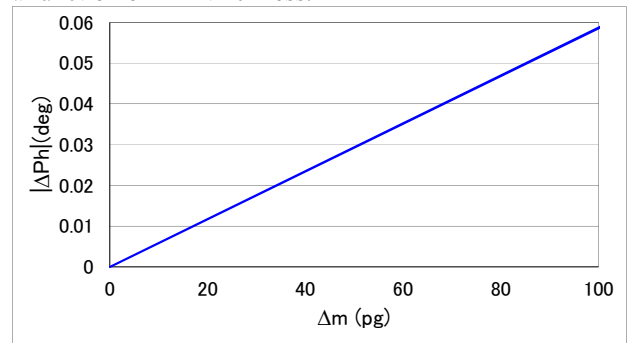

Fig.4 Numerical calculation results of phase changes as a function of mass loading.

4. Conclusions

This paper shows the theoretical results of mass loading effects on SH-SAW biosensors. In this calculation, a mass-sensitivity of 0.0006 deg/pg was obtained for our 250MHz SH-SAW biosensors on Quartz. However the value is not good enough to prove our SH-SAW biosensor sensitivity performance.

We have made it clear that a viscosity-sensitivity can be dominant rather than mass-sensitivity for our SH-SAW biosensors. In order to develop SH-SAW biosensors, theoretical analysis is very important. We are going to show theoretical analysis of viscosity-sensitivity for SH-SAW biosensors in the near future.

References
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