Scholar Publishing Group
International Journal of Multimedia Computing
https://doi.org/10.38007/IJMC.2022.030303
ISSN 2789-7168 Vol. 3, Issue 3: 33-46

Biosensors in Testing Children's Food Quality and Quality Safety

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Keywords: Grating Waveguide Mode Biosensor, Food Safety Product Detection, MEMS Micromirror

Abstract: In recent years, children's food safety has many problems, and it is particularly important to detect food quality and quality and safety in a timely and efficient manner. At present, the application of biosensors for food safety testing is becoming more and more extensive. Label-free, real-time, and highly sensitive bioassays are currently important technologies in the analysis of biology. In this paper, the optical waveguide lightmode spectroscopy (OWLS) biosensor is used to propose a biosensing detection method based on MEMS micromirror, and the concentration of glucose solution is directly detected by this method. Through the calculation and simulation, the relationship between the thickness of the waveguide film and the sensitivity is obtained. The refractive index sensitivity is proportional to the inverse ratio of the effective refractive index, and an extreme value appears during the period. The glucose solution was detected in TE mode and TM mode. The experimental results show that the solution concentration has a good linear relationship with the incident angle, and the sensitivity can reach 5 ng/m L, which is more sensitive than the traditional immunological detection method. The dynamic characteristics of the MEMS micromirrors were tested and analyzed. The method has the advantages of small volume, simple structure and no labeling, and can realize in-situ detection and avoid damage to protein activity. It is a protein-free optical detection method with great potential.

1. Introduction

From 2003 to 2016, there have been incidents of Anhui Fuyang Big Head kid, Zhengmeng infant milk powder incident, Nestl é children's milk powder iodine content exceeding the standard event, Sanlu “melamine milk” case incident, spicy strip safety issue, etc. These incidents have occurred in children's food safety, revealing that China's food safety problems are not optimistic, making children's food quality and safety once again become a hot spot in China. Table 1 shows the quality test results of children's favorite foods in the past ten years. From the test results, it can be seen that there are great quality and safety problems in various children's foods. Excessive standards such as

preservatives and sweeteners are the main causes of food failure.

Table 1. Qualification rate and non-conformity reasons for children of various varieties

Variety	Average pass rate(%)	Reason for failure
Fresh milk, yogurt	85	Preservatives, proteins
Fruit	83	Preservative
Jelly	73	Sweetener, indicator bacteria, preservative
Chocolate	87.7	Fat, microorganism
Cake, bread	85.3	Microorganisms (referring to the total number of colonies exceeding the standard, coliforms exceeding the standard, mold count indicators, bacteria-causing yellow grapes, etc.), leavening agents
Candy	89.8	Microorganism
Puffed food	81.2	Heavy metal (aluminum residue), additives (sulphur dioxide), sweetener (sweetener)
Candied food	78	Heavy metals, preservatives (sulphur dioxide), Sweetener (sweetener)
Drink	68.9	Protein, microbes, sweeteners
Biscuits	94.3	Heavy metals, preservatives (sulphur dioxide), Pathogenic bacteria, leavening agent
Spicy strip	50	Pathogenic bacteria, food additives

It can be seen from the statistics in Table 1 that food additives exceeding the standard is the number one public enemy of food safety for children in China. The seriousness of food additives is mainly reflected in preservatives, colorants and sweeteners. Sulfur dioxide, benzoic acid, sorbic acid exceeded the standard is the main cause of preservative problems. Colorants such as citric acid and carmine are severely exceeded, and sweeteners such as sodium cyclamate and sodium saccharin are also exposed to different levels. It can be seen that the two additive types of coloring agents and sweeteners are accomplices leading to food safety problems in children. There are also food additives such as carmine, lemon yellow, sunset yellow, bright blue, etc., which are banned in Europe and the United States, and are still widely used in China. In addition to the above, a large number of hormonal foods have emerged, and the use of ripeners and preservatives in fruits can be seen everywhere. These excessive chemical substances can have fatal effects on children in development, and more serious cause genetic mutation and precocious puberty and hypoplasia, which seriously endanger children's health and safety. There are also children's food problems such as excessive pesticide residues and food spoilage. This series of food safety problems are prone to children's body organ dysplasia, organ function damage, more serious cases of poisoning, suffocation, deformity, cancer and other health problems.

Therefore, it is particularly important to promptly and quickly test the quality of food and its quality and safety. There are many methods for food testing. Physical, chemical, and instrumental tests are commonly used. These methods usually require more complicated and expensive instruments and related pretreatments, making it difficult to perform on-site testing. With the development of bio-detection technology, this state has gradually changed, and bio-detection technology has been widely used in food inspection and achieved good results. For example, Li Junhua [1] used glutaraldehyde as a cross-linking agent to immobilize glucose oxidase, and ferrocene as an electron mediator to construct a novel glucose sensor with a detection limit of 0.2

mmol/L. It can effectively eliminate the effects of common interfering substances such as ascorbic acid and uric acid and successfully apply to the determination of glucose content in beverages. The biosensor can be used to determine the L(+) lactic acid content in dairy products such as yogurt, which takes only 2~3 minutes and can be used to analyze colored and turbid samples. Wang et al. [2] used nanogold as a fluorescent agent and a nucleic acid aptamer as a probe to develop a nucleic acid aptamer biosensor using modified DNA nanogold for protein detection. Mannino [3] used Eastman AQ polymer membrane to embed horseradish peroxidase, and used ferrocene as electron mediator to construct an organic phase enzyme electrode as an active center of the enzyme and an electron mediator on the surface of the electrode. This was used for rapid flow injection analysis to detect peroxide measurements in untreated edible oil samples. Sub-ramanian A [4] used a surface-plasma resonance SPR biosensor based on an alkanethiol self-assembled membrane to detect E. coli 0157:H7 quickly and specifically using a "sandwich" method. Campanella [5] studied an organic phase catalase sensor for the determination of cumene hydroperoxide and butyl ester peroxide in refined olive oil directly in n-decane and toluene.

Compared with traditional analytical methods, biosensors have the following characteristics: high sensitivity, good selectivity, fast response, simple and rapid measurement, low cost, miniaturization, portability, and on-site detection [6-8]. A biosensor is actually a device that uses biological factors or biological principles to detect or meter compounds. It is an analytical tool that uses biologically active substances (such as antibodies, purified enzymes, immune systems, tissues, organelles, or intact cells) as identification elements, coupled with appropriate physical or chemical signal transducers. The structure and working diagram of the biosensor are shown in Figure 1 [1].

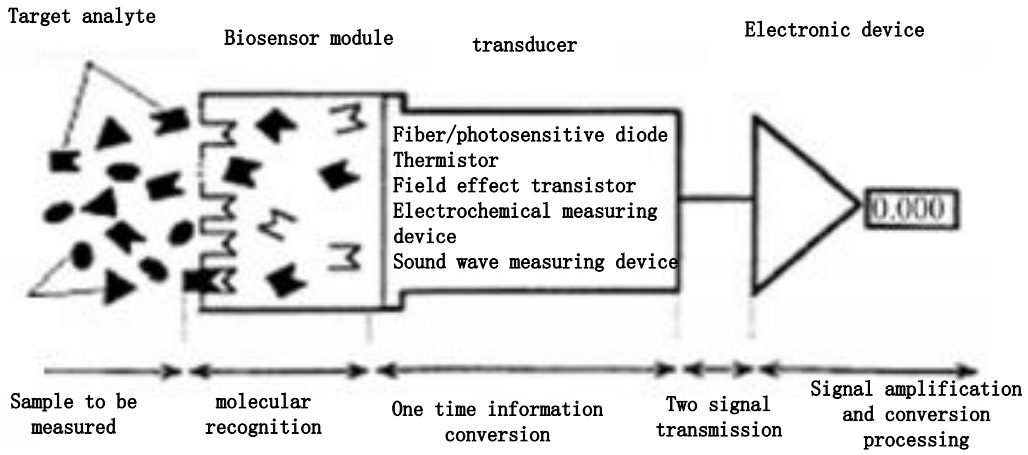

Figure 1. Schematic diagram of the composition and working principle of the biosensor

It consists of three parts: bio-sensitive component, transducer, and electronic signal processing device. The biosensing component is immobilized with a biometric component (including an enzyme, an antibody-antigen, a microorganism, a cell, an animal and plant tissue, a gene, etc.). The transducer is a signal conversion element (including a semiconductor, an optical element, a thermosensitive element, an electrochemical electrode, a piezoelectric device, etc.). The working principle of biosensors is: The substance to be tested enters the fixed biofilm sensitive layer by diffusion, and undergoes biological recognition through molecular recognition. The generated information such as light, heat, sound, etc. is converted into a quantifiable and processed electrical signal by the corresponding signal converter, and then amplified and output by the secondary

instrument, and the current value or voltage value is measured by the electrode, thereby converting the amount or concentration of the substance to be tested.

With the development of optical technology, the application of guided wave optics has been deepened, and many methods based on optical detection have been widely applied to the field of analytical biology. Such as optical waveguide lightingmode spectrum (OWLS) based optical sensor [9], SPR based biosensor [10-11], ellipsometry based sensor [12-15] and so on. The temperature-compensated long-period fiber grating sensor developed by Sirkis et al. [16] can measure the concentration of many chemical components, including alcohol, ethanol, methylcyclohexane and n-hexadecane. The long-period fiber grating sensor developed by Gu [17] et al. can measure calcium chloride and sodium chloride in aqueous solution; OWLs products produced by Microvacuum and Lasys series optical biosensors from Affinity Sensors have been commercialized. Currently, label-free, real-time, and highly sensitive bioassays are important technologies for analyzing biological fields. Therefore, this paper proposes a grating waveguide biosensor detection method based on microelectromechanical system (MEMS) micromirrors, which can adjust the incident light and use the grating coupling to excite the guided mode. The mass of the adsorbed substance on the surface of the sensor is detected by angle modulation, and the in-situ and real-time rapid detection of various components (such as glucose, protein, etc.) in a solution such as milk or beverage can be performed.

2. MEMS-based Grating Waveguide Mode Biosensor Detection Method

2.1. Biological Sensor

Due to the rapid development of biosensors, various biosensors have emerged. Biosensors can generally be classified according to the two major functional components of the biosensor (molecular recognition components and transducers) [18-19]. The classification is shown in Figure 3.

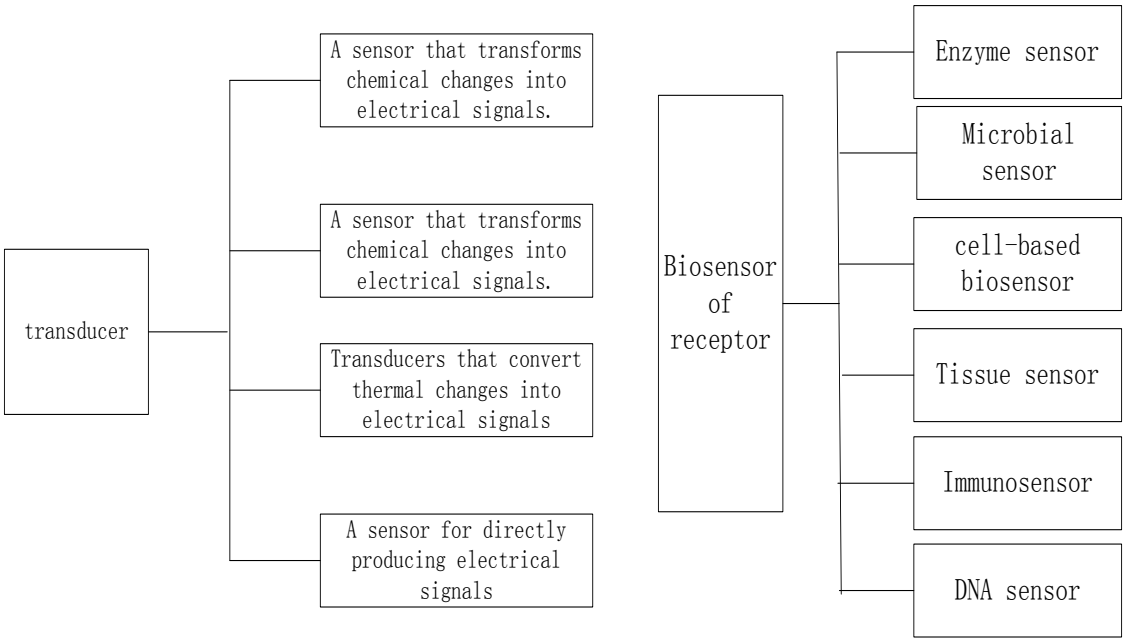

Figure 2. Classification of biosensors

1) Classified by transducer

Converting chemical changes into electrical signals

In the case of an enzyme sensor, the enzyme catalyzes the reaction of a specific substrate, thereby increasing or decreasing the amount of a specific product. An enzyme sensor is constructed by coupling a device capable of converting a change in the amount of such a substance into an electrical signal and an immobilized enzyme.

Converting thermal changes into electrical signals

The chemical reaction of the immobilized biological material with the corresponding analyte is often accompanied by a change in heat. The working principle of this type of biosensor is to convert the thermal effect of the reaction into a change in resistance by a thermistor, which is input to the recorder through a bridge with an amplifier.

Converting optical signal into electrical signal

Direct electrical signal generation

In this way, the electron transfer accompanying the enzymatic reaction, the oxidation of the microbial cells can be directly (or by the action of the electron donor) occur on the surface of the electrode, and the substrate concentration can be obtained according to the amount of current obtained.

2) Classification by biosensors of receptors

Enzyme electrode sensor

The enzyme biosensor [20] is an analytical instrument capable of quantitatively measuring a target. It uses the enzyme as a biosensing element to capture a measurable signal proportional to the concentration of the target produced by various physical and chemical signal converters to capture the reaction between the target and the sensitive element. Compared with traditional analytical methods, enzyme biosensors consist of an immobilized biosensing membrane and a transducer system that is closely integrated with it. It combines a curing enzyme with an electrochemical sensor and thus has the following unique advantages. (1) It has the advantages of both an insoluble enzyme system and a high sensitivity of an electrochemical electrode; (2) Due to the exclusive reactivity of the enzyme, it has high selectivity and can be directly measured in complex samples. Its working principle is that when the enzyme electrode diffuses into the tested solution, the substrate to be tested enters the inside of the enzyme layer and participates in the reaction, and most of the enzyme reactions produce or consume a substance that can be measured by the implantable electrode. When the reaction reaches a steady state, the concentration of the electroactive substance can be measured by a potential or current mode. At present, enzyme biosensors play a very important role in the field of biosensors.

Microbial sensor

The microbial sensor [21] consists of an immobilized microbial membrane and a transducer. It not only utilizes the function of single or multiple enzymes in a microbial cell (similar to an enzyme sensor), but also utilizes the physiological functions of the microorganism (similar to respiratory activity, metabolites, etc.). It can be divided into three categories: the activity of the microbial enzyme catalyzed reaction as the index, the microbial respiration activity as the index and the microbial metabolite as the index. This type of sensor is characterized by relatively low price, good stability, and longer response time than the enzyme sensor, but the specificity is inferior to that of the enzyme sensor.

Immunosensor

The immunoelectrode is an electrochemical biosensor that uses an immunological substance (antigen or antibody) as a sensitive element. The high specificity of the immune substance allows

the immunoelectrode to be highly specific. According to whether the marker is required in the measurement process, it can be divided into a direct immunoelectrode and an indirect immunoelectrode. Indirect immunoelectrodes require the preparation of an enzyme-labeled antibody or an enzyme-labeled antigen conjugate as a label. The indirect immune electrode uses a label to amplify the signal of the immune response to indirectly determine the antigen or antibody.

DNA sensor

The DNA sensor [22] is a single-stranded DNA molecule that has a nucleotide sequence that has been immobilized on the surface of the susceptor, and hybridizes to form a double-stranded DNA according to the principle of complementary pairing of bases, and then undergoes a hybridization process through a transducer. The changes are converted into physical quantities such as electricity, light, and sound, and the response signals or related gene information are analyzed by means of microelectronics. It then converts the hybridization process or the changes produced into electrical, optical, acoustic and other physical quantities through a transducer, and analyzes the response signal or related genetic information by means of microelectronic technology. The DNA sensor consists of a genetic recognition system and corresponding transducers. According to the different transducers, it can be divided into electrochemical type, photochemical type and piezoelectric line DNA sensor. Among them, electrochemical DNA sensors are currently the most promising DNA analysis methods. It forms double-stranded DNA (dsDNA) by specific recognition of a specific sequence of ssDNA immobilized on the surface of the electrode and complementary DNA in solution (molecular hybridization). At the same time, by detecting the change of the electrochemical response signal of the hybridization indicator capable of recognizing ssDNA and dsDNA, it is possible to achieve the purpose of detecting the presence of the gene. At the same time, when the concentration of the complementary sequence DNA changes, the response signal after the indicator is embedded also changes in response. Within a certain range, the response signal of the indicator is linearly related to the concentration of the DNA substance to be tested, thereby enabling detection of the gene content for quantitative purposes. The advantage of the DNA sensor is that it enables simpler, faster, and cheaper detection. In addition, it has higher sensitivity and specificity than traditional DNA hybridization assays.

Tissue electrode and organelle electrode sensor

The tissue electrode sensor [23] refers to an electrochemical sensor that directly uses an animal and plant tissue sheet as a sensitive element, and the principle is to utilize an enzyme in animal and plant tissues as a catalyst for the reaction, and its working principle and structure are similar to those of the enzyme electrode. The advantage of the tissue electrode sensor is that the enzyme activity and its stability are higher than that of the enzyme, the material is easy to obtain, the preparation is simple, and the service life is long. However, there are still shortcomings in terms of selectivity, sensitivity, and response time. Animal tissue electrodes mainly include renal tissue electrodes, liver tissue electrodes, intestinal tissue electrodes, and so on. Plant tissue electrodes include roots, stems, leaves, flowers, and the like of different plants. Plant tissue electrode preparation is simpler, less expensive, and easier to store than animal tissue electrodes.

The organelle electrode sensor [24] is a sensor that utilizes animal and plant cells as sensitive elements. The organelle refers to tiny "organs" surrounded by the envelope contained in the cells, such as mitochondria, microsomes, and lysosomes. Different organelles contain some unique enzymes, often multi-enzyme systems. Therefore, it can be used to determine substances that cannot be measured by a single enzyme sensor. The enzyme is in a stable state therein. Therefore, its performance is stable. However, its preparation and extraction process is more complicated.

2.2. Detection Method

2.2.1. Optical Waveguide Lightmode Spectroscopy Biological Sensor (OWLS)

This detection method uses OWLS as a sensor chip. The OWLS chip [25] is a sensor that specifically detects molecules present on the surface. It consists of a high quality glass substrate layer and a refined grating waveguide film layer located thereon. The grating is located at the interface of the waveguide, which is a sub-wavelength grating $\Lambda<\lambda$, the refractive index of the waveguide layer is nF, the refractive indices of the substrate and the cladding are nS, nC, respectively, and nF> nS, nF> nC, and the thickness of the waveguide is dF. The incident light excites the guiding mode through the grating coupling, and the guiding mode generates an evanescent wave at the interface between the slab waveguide and the adjacent medium, and penetrates into a certain depth in the adjacent medium, so the method is sensitive to the refractive index of the adjacent medium. The method is simple in structure, easy to control surface modification, and can realize label-free detection, avoiding the destruction of protein activity by labeling, and is a label-free optical method for detecting protein chips with great potential.

The principle of OWLS sensing [26] is shown in Figure 3. The analyte A to be tested in the solution flows through the surface of the ligand B immobilized on the surface of the optical biosensor to form a structure of AB combination. The theoretical basis of OWLS is the theory of planar optical waveguides, that is, some discrete optical modes that can propagate inside the optical waveguide. The characteristic parameters of these modes of light are closely related to the environment in which the waveguide is located. All of the characteristic parameters of the guided light will change as the biological environment in the optical waveguide cladding changes. Therefore, changes in the biological environment to be measured can be sensed by measuring the guided light parameters.

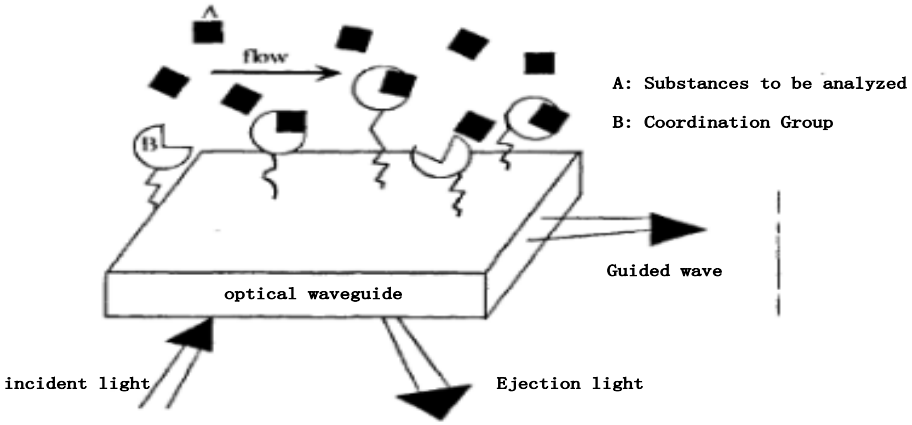

Figure 3. Schematic diagram of a typical optical biosensor

The total phase difference of the light wave going back and forth between the upper and lower interfaces in the waveguide includes the phase difference of total reflection occurring at the upper and lower interfaces and the phase difference of the optical path within the waveguide. Therefore, the change of the refractive index of the adjacent medium causes a change in the phase difference of the total reflection of the light wave at the upper interface, thereby causing a change in the total phase difference of the light wave between the upper and lower interfaces in the waveguide, thereby causing a change in the light intensity [5]. In the slab dielectric optical waveguide, the grating

dielectric waveguide propagation mode equation is

$$\phi = K_{z,F} 2d_F + \varphi_{F,S} + \varphi_{F,C} = 2\pi m \tag{1}$$

Wherein $K_{z,F}$ represents the wave number of the wave vector in the z direction perpendicular to the surface of the waveguide film in the waveguide film.

$$K_{z,F} = kn, \cos\theta = k(n_F^2 - N^2), k = 2\pi / \lambda \tag{2}$$

Where θ represents the mode angle of the light wave in the waveguide film, d_F represents the thickness of the waveguide film, $\varphi_{F,S}$ and $\varphi_{F,C}$ are the phase shifts when the upper and lower interfaces of the waveguide are totally reflected.

The sensor is used to detect the waveguide, and when the biomolecule layer formed on the surface changes, the biofilm is formed along the surface of the grating waveguide. This can be seen as a modified three-layer structure of the waveguide overlay [27], which has:

$$\phi \cong K_{z,F} 2(d_F + \Delta d_F) + \phi_{F,S} + \phi_{F,C} = 2\pi m \tag{3}$$

If the phase change is equal to zero during a total internal reflection period, that is, the mode equation of equation (3) is satisfied, the single-mode resonance propagation mode at m=0 is excited.

$$0 = \frac{2\pi}{\lambda}\sqrt{n_F^2 - N^2} \times \left\{ d_F + \frac{n_A^2 - n_C^2}{n_F^2 - n_C^2}\left[\frac{(N/n_c)^2 + (N/n_A)^2 - 1}{(N/n_c)^2 + (N/n_F)^2 - 1}\right]^p d_A \right\} - \arctan\left[(\frac{n_F}{n_C})^2 \frac{\sqrt{N^2 - n_S^2}}{\sqrt{n_F^2 - N^2}} \right] \tag{4}$$

Where n_A is the refractive index of the adsorption layer, ρ is a constant, $\rho = 0$ is the transverse electric (TE) mode, $\rho = 1$ is the transverse magnetic (TM) mode, N is the effective refractive index, including N_{TE} and N_{TM}, and they meet the respective incident angles θ_{TE} and θ_{TM}:

$$N = n_0 \sin\theta_0 + l\frac{\lambda}{\Lambda} \tag{5}$$

Where n0 is the refractive index of air, θ0 is the incident angle of incident light in air, k0= 2π/λ0, l is the order of the diffracted light, andΛis the grating period. The thickness dA of the adsorption layer and the refractive index nA of the adsorption layer can be simultaneously determined by the parameters measured in the formula (5). According to the Feijter formula, as long as the molecular film thickness and the refractive index of the molecular film layer are determined, the absolute value of the mass density of the surface adsorption can be directly calculated by the thickness and the refractive index [28].

$$M = d_A(n_A - n_C)/(d_n / d_c) \tag{6}$$

In the formula, dn/dc is the refractive index change rate. In general, the refractive index change rate of the protein is 0.12 m L/g.

When the guided mode propagates in the waveguide and penetrates into the adjacent medium, the mode in the waveguide changes as the refractive index of the adjacent medium changes, and the

incident angle condition of the excitation guided mode changes. Therefore, by continuously measuring the change in the incident angle, the adsorption quality of the biomolecule can be directly monitored on the grating in real time without any mark.

2.2.2. OWLS Performance Evaluation Indicators

The amount tested in OWLS is the effective refractive index N, and any change in the environment in the sensor will cause a change in the effective refractive index. Therefore, the sensitivity of this sensor is defined as the first derivative of the effective refractive index relative to any environmental change [29]:

$$s\{x_i\} = \partial N / \partial x_i \tag{7}$$

Where $x_i = \{n_S, n_F, d_{F,} n_{c,} n_A, d_A\}$

When the sensor is used in a single mode, the object to be tested is the refractive index $n_{c,}$ of the overlay. According to equation (7), the sensitivity of the sensor is:

$$s\{n_c\} = \partial N / \partial n_c \tag{8}$$

When the sensor is used in the surface mode, there are two sensitivities in the evaluation index. One is the sensitivity to the rate of change of the refractive index of the overlay layer in the same mode as the single mode, and the other is the sensitivity to the rate of change of the thickness of N with respect to the adhesion layer, which can be defined as:

$$s\{d_A\} = \partial N / \partial d_{Ac} \tag{9}$$

The third indicator is the effective thickness d_{eff} of the sensor, which is calculated as [30]:

$$d_{e\,f\,\bar{f}} = d_{\,\bar{F}} + \sum_{J=\$\;\;C} \Delta Z_{,\;\;I} \tag{10}$$

3. Experiments

3.1. Experimental Equipment and Materials

The grating waveguide angle modulation biosensor uses the optical waveguide mode theory. The sensor structure is shown in Figure 4. It consists of a flat grating waveguide, a substrate and a cladding. Wherein S represents a substrate, F represents a waveguide, and C represents a cladding. The word “grating” represents a grating. The corresponding parameters of the grating waveguide mode biosensor in this experiment are: the grating adopts the imprint method, the refractive index of the waveguide layer is nF = 1.65, the refractive index of the base glass layer is nS = 1.37, the waveguide width is Lx=4 mm, and the period is Λ= 404.8 nm. The biomolecule solution to be detected is a cladding layer.

In addition, in the detection and biochemical analysis using OWLS, the purpose of the incident angle modulation by rotating the storage platform is a generally adopted method. In this paper, the purpose of modulating the incident light source is achieved by rotating the laser source. At the same time, in order to reduce the system volume, this paper also uses MEMS micromirrors for angle modulation. The MEMS micromirror is produced by Wuxi Micro Technology Co., Ltd., with a chip size of 1.45 mm × 1.25 mm and a micromirror size of 440 mm × 440 μm. The electro-thermal driving method has a driving voltage of 0~7 V, and the supporting arm is a folded double-S

micromirror driver structure, and the physical rotation angle of the mirror surface is ±8°[12], and the structure is as shown in Figure 5.

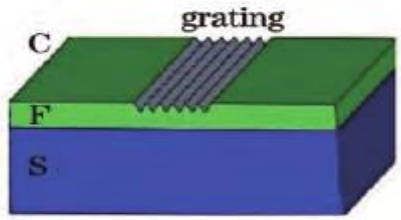

Figure 4. Grating waveguide mode biosensor structure

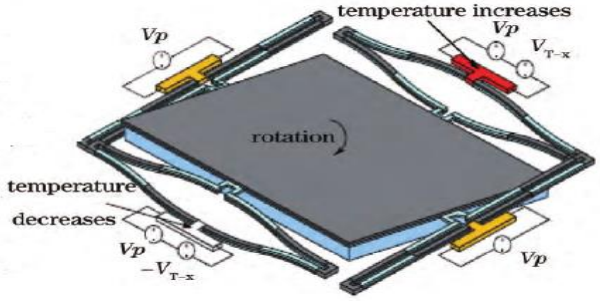

Figure 5. Folding double S-type structure micromirror

3.2. Testing Process

The experiment firstly uses a laser with a wavelength of 640 nm, which is incident on the MEMS micromirror through a collimating mirror, and rotates the micromirror to couple the incident light into the waveguide layer through the light; then the grating waveguide chip is placed in the sample cell; the sample cell is fixed on the two-dimensional displacement platform, different observation points on the surface of the chip are then selected by adjusting the two-dimensional displacement platform. To verify the sensitivity of the biosensor, the solution concentration was measured using a direct method. In the experiment, 12 different concentrations of glucose solution were prepared as the environment solution to be tested, and the concentrations were 10000, 5000, 2500, 1250, 625, 312.5, 156.25, 78.125, 39.0625, 19.53125, 9.765625, 4.8828125. The unit is mg/m L. The detection mode is detected in two modes, TE mode and TM mode.

4. Result Analysis

4.1. Waveguide Film Thickness and Sensitivity

In order to obtain the relationship between the refractive index sensitivity of the three-layer slab waveguide and the thickness of the waveguide, the refractive index of the substrate and the cladding, the refractive index of the waveguide layer is nF = 1.65, the refractive index of the base glass layer is nS = 1.37, and the waveguide width is Lx=4. Mm, incident wavelength is 640 nm. Adjusting the angle of incidence produces guided mode resonance. 1) When nC < nS, take nC = 1.3; 2) When nC> nS, take nC = 1.5, through computer simulation, the result is shown in Figure 6.

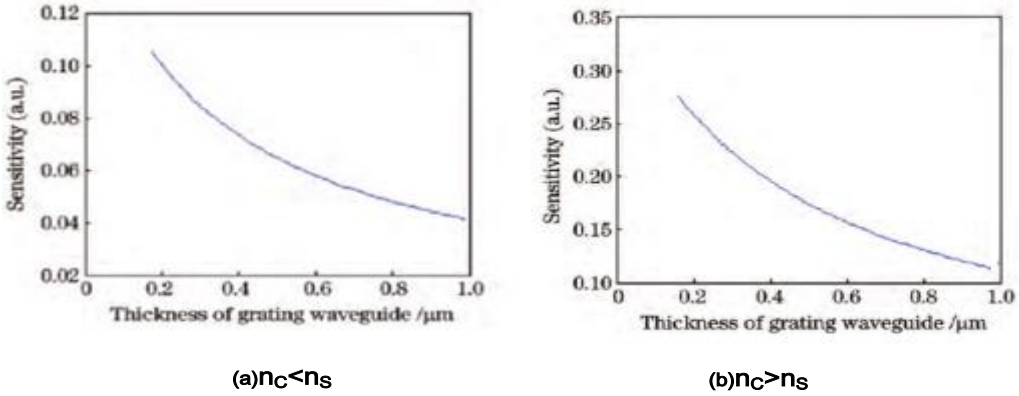

Figure 6. Waveguide refractive index sensitivity and waveguide thickness

It can be seen from Figure 6 that the polarization sensitivity of the flat-plate dielectric grating decreases with the increase of the waveguide thickness, regardless of nC < nS or nC > nS. When nC > nS, when nC > nS at the same waveguide thickness, the refractive index sensitivity is higher than that of nC < nS. When nC < nS, when dF = (dF)min = 163.1 nm, there is (∂N/∂nC)max = 0.19. When nC > nS, when dF=(dF)min= 163.1 nm, there is (∂N/∂nC)max= 0.29. In the above theoretical model, the relationship between the refractive index sensitivity and the effective refractive index is shown in Figure 7.

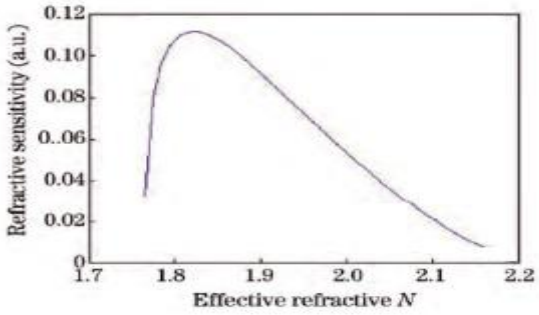

Figure 7. Relationship between refractive index sensitivity and effective refractive index of the waveguide

The refractive index sensitivity increases as the effective refractive index increases, and an extreme value appears, and the refractive index sensitivity begins to decrease gradually. Therefore, in order to make the grating waveguide have higher sensitivity, the detection should be performed in a mode near the maximum value of the refractive index sensitivity. As shown in Figure 7, the mode with an effective refractive index of around 1.82 makes the grating waveguide have higher sensitivity.

4.2. Relationship between Environmental Solution and Incident Angle

1) When TE mode is detected, when the concentration of glucose solution is in the range of 3~10000 mg/m L , the resonance angle and the concentration of the solution to be tested increase with the increase of concentration, which is basically linear, and the angle varies from 14.2 ° to 19.1. °, the lower limit concentration of the detection environment solution is 27.6 ng / m L.

2) In the direct detection of TM mode, when the concentration of glucose solution is in the range

of 1~600 ug/m L, the resonance angle and concentration increase with the increase of concentration, which is also linear, and the angle varies from 23.3 ° to 26.8. °, the lower limit concentration of the detection environment solution is 4.8 ng / m L.

4.3. MEMS Micro-mirror Analysis

The MEMS driver adopts a single-pin driving method, and the mirror surface is vertically displaced. When a pair of inverting differential voltages is applied to the two sets of driving arms, the MEMS mirror is rotated by the X-axis or the Y-axis. Through testing, the relationship between the driving voltage of the MEMS micromirror and the mechanical rotation angle and power is shown in Figure 8. There is a good linear trend between voltage and mechanical rotation angle. After a period of time, the thermomechanical oscillation of the MEMS micromirrors tends to be stable.

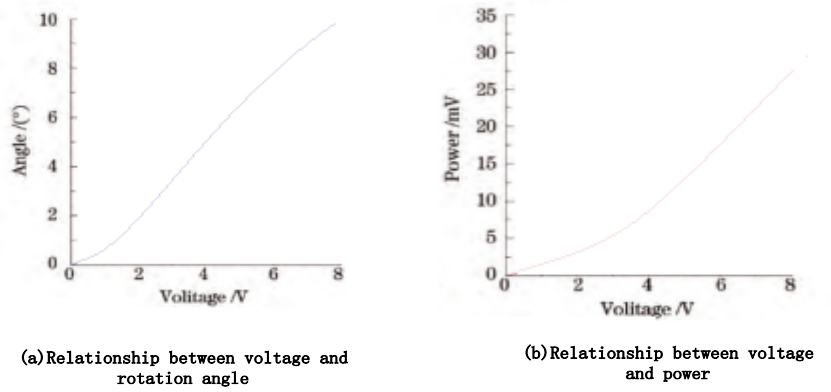

Figure 8. Voltage and rotation angle and power relationship

5. Conclusion

In recent years, with the development of biological science, information science and materials science, biosensor technology has developed rapidly. Biosensors have been used in many fields and are also widely used in food testing. Among them, modern optics plays a very important role in biosensors. Therefore, this paper introduces the application of a modern optical biosensor - optical waveguide mode spectral biosensor in the detection of children's food quality and safety. That is, optical waveguide lightmode spectroscopy (OWLS) biological sensor detection method based on microelectromechanical system (MEMS) micromirrors, which is a novel optical label-free method. The method utilizes a grating waveguide as a sensing chip, adjusts incident light by MEMS, and excites the guiding mode by grating coupling. The mass of the adsorbed material on the sensor surface is detected by angle modulation. The method has the advantages of simple structure, easy surface control, and label-free detection, thereby avoiding the destruction of the protein activity by the label. In the OWLS detection experiment, the direct immunosorbent reaction using surface-immobilized antibodies has a detection sensitivity of 1-2 ng/mL. The method directly detects the concentration of the glucose solution, and its detection sensitivity can reach 5ng/mL, which has higher sensitivity than the traditional immunological detection method. This method is valuable in the dynamic analysis of the material adsorption process on the surface of the sensor. The optical detection sensor based on this will be the main detection device in the future biological detection.

Funding

This article is not supported by any foundation.

Data Availability

Data sharing is not applicable to this article as no new data were created or analysed in this study.

Conflict of Interest

The author states that this article has no conflict of interest.

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