# DETECTION OF TEETH SURFACE DEFECTS USING FIBER OPTIC DISPLACEMENT SENSOR


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**DETECTION OF TEETH SURFACE DEFECTS USING FIBER OPTIC DISPLACEMENT SENSOR**

Field of Study: **FIBER OPTIC SENSOR IN DENTISTRY**

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# ABSTRACT

Fiber optic displacement sensor has been used recently for general medical diagnostics and in industry for the characterization and detection of defects in engineering components. However the lack of understanding of parameter performance of such a sensor in teeth has largely precluded its application to dentistry. This research describes sensor system adopting an intensity modulation technique that used a concentric type bundled optical fiber as a probe in conjunction with the real teeth, artificial teeth, hybrid composite resin teeth and mirror as the reflecting targets. The performance of the sensor is investigated on the linear range, sensitivity and the peak position. In this experiment, the sensor system is used for detecting defect on real teeth. A 2 mm hole is drilled to create defect on the teeth surface to stimulate dental cavity. The sensor probe is consequently fixed within the linear range of the displacement and the intensity of the collected light as a function of lateral movement (x and y axis) of the teeth surface is recorded while being maintained in perpendicular and constant in axial position (z axis). The image surface of the defect teeth is reconstructed from the recorded output voltage using the MATLAB software. This method is practically suitable when used in conjunction with a micro-computed tomography system in determining the defect of surface profile of a 3D tomography object being inspected prior to the image reconstruction.

# ABSTRAK

Sensor anjakan gentian optik telah mula digunakan sejak kebelakangan ini dalam diagnostik perubatan dan di dalam industri untuk pencirian dan pengesanan kecacatan pada komponen kejuruteraan. Walaubagaimanapun, pemahaman terhadap prestasi paramanter sensor dalam bidang pergigian masih kurang mewujudkan batasan untuk aplikasi dalam bidang perubatan. Penyelidikan ini menggunakan fiber optik yang disusun secara sepusat sebagai kuar dengan teknik keamatan modulasi yang disasarkan pada gigi sebenar, gigi tiruan, gigi hybrid kandungan resin dan cermin. Prestasi sensor diselidiki pada keluaran sensor untuk menentukan julat kelelurusan, kepekaan, dan kedudukan puncak.  Dalam uji kaji ini, sensor digunakan untuk mengesan kecacatan pada gigi sebenar. Satu lubang 2 mm dibuat untuk mewujudkan kecacatan seakan lubang pada permukaan gigi. Penguji sensor dipasang tetap di dalam julat kelelurusan dan kepekaan sensor di kira di paksi sisi (paksi x dan y) sepanjang permukaan gigi serta paksi tegak (paksi z) dibiarkan tetap semasa sepanjang penguji sensor dipasang.  Imej permukaan gigi yang rosak digambarkan daripada nilai-nilai voltan keluaran yang direkod menggunakan perisian MATLAB. Secara praktikal, kaedah ini sesuai dengan satu sistem pengkomputeran tomografi bagi menentukan kerosakkan permukaan obejek 3D tomografi yang sedang diperiksa sebelum pembentukan imej.

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## LIST OF SYMBOLS AND ABBREVIATION

| | |
|---|---|
| FODS | Fiber Optic Displacement Sensor |
| HA | Hydoxyapatite |
| FA | Fluroapatite |
| FOTI | Fiber Optic Transillumination |
| QLF | Quantitative Light-Induced Fluorescence |
| DiFOTI | Digital Image Fiber Optic Transillumination |
| ECM | Electrical Conductance Measurement |
| SMF | Single Mode Fiber |
| MMF | Multimode Fiber |
| He-Ne | Helium-Neon |
| POF | Plastic Optical Fiber |
| CW | Continuous Wave |
| PMMA | Polymethylmethacrylate |
| LED | Light Emitting Diode |
| E-H | Electron-Hole |
| OCT | Optical Coherence Tomography |
| 2D | Two-Dimensional |
| 3D | Three-Dimensional |

# CHAPTER 1

# INTRODUCTION

## 1.1 Overview

Biophotonics is the study of the interaction of light with biological materials. Recently, development of biophotonics is focused on translating discoveries in the sciences around lasers, light, and imaging into useful medical equipments (Wilson et al., 2005). Among the biophotonics applications, optical fiber sensors possess various advantages in electrical sensing technologies, which make it interesting for many other areas. The advantages include the safety in chemically or explosive environment and most important is low susceptibility to electromagnetic interference.

Fiber optic based sensor is commonly used in control and monitoring systems for material temperature, strain, deformation and others field. These fields use the work range in fiber optic displacement sensor (FODS) mechanism. The mechanism of FODS can be categorized into three categories, based on 1) wavelength modulation, 2) phase modulation, and 3) intensity modulation. The first and second categories have good accuracy but short dynamic range, while the third category has lesser accuracy but longer dynamical range.  Moreover the third category is lower in cost and contributes measurements that are easily interpretable.

FODS based on intensity modulation can be designed by using transmission technique (Harun et al., 2010), where the coupling loss of power light is measured at two-movable ends of the fiber, and the reflection technique, where the coupling loss factor of power lights is measured due to the reflection of the moving object.

In the reflection technique, the sensor can be divided into two forms: 1) single mode fiber (SMF) and 2) multi mode fiber (MMF) coupler. This research conducted an experiment using the MMF coupler for displacement sensor on artificial teeth, hybrid composite resin teeth, and human teeth.

Furthermore, this study investigates the potential of intensity modulated FODS imaging for the identification of dental cavity. In this research, the discussion is on preliminary results in the detection of flaws on various teeth surfaces, as well as measurement of the flaws which are represented by drilled holes that simulate dental cavities. Measurements are obtained by launching a light beam onto a tooth surface via a bundled fiber and collecting the reflected light from the surface via a silicon photodetector. The output voltages are measured as a function of lateral distance to produce the tooth surface profile. The results show a clear distinction between the flawed and unflawed tooth region. This method is simple, cost effective, non-destructive and does not require ionizing radiation, hence providing great potential for quality assessment of the impressional mapping methods in restorative dentistry or as an alternative non-impressional method.

## 1.2 Problem Statement

FODS has become popular in sensing application in medicine (Jensen et al., 1989). The sensor relies on the change in retroreflectence of light into a fiber because of the movement of a target surface. The unique features of FODS have spread interestingly in dentistry applications that provide with capabilities, less cost and performance benefits over existing sensors. However, in dentistry there are many challenges in bringing optical sensing to the clinical environment. The information on the use of FODS and also their displacement properties on real human teeth information are still not readily available. Researchers require these properties to enhance the application of FODS in dentistry.

Tooth decay, which is also called dental cavity or dental caries, is the destruction of the outer surface of a tooth. Dental cavity occurs when the plaque bacteria that live in the mouth sticking to tooth surface use the sugar from eaten food to produce acid. The acid will dissolve the minerals on the tooth surface, creating holes and weak spots.

The understanding of the dental cavity process has continued to develop on the risk and modifying factors for cavity process, comes an opportunity to promote preventative therapies that encourage the remineralisation of non-cavitated lesions (Iain A, 2006). This will results in prevention of dental cavity with the preservation of tooth for its surface and function. The failure to detect early stage of dental cavity, leaving the cavity at the deep enamel has resulted in poor outcomes for remineralisation therapies. In conclusion, the early detection system for dental cavity may help dentist to prevent the lost of the teeth due to late detection of the cavity.

## 1.3 Objective of the Study

In this study, the performance and the displacement properties of the sensor on real human teeth by using multimode bundled type of fiber optic sensor are investigated by measuring the reflected light in voltage by FODS. The goals of the study are as follows:

(i) To design FODS to be applied on the real human teeth.

(ii) To design FODS for the detection of surface defect that simulate dental cavities on real human teeth

(iii) To demonstrate the designed FODS experimentally.

(iv) To analyze and evaluate the performance of the designed displacement sensor.

## 1.4 Hypothesis

The properties of displacement on real teeth such as displacement response, sensitivity, and dynamic range are much similar towards each other with excellent displacement properties. Furthermore, the detected voltages by the proposed method with an analysis can measure the surface defect areas that simulate dental cavities on real human teeth.

## 1.5 Significance of the Study

The significance of the study is to propose a detection system for teeth surface defect for profile inspection using a simple displacement fiber optic bundled sensor system. The FODS is interesting due to its inherent simplicity, small in size, mobility, wide frequency capability, extremely low displacement detection limit and ability to perform non-contacting measurement. The proposed method of detection of teeth surface defect is accurate to overcome the shortcomings of currently available methods. The novel detection of teeth surface defect is highly advantages over the conventional visual method that needs experience from the dentist since earlier detection can prevent the lost of the teeth.

## 1.6 Scope of the study

The scopes of this study are as follows:

- A concentric type of bundled fiber optic consisting of both transmitting and receiving fiber as the probe, a red He-Ne laser with 632nm wavelength and a silicon photo detector.
- The real human teeth (namely molar and canine), artificial teeth and hybrid composite resin teeth are used as samples.
- A hole of 2.0 mm to 2.5 mm diameter is drilled on the teeth to create a defect on the teeth surface to simulate dental cavity. The deep of the hole is not significant in this study.

## 1.7 Organization of Research Report

Chapter 1 introduces the overview of optical fiber based sensor in medical applications and the importance of early detection for dental cavities. The problem statement, objectives of the study, hypothesis, significant of the study, and scope of the study are stated in this chapter.

Chapter 2 presents the literatures that are related to teeth as a hard tissue and optical fiber. Furthermore, the usage of fiber optic sensors in dentistry is reviewed. In additional, the explanation of the component for FODS used in this research is also discussed.

Chapter 3 covers the performance of fiber optic displacement on real human teeth. The explanations cover the introduction, experimental method, as well as results and discussion.

Chapter 4 covers the usage of FODS as detection system for teeth surface defect that simulate dental cavity. The explanations cover the introduction, experimental method, as well as results and discussion.

Finally, the research report is concluded in Chapter 5.

# CHAPTER 2

# LITERATURE REVIEW

## 2.1 Introduction

This chapter reviews the fiber optic sensor technology as well as the application of FODS in medicine, focusing on dental hard tissue.

## 2.2 Dental Hard Tissue

Human teeth (singular tooth) consist of three primary tissue components which are the enamel, dentine and pulp cavity as shown in Fig. 2.1. Human usually have 20 primary teeth (baby) and 32 permanent teeth (adults). A human tooth consists from incisors, canines, premolars and molars. Incisor is a tooth adapted for cutting or gnawing, located at the front of the mouth along the apex of dental arch. Molar is a tooth with a broad crown used to grind the foods, located at the back of the mouth. Enamel is the hardest and most highly mineralized substance of the body. Enamel is one of the major tissues in single tooth, together with the dentine, cementum and pulp cavity. Dentine is the substance between enamel or cementum and the pulp chamber. The dentine secreted from the odontoblast of the dental pulp.

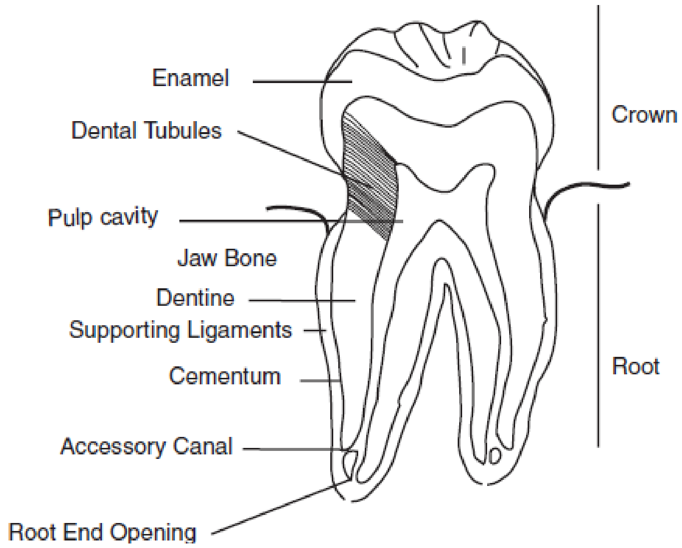



Figure 2.1: The section of a human tooth (Adapted from Thomas et al., 2005)

### 2.2.1 Composition of Dental Hard Tissue

Enamel and dentine tissue consists of mineral, lipid, protien and water. Enamel and dentine are very different in their structure, but consists of similar components. Table 2.1 shows approximate composition for enamel and dentine as volume percent of each of the components involved. The organic mineral and water components are the main role in enamel compared to dentine. The mineral in the teeth is composed of a highly substituted hydoxyapatite (HA). This is actually similar with our bones. However, the substitutions in the mineral crystal lattice disturb the structure and make it much more acid soluble than the pure HA (Nelson el al., 1983). The primary major substitution is carbonate that replaces some of the phosphate leading to an acid soluble mineral (Nelson et al., 1982).

Table 2.1. Approximate composition of enamel and dentine as volume percent of total tissue

| Component | Enamel approximate percent by volume | Dentine approximate percent by volume |
| --- | --- | --- |
| Carbonated HA | 85 | 47 |
| Water | 12 | 20 |
| Protein and lipid | 3 | 33 |

The proteins in enamel are primarily a very thin covering of the individual crystals and comprise approximately half of the organic material. While lipid is the other half of the organic material in enamel (Nelson et al., 1987). Dentine has a similar mineral composition, although the carbonate content is much higher, the mineral is much more soluble, and a large component of the tissue is collagen type I with about 10% of the protein comprised of a range of non-collagenous proteins (McCormack et al., 1995).

### 2.2.2 Dental Caries and Detection Methods

A tooth cavity is an oral disease that comes from plaque bacteria. Dental plaque (bacteria) that present in and around the teeth for period of time will produce acids that lead to dental cavities. Dental cavity is a transmissible infectious disease but can be prevented by noticing the presence of cavities in the mouth. Frequent ingestion of sweets and low saliva flows is the cause of the development of cavities in the presence of dental plaque.

Researchers have suggested evidence that lead to caries detection process where numerous modifiers tending to push the mineral equilibrium in one direction to another, i.e. towards remineralisation or demineralisation (Holt et al., 2001). The modifiers of this system are well known and are summarised in Table 2.2 with Fig. 2.2 presenting an overview of the dynamics of the caries process.

Table 2.2. Risk and modifying factors for caries (Adapted from Iain et al., 2006)

| | |
|---|---|
| **Primary risk factors** | |
| **Saliva** | |
| 1 | Ability of minor salivary glands to produce saliva |
| 2 | Consistency of unstimulated (resting) saliva |
| 3 | pH of unstimulated saliva |
| 4 | Stimulated salivary flow rate |
| 5 | Buffering capacity of stimulated saliva |
| **Diet** | |
| 6 | Number of sugar and acid exposure per day |
| **Fluoride** | |
| 7 | Past and current exposure |
| **Oral** | |
| 9 | Differential staining |
| 10 | Composition |
| 11 | Activity |
| **Modifying factors** | |
| 12 | Past and current dental status |
| 13 | Past and current medical status |
| 14 | Compliance with oral hygiene and dietary advice |
| 15 | Lifestyle and socioeconomic status |

Figure 2.2: Demineralisation and remineralisation cycle for enamel caries (Adapted from Iain et al., 2006)

| | | | Critical pH of HA | | Critical pH of FA | | | |
|---|---|---|---|---|---|---|---|---|
| pH | 6.8 | 6.0 | 5.5 | 5.0 | 4.5 | 4.0 | 3.5 | 3.0 |
| Production of HA and FA Calcium & Phosphate in saliva | | | Demineralisation Dissolution of HA FA forms if fluoride available Remineralisation FA reforms | | | Acid dissolution of crystal | | |
| 8.0 | 6.8 | 6.0 | 5.5 | 5.0 | 4.5 | 4.0 | 3.5 | 3.0 |
| Formulation of calculus | | Remineralisation Demineralisation | | Caries | | | Erosion | |
| HA is hydroxyapatite | | | | | | FA is flurorapatite | | |

The diagnosis methods for dental caries are based upon the measurement of a physical signal – these are surrogate measures of the caries process. Examples of the physical signals that can be used in this way include X-rays, visible light, laser light, electronic current, ultrasound, and possibly surface roughness (Pretty et al., 2004). The device for dental cavity detection should be able to initiate and receive signal as well interpret the strength upon signal detection.  Table 2.3 demonstrates the physical principles and the detection methods that have taken advantages of them.

The early detection of dental cavity can help patient to prevent and preserve the teeth. Practitioners agree that identifying and preventing or treating dental cavity at their early stages is one of the most important things in dentistry. (Fontana et al., 2010). The late detection of dental cavity may lead to the continue growing of the lesion that will destroy the enamel.

Table 2.3. Methods of caries detection based on their underlying physical principles with its advantages and disadvantages. (Modified from literature Angmar et al., 1998; Iain et al., 2006 )

| **Physical Principle** | **Application in caries detection** | **Advantages** | **Disadvantages** |
|---|---|---|---|
| X - rays | Digital subtraction radiography<br>Digital image enhancement | (1)Significantly lower resolution, with potential of 256 grey level | (1)Exposure of radiographic dose to patient |
| Visible light | Fiber optic transillumination (FOTI)<br>Quantitative light-induced fluorescence (QLF)<br>Digital image fiber optic transillumination (DiFOTI) | (1)Detection of caries on all surface; useful at proximal lesions<br>(2)Less time needed during diagnosis<br>(3)Economic | (1)The diagnosis is rather subjective due to analysis is based on visually of doctor<br>(2)Training is required to be competent to used the device |
| Laser light | Laser fluorescence measurement (DiagnoDent) | (1)Can measure the degree of bacterial activity | (1)Need only to apply on clean and dry teeth only<br>(2)May lead to false reading due to existing of stain and plague on teeth |
| Electrical current | Electrical conductance measurement<br>Electrical impedance measurement | (1)Simple measurement by exploiting the conductivity of lesion and tissue | (1)Surrounding conditions such as temperature and thickness of the teeth may affect result |
| Ultrasound | Ultrasonic impedance measurement | (1)Images of tissue can be acquired by collecting the reflected sound wave | (1)Low in sensitivity and specificity when measurement<br>(2)Discriminate between cavitated and non-cavitated inter-proximal lesions |

## 2.3 Optical Fiber Technology

Fiber optic was first envisioned as optical elements in yearly 1960s. It was believed that scientist well acquainted with the microscopic structure of the insect eye who realized that an appropriate bundle of optical waveguides could be made to transfer an image and the first application of fiber optic to imaging was conceived. Fiber optic is made from a glass or plastic, which will carry lights along its length. Fiber optic mostly is used in telecommunications, which allows transmission of signal over longer distances and at higher bandwidths (data rates) compared with others method of communications. The benefit of fiber optic is less loss compare to metal wires and also immune to electromagnetic interference. Moreover, fiber optic that wrapped in bundles may used to transmit images, thus allowing viewer to view it in a tight space.

### 2.3.1 Principles Operation of Fiber Optic

The main theory of the principles operation of wave guidance in fiber optic is about the total internal reflection inside the fiber optic cable. Theoretically if a light rays passed from one medium with a refractive index of $n_1$ to another medium with a refractive index of $n_2$ where $n_2$ is larger than $n_1$ for example air to glass, the refracted wave in the second medium will bend towards the normal. If $n_2$ is less than $n_1$, the wave will be bent away from the normal to the surface. However, there is one instance, where the penetrating ray will not deviate from its original path at all, that is, if it enters the medium perpendicular to the surface or head on. In the case where $n_2$ is less than $n_1$ there will be a point where an incident where the ray will be totally internally reflected where it will not enter the second material. There is an angle where the refracted wave will be placed between the two media and also will not enter the second material, is called the critical angle $\theta_c$. This means that all light rays will be reflected at the boundary if all the angles of incidence is greater than the critical angle. The further increased of incident angle of the ray, the reflected wave is actually turned back into the first medium, and this achieved the total internal reflection. The critical angle can be defined by putting $\theta_2$ = 90 deg in the Snell's law of reflection.

$$n_1 \sin \theta_c = n_2 \sin 90\, deg \quad (2.1)$$

$$\theta_c = \sin^{-1} \frac{n_2}{n_1} \quad (2.2)$$

Optical fibers are the fiber media that guides the light ray which can be made from plastic or glass. Since the plastic fibers have more loss and tend to have low bandwidth so glass fibers are more common to be used. Commonly, fiber is made up of a core, cladding and a jacket. The core is the centre or the actual fiber where the light propagates and it has dimensions on the order of 5 to 600 µm. The cladding surrounds the core and has an index of refraction lower than that of the core, in this way light will propagate through the core by means of internal reflection. The jacket is surrounding the cladding and used to protect the entire optical fiber.

### 2.3.2 Types of Optical Fibers

There are several types of fibers optic used commercially in the areas of communication, medical and sensors which all the fibers optic are made differently for their usage. Low attenuation fibers optic is required to maximize spacing of repeaters or amplifiers in transmission link. In sensor technology, there are two most common operating classes of fibers optic based on their properties; SMF and MMF. SMF are classified into three different types of fibers i.e. step index, dispersion-shifted and dispersion-flattened. MMF are grouped into two types; step index and graded-index fibers.

SMF are most available with step-index profile, which has a higher index of refraction for the core compared to the cladding. This allows the total internal reflection to take place at the core/cladding interface. The peak difference in refractive indices of the core and cladding is estimated around 1 to 20% in this fiber optic. The index difference determines the numerical aperture of the fiber optic, which in return determines the amount of light that the fiber optic can be collected from optical source. The SMF usually has a very small core diameter (5-10μm) and allows only one mode to propagate effectively in the core.

MMF let the light to propagate in the same wavelength along different ways through a fiber which causes the waves to arrive at the opposite end of fiber at different time. These fibers are not common to be used for optical communication but very useful for sensor application. The major attraction of MMF is due to its relativity easy in collecting light because of larger core size compared to SMF. Typical numerical apertures (NA) of this fiber range from 0.2 to 0.4 and their core size diameter usually ranges from 100μm to 200μm.

The experiment carried out in this research is using the MMF plastic bundled fiber as the probe to measure the signal from receiving fiber by moving the probe away from the zero point.

## 2.4 Fiber Optic Sensor in Dentistry

The applications and technology of optical fiber has progressed very rapidly nowadays; however, they still need further improvement (Angmar et al., 1998). Fiber optics offers the advantages of flexibility of beam manipulation. The intention of fiber optic based sensor is due to the safe, rapid and non-invasive testing method of clinically relevant physiological variables. Furthermore the advantages includes electrical isolation, physical flexibility and needless of electrical power for driving sensor unit. Researchers find application in designing customized probes that are tailored for specific applications (Jeon et al., 2003).

Iain et al. suggest enhancing the visual techniques using the fiber optic sensor technology (Iain et al., 2006). The basic of visual inspection of carries is based upon the phenomenon of light scattering. The colour of teeth is influenced by the underlying dental shade where when is normal, visible light, this appear as a whiter area-the so called white spot. But, when the enamel is disrupted, the penetration photons of light are scattered which results in an optical disruption. The suggested FOTI takes advantages of these optical properties of enamel and enhances them by using high intensity white light that is presented through a small aperture in the form of a dental handpiece. Light is shone through the tooth and the scattering effect can be seen as shadows in enamel and dentine. The advantages of this method are the ability to help discriminate between early enamel and early dentine lesions. Furthermore, it can be used for the detection of caries on all surfaces and is particularly useful at proximal lesions make it one would expect this method to be at least as effective as a visual examination.

Milczewski et al. used polymer optical fiber on the measurement of orthodontic forces (Milczewski et al., 2007). During orthodontic treatment, forces are transferred to the teeth by using appliances designed to displace teeth a prescribed amount in desired direction. The suggested polymer optical fiber had allow to control the applied forces where at conventional method too much applied force that displaces a tooth or segment of teeth may caused permanent damage at periodontal ligament. The researcher exploited the structured properties of polymer photonic crystal fibers to demonstrate improved sensitivity to applied pressure and apply it to orthodontic tooth measurement. Hence, to develop polymer optical fiber optic sensor for measurement of the magnitude of forces received by a single tooth during orthodontic measurement. Polymer optical fibers have been used in sensing application and characteristic such as flexibility, electrical and chemical immunities make them attractive for biomedical sensing (Jiang et al., 2002).

## 2.5 Fiber Optic Displacement Sensor (FODS)

Displacement sensors are used wherever it is necessary to acquire the exact position of two parts that can be moved relatively to one another. Currently, there are several types of displacement sensors in the market. For instance, optical and magnetic displacement sensors are used for detecting the amount of a linear displacement in stages of machine tools and three-dimensional measurement.

In a FODS, two methods are commonly adopted, namely laser interferometry and reflective intensity modulation techniques. Laser interferometry is based on the fringe counting method and has high resolution and stability, however it wavelength of light limit its precision and stability. In contrast, the reflective intensity modulation technique is much more simple method for non-contact displacement measurements providing high resolution and stability (Zivanovic et al., 1997).

The reflective intensity modulation type of sensor the reflected light from the target sample is coupled back into a fiber from a reflecting surface and the power is compared to a portion of the power emitted from the light source. The interest of this type of sensor in dentistry application due to its simplicity, mobility, small in size, very low displacement detection limit and ability to perform non-contact measurement.

The main element that used in FODS can be divided into 3 parts, namely, 1) He-Ne laser, 2) plastic bundled fiber and 3) silicon detector.

### 2.5.1 Helium-Neon (He-Ne) Laser

A He-Ne laser, is a type of small gas laser. He-Ne laser operates at 632.8nm wavelength, in the red portion of the visible spectrum make it popular to be used in industrial and often used in laboratory demonstration of optics. The diagram of He-Ne laser is shown in Fig. 2.3.

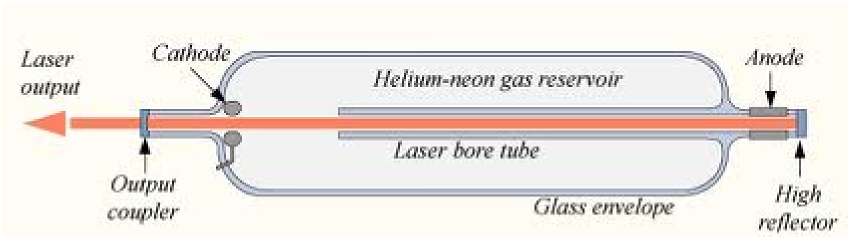



Fig 2.3: Helium-Neon (He-Ne) laser (Adapted from Wikipedia, 2012)

The gain medium of the laser, as referred to its name, is a mixture of helium and neon gases, in a 5:1 to 20:1 ration, contained at low pressure Helium at 1 torr and Neon at 0.1 torr in a glass envelope. An electrical discharge through an anode and cathode at the end of the glass tube with voltage around 1000 volts provided the energy or pump source to the laser. A current of 5 to 100mA is typical for continuous wave (CW) operation. The optical cavity of the laser typically consists of a plane, high-reflecting mirror at one end of the laser tube, and a concave output coupler mirror of approximately 1% transmission at the other end.

The interaction of visible light in the red region of He-Ne laser with hard tissue modification in dentistry can be described in Table 2.4 (Fried et al., 1995). The absorption coefficient is about the measurement of the level of absorption occurs in specific tissue by a specific wavelength of laser light. The visible light in the red region of He-Ne laser can be considered as low absorption rate compared to others visible light (Featherstone et al., 2001). From the table also, it can be seen that at the green end of visible light, the scattering coefficient in enamel is relatively high and this reduces towards the red. The absorption and scattering coefficient must be taken into account when determining displacement properties of the sensor.

Table 2.4. Selected optical properties of dental hard tissue (Adapted from Featherstone et al., 2001)

| Wavelength | Absorption coefficient, $cm^{-1}$ | | Scattering coefficient, $cm^{-1}$ | |
|---|---|---|---|---|
| | Enamel | Dentine | Enamel | Dentine |
| 543nm (green) | <1 | 3-4 | 105 | 280 |
| 632nm (red) | <1 | 3-4 | 60 | 280 |

### 2.5.2 Plastic Bundled Fiber

Plastic optical fiber, abbreviated POF, typically uses polymethylmethacrylate (PMMA), a general purpose resin as the core material, and fluorinated polymers for the clad material. The structure of POF can be described in Fig. 2.4 where in large diameter, 96% of the cross section is the core that allows light transmission. The POF consider as consumer fibers where the cost of POF, associated optical links, connectors and installation costs are low. POF is having much larger core diameter compared to others fiber as shown in Fig. 2.5. Most POF used 1000μm in diameter, with a core diameter of 980μm. Due to this large diameter, transmission is possible even if the ends of the fiber are slightly soiled or damages, or if the light axis is slightly off centre. As a result, a part such as optical connector is made cheaper and easy to be installed.

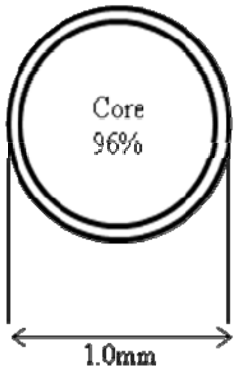



Fig. 2.4: Structure of plastic optical fiber (POF)

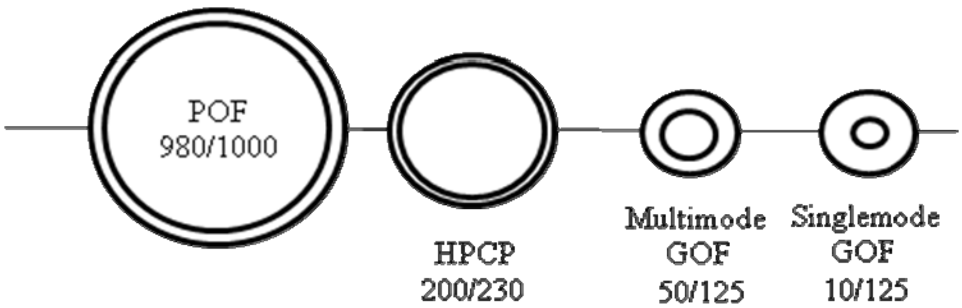



Fig. 2.5: Size of plastic optical fiber (POF) compared to other types of fiber

POF is strong and very difficult to bend and within 25mm radius bending the loss is very small. Due to that, it can be installed within walls or other narrow areas and suitable for lighting in tight locations. Furthermore the installation is simple and easy for beginner users. Normally, a red 650 nm light emitting diode (LED) is used as the light source. Since the light source is within the visible light spectrum, users can protect the eyes from directly viewing the light beam.

For sensing applications, the POF is normally bundled together to form a bundled fiber. This bundled fiber can be classified into several types such as pair and concentric types. This bundled fiber will be used in this experiment.

### 2.5.3 Silicon Detector

As for its name, silicon is used as the detector due to the moderate band gap of the semiconductor is 1.12eV when compared to the thermal energy at room temperature of kT = 1/40eV. The energy to create an electron-hole pair is 3.6eV, compared to the ionization energy of 15eV in Argon gas, leading to an ionization yield for minimum ionizing particles of about 80 electron-hole (E-H) pairs per micron. Thus, about 23000 E-H pairs are produced in the customary wafers thickness of 30μm, and collected in about 30ns without a gain stage.

The wafers are normally n-type with a high resistivity of about 5kΩ-cm and with a low-resistivity p-implant in form of pads, strips or pixels to create junction. With a reverse bias of less than 100V, the detectors can then be fully depleted such that only the thermally generated current contributes to the leakage current. The area of the detectors are limited to the standard wafer sizes used in high-resistivity processing by industry, which has increased the wafer size recently. Larger area detectors are now routinely made by assembling and wire bonding several detectors into so-called ladders, with fairly ling readout strips.

The silicon detector is used in this experiment to detect and measure the power of the visible light. The detectors convert the photon intensity to the voltage. The lock-in amplifier is also used together with the silicon detector in the conducted experiment.

# CHAPTER 3

# PERFORMANCE CHARACTERISTICS OF FIBER OPTIC DISPLACEMENT SENSOR (FODS) ON REAL TEETH

## 3.1 Introduction

FODS are commonly constructed from plastic multimode optical fibers, which offer benefit of low optical signal transmission loss, low production cost, compact size and compatibility with optical fiber technology. These sensors engineering have been used in branches of science and engineering, as it is evident from the range of properties which has been sensed optically, ranging from light intensity, vibration, temperature, pressure, strain, liquid level, pH, chemical analysis, concentration, density, refractive index of liquids, teeth colorimeter and etc (Harun et al., 2011; Patil et al., 1997). The parameter of performance for FODS is depend on blind region, linear range, sensitivity and the peak position of optically sensed object (Patil et al., 2011).

Early studies approached the use of optical fiber sensors in medical is an empirical fashion, while at the same time demonstrating the potential of this technology to have a real effect in clinical industry. Rahman et al. (Rahman et al., 2011) proposed a simple intensity modulated displacement sensor for sensing salinity based on different concentration of sodium chloride (NaCI) in de-ionized water. While, Hong, et al. (Hong et al., 1988) has suggested a novel type of fiber optic displacement pressure transducer that has its pressure sensitive membrane and shows that this new type of pressure transducer is much more better that conventional transducer.

In this chapter, a simple design of a FODS is demonstrated using a multimode plastic fiber bundled based on reflective intensity modulation technique. The performances of this sensor are investigated by correlating the detector output with different reflectivity properties of hybrid composite resin teeth, artificial teeth and real teeth. The simplicity of the design, high degree of sensitivity, dynamic range and the low cost of the fabrication make it suitable for dentistry application.

### 3.2 Experimental Setup

Fig. 3.1 shows the schematic diagram of the experimental set-up for the fiber optic sensor to detect axial displacement with mirror and different type of teeth. The sensor consists of a fiber optic transmitter, mechanical chopper, fiber optic probe, mirror or different type of teeth, a silicon photo detector, lock-in amplifier and computer. The fiber optic probe is made of two 2 m long PMMA which consists of one transmitting core of 1mm in diameter and 16 receiving cores of 0.25mm in diameter, numerical aperture 0.5, core refractive index 1.492 and cladding refractive index 1.402. A red He-Ne laser ($\lambda$=633nm) is used as the light source with an average output power of 5.5mW, beam diameter of 0.8mm and beam divergence of 1.01mRads. Laser diode ($\lambda$=785nm) has a collimated output beam with 2.4 x 3.4 mm in beam diameter and output power of 20mW is also used as another light source in this experiment. This two laser is used due to investigate which light source is the most suitable at human teeth when is used as the target sample.

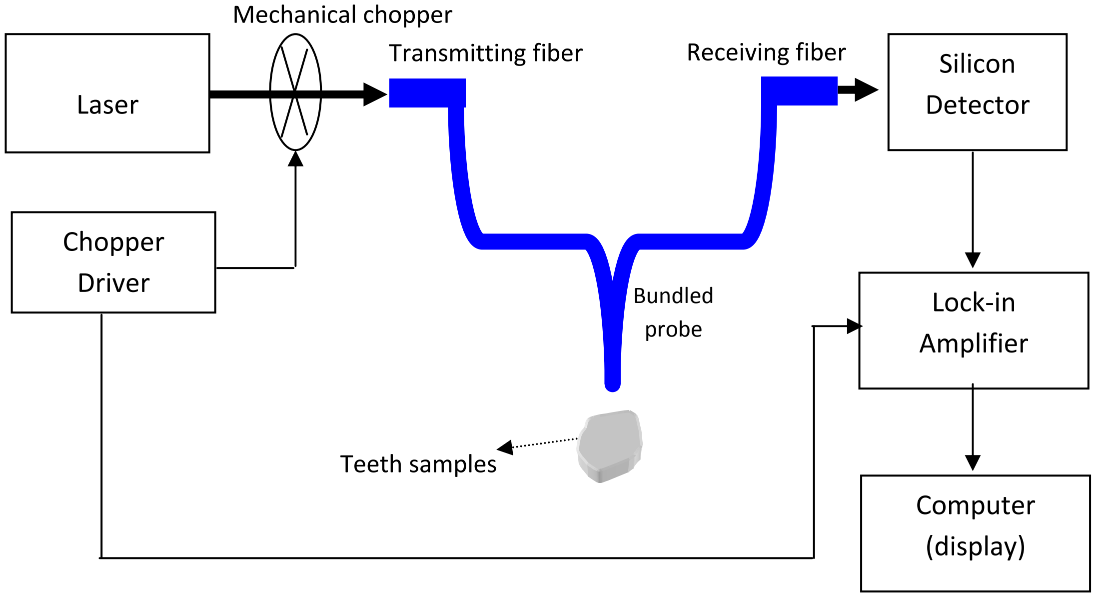



Fig. 3.1: Schematic experimental setup for displacement sensor using fiber bundled probe and teeth samples

The photodetector is a high speed silicon photodiode with an optical response which extends from 400 to 1100 nm, making it compatible with a wide range of visible light including the 633 nm visible red He-Ne lasers used in this set up. The light source is modulated externally by a chopper with a frequency of 113Hz as to avoid the harmonics from the line frequency which is about 50 to 60 Hz. The modulated light source is used in conjunction with a lock-in amplifier to reduce the dc drift and interference of ambient stray light.

The displacement of the fiber optic probe is achieved by mounting it on a micrometer translation stage, which is rigidly attached to a vibration free table. Light from the fiber optic transmitter (peak wavelength at 633nm) is coupled into the transmitting fiber. The signal from the receiving fiber is measured by moving the probe away from the zero point, where the reflective fiber is measured by moving the probe away from the zero point, where the reflective surface of samples and the probe are in close contact. The signal from the detector is converted to voltage and is measured by a lock-in amplifier and computer via RS232 using a Delphi software.

Mirror is used as the control sample while the other teeth samples are used consecutively as the reflecting target while measuring the output intensity by changing the position of the fiber optic probe from 0 to 4.6 mm in a step 50µm. The displacement profiles of the teeth samples are compared with a flat mirror due to its 100% reflectivity.

### 3.3 Results and Discussions

Fig. 3.2 shows the comparison of output voltage between He-Ne laser and laser diode against displacement on the human teeth sample. All results exhibit increasing at the front slope until it reached peak position and decrease at back slope when further displacement is increased. He-Ne laser shows more gradually increasing on the output voltage as the displacement is increased. In contrast, laser diode show less increasing on the output voltage as the displacement is increased. From the front slope until it reached to the peak position, the increasing output voltage value starts from 0.765mV until 1.050mV for He-Ne laser and 0.700mV until 0.7500mV for laser diode respectively.

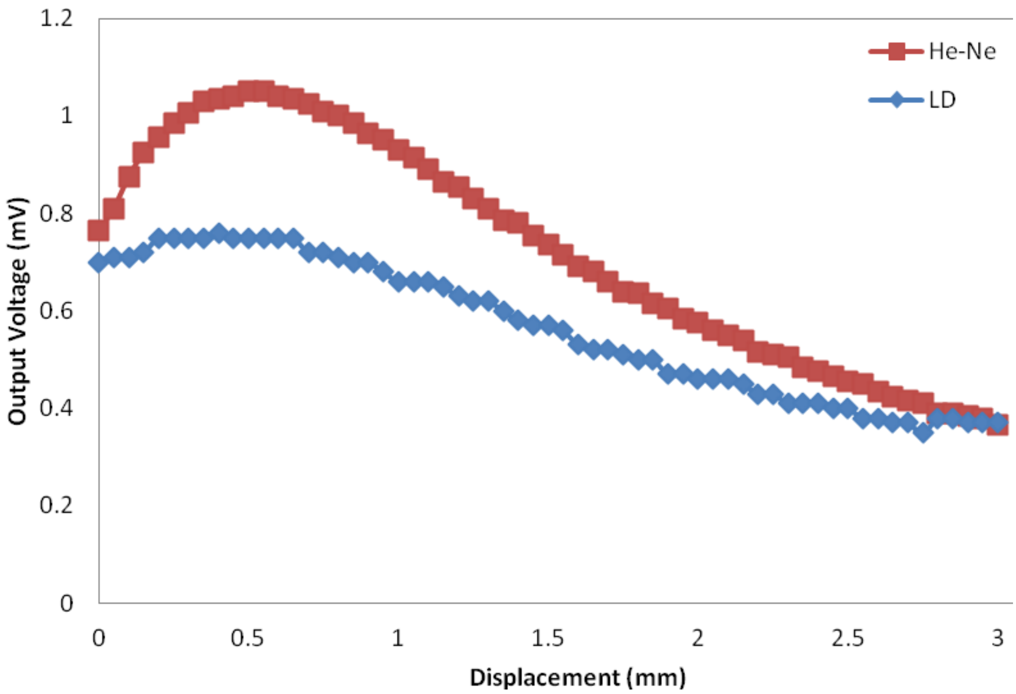



Fig. 3.2: Comparison of voltage output between Helium-Neon (He-Ne) laser and laser diode against displacement.

Table 3.1 summarize the features of the fiber optic displacement at the front slope for He-Ne laser and laser diode. The sensitivity of the sensor is determined by a slope of straight line portion of the curves. The peak voltage and sensitivity for He-Ne laser and laser diode are obtained at 1.05mV and 0.8214mV/mm and 0.76mV and 0.1929mV/mm respectively. Furthermore, Featherstone et al. based on fundamental laser interaction with dental hard tissue mentioned that visible light towards red, in this case He-Ne laser is less scattering where He-Ne laser is more reflective compared to the others light source (Featherstone et al.,2001). Further results and discussion will be focusing only at He-Ne laser as the light source throughout the experiment.

| No | Type of light source | Linear range, mm | Peak voltage, mV | Sensitivity, mV/mm |
|---|---|---|---|---|
| 1 | He-Ne laser | 0 - 0.3 | 1.05 | 0.8214 |
| 2 | Laser diode | 0 - 0.3 | 0.76 | 0.1929 |

Table 3.1. The features of the fiber optic displacement (front slope) on human teeth sample for various type of light source.

Fig. 3.3 shows the comparison between flat mirror and human teeth sample against displacement of the sensor. All curves exhibit a gradual and linear increase in the front slope while back slope follows an almost inverse square law relationship. The starting output voltage is 0.165mm and 0.565mV for flat mirror and human teeth sample respectively at the minimum displacement. Since the human teeth in general have a higher surface roughness and even more uneven than that of a mirror, the observed curves start at a much higher point compared to flat mirror. When the displacement is increased, more light will be collected by the receiving cores due to the increased size of the reflected cone of light. A further increase in the displacement leads to larger overlapping which results in a further increase in the output voltage as seen in the first part of the curve until it reaches the peak value. At this peak value, the reflected cone power falls within the surface area of the receiving fibers. Further increase of the displacement will results in a reflected cone size bigger than the size of receiving fibers. Therefore, only a fraction of the power of the reflected light will be detected, as seen in the second part of the curve. The different of front and back curve upon displacement for human teeth sample and flat mirror due to mirror has more light reflectivity compared to human teeth sample. However, this does not affect the performance of the sensor since the evaluation is based on the first slope and peak voltage of the displacement profile.

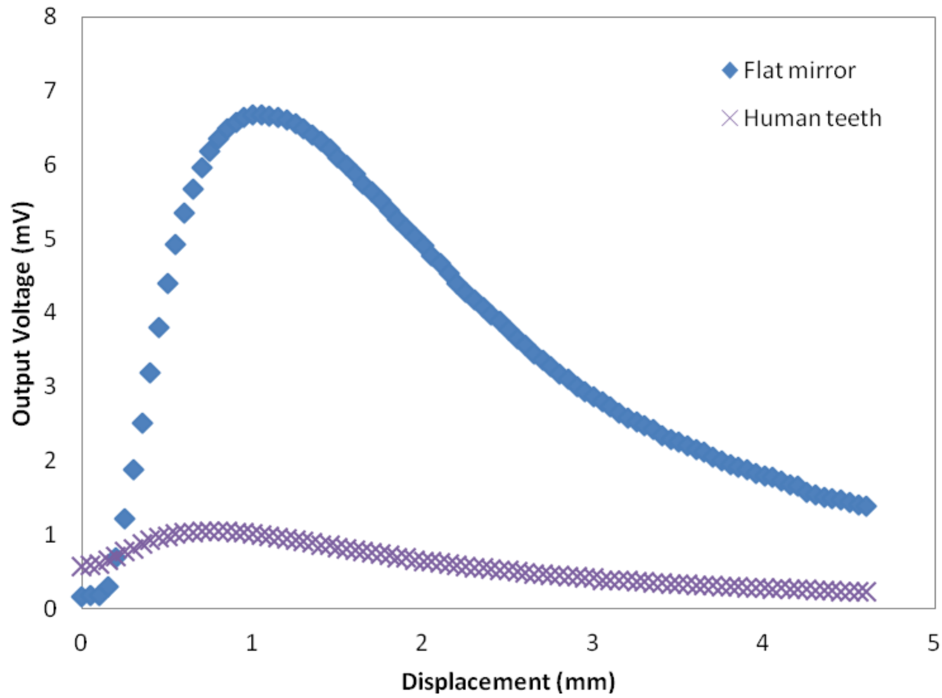



Fig. 3.3: Comparison between flat mirror and human teeth sample against displacement.

The features of the sensor for different teeth surface are summarized as shown in Table 3.2. Fig. 3.4 shows the reflected light intensity versus distance of the fiber optic probe from various reflecting teeth surfaces, namely molar, canine, hybrid composite resin and artificial. All curve again exhibit a maximum with a gradual and linear increase in the front slope while the back slope follows an almost inverse square law relationship. The received light intensity varies considerably among the various teeth surface due to different reflectivity on teeth surface. The reflectivity of molar, canine, hybrid composite resin and artificial teeth surfaces are obtained at 4.72, 4.18, 2.16, and 1.82 %, respectively by comparing the power of the light source before and after the reflecting surface. As shown in Fig.4 the sensor again has two slopes; front and back slope, which the sensitivity is higher in the front slope. Focusing on the front slope analysis, the sensitivities and linear range for molar, canine, hybrid composite resin and artificial teeth surfaces are obtained at 0.9667 mV/mm and 0.5 mm, 0.775 mV/mm and 0.45 mm, 0.5109 mV and 0.5 mm and 0.25 mV/mm and 0.5 mm respectively with a good linearity of more than 90 %. The peak voltages also decrease in the same order as the decrease in sensitivities which are 1.05, 0.685, 0.475 and 0.415 mV for the molar, canine, hybrid composite resin and artificial teeth surfaces, respectively. The highest resolution of approximately 0.0025 mm (front slope) is obtained with the molar tooth surface. The stability of the displacement sensor is also investigated and the measurement errors are observed to be less than 0.3, 0.88, 2.35 and 0.67 % for molar, canine, hybrid composite resin and artificial teeth surfaces, respectively.

Table 3.2. The features of the fiber optic displacement (front slope) for various flat teeth surfaces.

| No | Type of teeth and dimension | Linear range, mm | Peak voltage, mV | Sensitivity, mV/mm | Resolution, mm |
|---|---|---|---|---|---|
| 1 | Molar, 7mm x 7mm | 0.05 – 0.5 | 1.05 | 0.9667 mV/mm | 0.0025 |
| 2 | Canine, 6mm x 7mm | 0.05 – 0.45 | 0.685 | 0.775 mV/mm | 0.0067 |
| 3 | Hybrid composite resin, 9mm x 9mm | 0 – 0.5 | 0.475 | 0.5109 mV/mm | 0.0053 |
| 4 | Artificial, 8mm x 10mm | 0 – 0.5 | 0.415 | 0.25 mV/mm | 0.0084 |

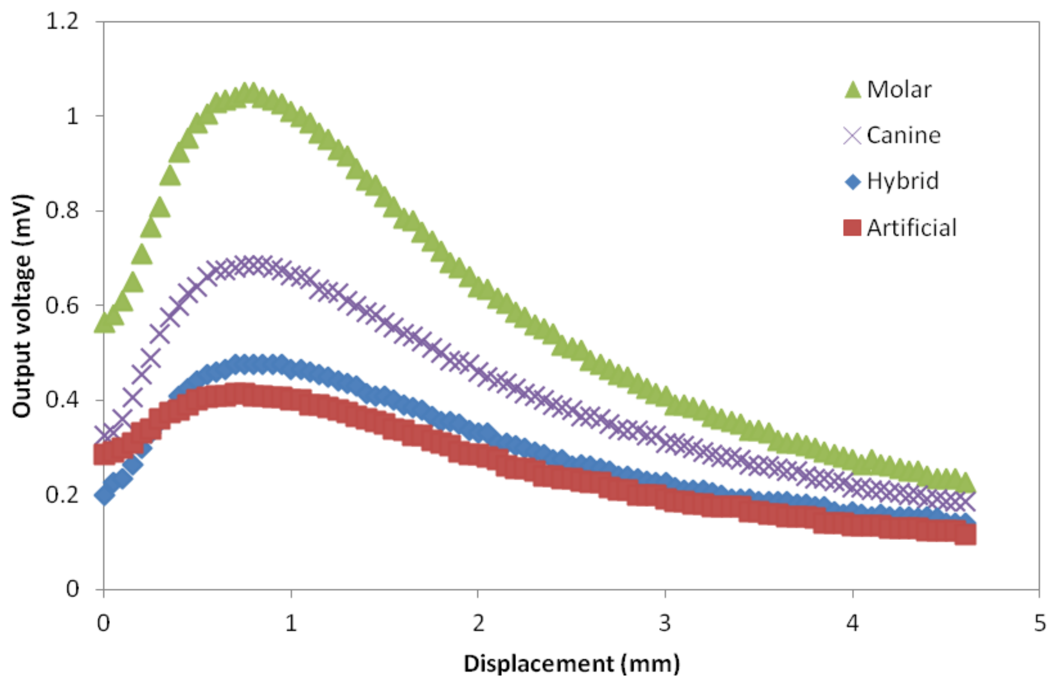



Fig 3.4: Variation of output voltage against displacement using He-Ne laser.

# CHAPTER 4

# FIBER OPTIC DISPLACEMENT SENSOR (FODS) FOR DETECTING TEETH SURFACE DEFECT THAT SIMULATE DENTAL CAVITIES

## 4.1 Introduction

In dentistry, a successful restorative procedure depends strongly on the precise mapping of the shape of dental cavities. Impressional methods are often used for that purpose (Jung et al., 2010; Rao et al., 2010) but the impression material might suffer from shape and size deformity during the course of mapping, copying and storage, leading to defects in the process. Visualization of the surface topography may help in the estimation of the quality of the mapping process. X-ray is one is one of the main method used for visualization in clinical practice (Nakata et al., 2006; Vandenberghe et al., 2010; Zabler et al., 2007). However, it involves relatively high doses of ionizing radiation which will have a negative impact on the healthy and safety of both dental patients and practitioners.

Optical methods are non-ionizing and hence have potential clinical applications (Jones, Huynh et al., 2003; Shimamura et al., 2011; Wilder-Smith et al., 2009) QLF and optical coherence tomography (OCT) are among the optical techniques receiving much attention and are discussed in detail by Huysmans et al (Huysmans et al., 2011). QLF uses specific wavelength of light to induce the fluorescence of the surface of the tooth. The surface that contains cavity appears to be darker due to the increased scattering of incident light which makes the cavity identifiable. However, QLF are rarely being used in private clinical practice due to its high unit cost and the complex operation involved (Rochlen et al., 2011). OCT on the other hand uses near infrared lights that are able to generate high resolution images of enamel samples but often hindered by the difficulty of performing in vivo studies and positioning of the probe (Schlueter et al., 2011).

The aim of this experiment is to develop a novel non-destructive imaging tools for an intensity modulated FODS visualization of teeth surface defect due to dental cavities and apply it for the determination of cavity shape and size. In our approach, various teeth surfaces, namely molar, canine, hybrid composite resin and artificial contains a machine drilled hole that simulate dental cavity. These types of sensors are inexpensive and simple to operate, hence opening the feasibility for enhancements in restorative dentistry to a wider international community.

## 4.2 Experimental Setup

Fig. 4.1 shows a schematic diagram for the axial displacement measurement using a multimode plastic fiber bundled with different type of tooth sample namely, molar, canine, hybrid composite resin and artificial. The FODS consists of a fiber optic transmitter, mechanical chopper, multimode plastic fiber optic bundled probe, a silicon photodetector, lock-in amplifier and a computer. The multimode plastic fiber is used due to the ease of alignment between the light and the fiber. The MMF probe is a commercial step index fiber coupler having a 980μm core made of PMMA, surrounded by a 10μm layer of fluorinated polymer acting as the cladding with numerical aperture is closed to 0.5. A red He-Ne laser (λ=633 nm) is used as the light source with an average output of 5.5 mW, beam diameter of 0.8 mm and beam divergence of 1.01 mRads. The photodetector is a high speed silicon photodiode with an optical response extending from 400 to 1100 nm, making it compatible with a wide range of visible light including the 633 nm visible red He-Ne lasers used in this experiment. The red He-Ne laser is modulated externally using a chopper at a frequency of 113 Hz as to avoid the harmonics from the line frequency which is about 50 to 60 Hz. The modulated light source is used in conjunction with a lock-in amplifier to reduce the dc drift and interference of ambient stray light.

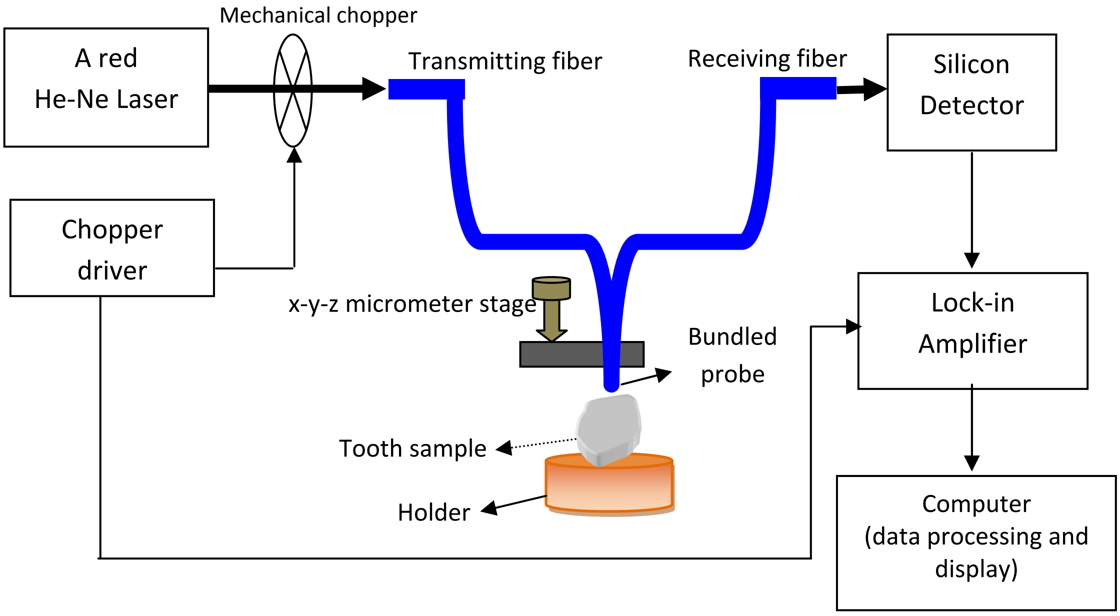



Fig. 4.1: Set-up for the imaging and detection of enamel tooth surface defect using fiber optic displacement sensor (FODS)

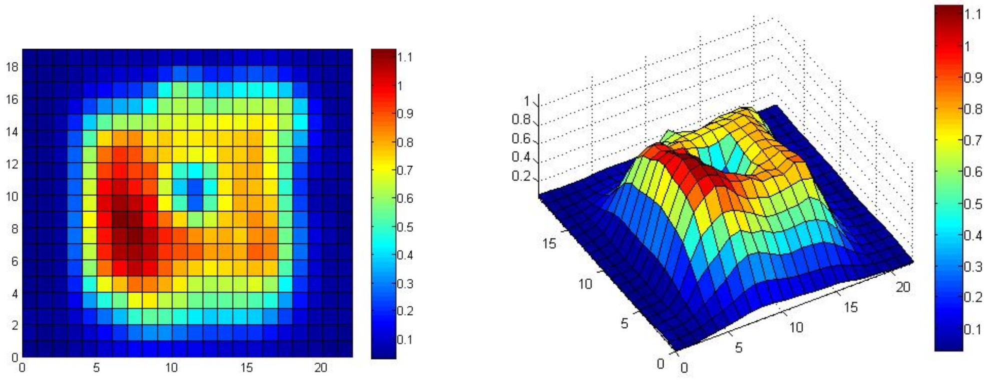
The displacement from the tooth sample is achieved by mounting it on a micrometer translation stage, which is rigidly attached to a vibration free table due to keep the laser and optic fiber free from outside vibration. The signal from the receiving fiber is measured by moving the probe away from the zero point, where the flat reflective surface of teeth samples and the probe are in close contact. The output light is then sent into the silicon phtotodetector and the electrical signal is then sent into the lock-in amplifier together with the reference signal of the mechanical chopper. The output results from the lock-in amplifier are then connected to a computer through a RS232C port interface and the signals are processed using Delphi Software.

In this experiment, the displacement of the bundled probe from the reflecting surfaces is done in steps of 0.05 mm. The probe is consequently fixed (z axis) within the linear range of displacement curve (as from chapter 3 results) and the intensity of the collected light as a function of lateral movement (x and y axis) of the tooth surface is recorded while being maintained in perpendicular and constant in axial position (z axis). Flaws are introduced in each of teeth samples by drilling holes that simulate dental cavities which will led to enamel surface defect. In order to use this system diameter quantification of the enamel tooth surface defect, we exploit the difference in the reflected light between the flawed and unflawed region. The experiment is carried out with minimum successive steps of 0.5mm for each of the tooth surface. Lastly, the raw data are processed and transformed into two-dimensional (2D) and three-dimensional (3D) images using Matlab and used for the detection of the tooth cavity. The experiment is repeated using all four different types of teeth surfaces. Due to the time constraint the experiment was conducted one time for each teeth sample.

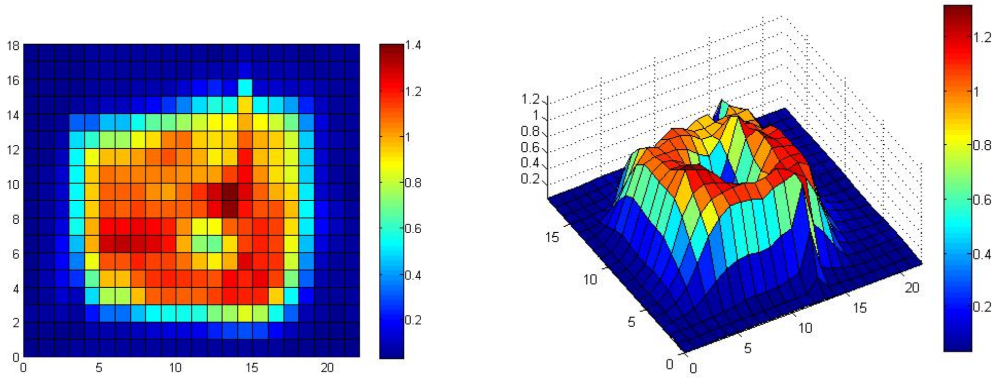
### 4.3 Results and Discussions

The performance of the FODS on sensitivity, linear range, peak voltage and stability was discussed in Chapter 3. Based on the linear range from Chapter 3, the recorded tooth surface was being maintained in perpendicular and constant in axial position (z axis). Further results and discussion in Chapter 4 will be focused on imaging of tooth surface prior to detection of enamel tooth surface defect.

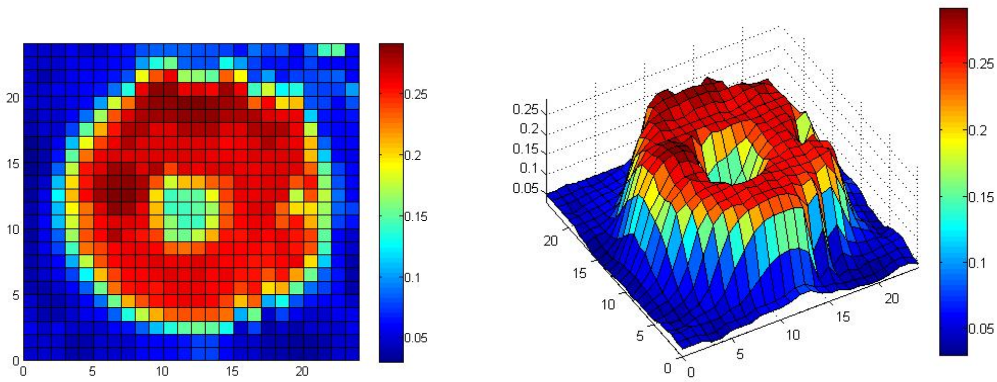
2D and 3D imaging of the various teeth surface profiles were required by scanning the teeth surface at various lateral positions at a fixed distance of 0.25mm between the fiber probe and teeth surface. The fixed distance was selected from the linear range discussed in Chapter 3. Fig. 4.2 (a) shows 2D and 3D views of the flawed molar surface, which was obtained by 27 x 27 lines of scanning along the row axis and column axis. The recorded signal from the flawed region is much more significantly reduced in amplitude than the surrounding region which occurs as results of a reduction in the reflected signal at that particular region. The intensity of the reflected light from the tooth surface depends upon the surface texture of the tooth and standoff distance between the surface and fiber tip. Thus, smaller reflected signal amplitude is expected. The experiment is then repeated for canine, hybrid composite resin and artificial surfaces with results as shown in Figs. 4.2(b), 2(c) and 2(d).



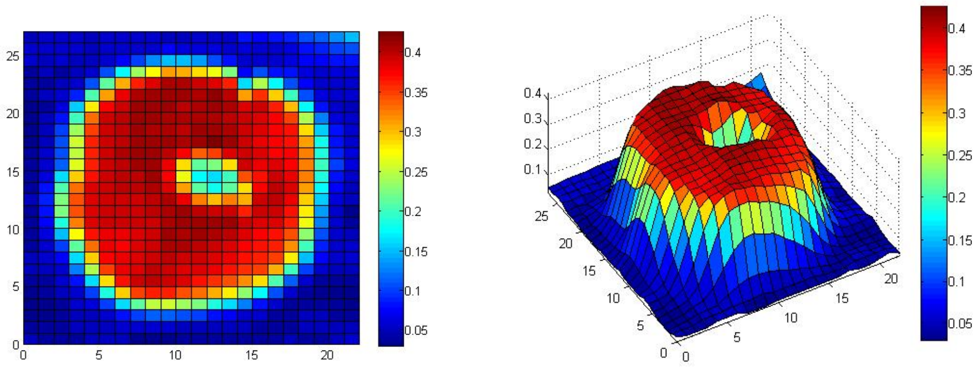
(a)



(b)



(c)



(d)

Fig. 4.2: 2D and 3D surface profile of the (a) molar, (b) canine, (c) hybrid composite resin and (d) artificial surface showing a clear difference in the reflected signal between the flawed and unflawed tooth region.

The figures clearly show the difference in reflected amplitudes between the flawed region and the surrounding region, again verifying the detection capabilities of the FODS on a enamel tooth surface. Fig. 4.3 shows the measured reflected signal taken along the same axis of a flawed and unflawed canine surface. The region of interest marked by the circle demonstrates the clear difference in the reflected signal between them. The diameter of the flaw is measured based on the total lateral displacement starting from the first occurrence of disparity in the reflected signals and ends with the coincidence of the signals. Taking into account the distance for each displacement which is 0.5 mm, the diameter is measured to be 2.5 mm, which is exactly the same value when measured with a micrometer. By using the same approach, the diameter of the surface defect in the molar, hybrid composite resin and artificial surfaces are obtained to be 2.0, 2.5 and 2.0 mm, respectively, as summarized in Table 4.1.

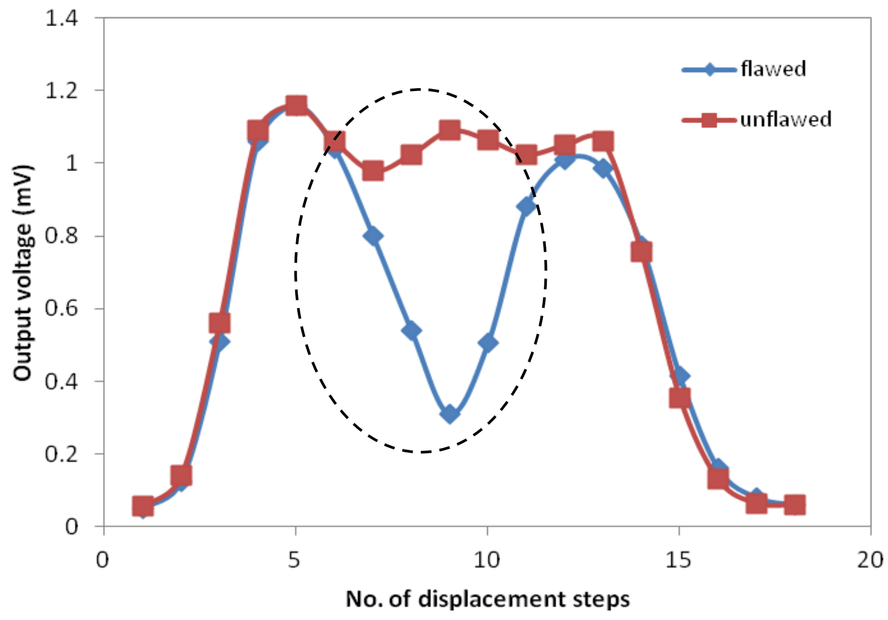



Fig 4.3: Variation of output voltage against position of flawed and unflawed canine surface.

Table 4.1. Diameter measurement of the flaw on the tooth surface using the fiber optic displacement sensor (FODS) and micrometer.

| **No** | **Type of tooth surface** | **Micrometer measurements (mm)** | **FODS measurements (mm)** |
|---|---|---|---|
| 1 | Molar | 2.0 | 2.0 |
| 2 | Canine | 2.5 | 2.5 |
| 3 | Hybrid composite resin | 2.5 | 2.5 |
| 4 | Artificial | 2.0 | 2.0 |

The results indicate the capability of implementation of the displacement sensor for the imaging and detection of enamel tooth surface defect. In order to cater for high resolution requirements, additional computer vision techniques and algorithms need to be implemented for the matching computations at each position of the output disparity. The threshold for then output disparity need to be clearly defined based on extensive analysis of a larger amount of data with higher sampling rate.

# CHAPTER 5

# CONCLUSION

A simple design of a FODS is demonstrated using teeth samples as the light reflective target. He-Ne laser is more suitable to be used as the light source compared to laser diode due to its less scattering behaviour on dental hard tissue. The difference in displacement response, sensitivity, and peak voltage on teeth samples are investigated. The sensitivity is decreased upon the decreasing of peak voltage of each voltage with good linearity of over 90%. The simplicity of the designed sensor will help future researchers to use the properties of displacement for further investigation of dental hard tissue properties.

A fiber optic sensor is introduced as a novel non-destructive method for quantitative imaging and measurement of tooth surface defect that simulate dental cavity in a non-contact mode. The sensor is based on intensity modulation technique and uses a multimode plastic bundled fiber as a probe and He-Ne laser as the light source. The feasibility of the sensor to identify teeth surface defect on different types of teeth has been demonstrated. While, the results are still some way from clinical applications, they demonstrate the possibility to detect tooth cavity in a safe and non-invasive manner compared to conventional method such as x-rays. Furthermore, they are particularly beneficial in the estimation of the quality of the impressional mapping methods currently used in restorative dentistry.

In future, the real surface defect teeth due to dental cavity should be used as the sample to investigate on the proposed sensor. In addition, the proposed sensor may also be used for teeth surface imaging of the teeth that is useful in the aesthetics fields.

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