PLASTIC OPTICAL FIBRE BASED SENSORS FOR MONITORING PHYSIOLOGICAL PARAMETERS IN CLINICAL ENVIRONMENTS

Submitted by Wern Kam
For the award of Doctor of Philosophy

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Submitted to the University of Limerick
November 2018
ABSTRACT

This dissertation focuses on a plastic optical fibre (POF) sensor based on intensity modulation for health monitoring applications in a modern clinical environment. The sensor is suitable for spine bending monitoring as well as respiratory assessment.

The sensor of this investigation has been designed and fabricated in all-plastic form using 3-D printing and plastic optical fibre to form a sensor which operates based on intensity modulation. For spine bending assessment, bending modes for lateral and sagittal directions are monitored through the change of light intensity coupled to three separate output fibres that are aligned to the input fibre. The sensor exhibits a working range of ±12° and has a good agreement when the bending response is evaluated with theoretical calculation. A good correlation with established image acquisition and Biometrics goniometer based techniques has been demonstrated in monitoring the spine bending movement. The same sensor has been utilized to monitor the respiration rate by detecting the upper body movement due to thoracic respiratory motion changes. The sensor has shown excellent feasibility and can be used on any of four different positions on the body to measure the respiratory signal. Comparison of the sensor output versus a commercial instrument (the Neulog NUL-236 respiratory sensor) demonstrates that the POF sensor is capable of accurately measuring the human breathing cycle and frequency. The all-plastic optical fibre sensor is compact, portable and capable of operating autonomously within a Physiology clinic and other specialized environments such as X-ray and Magnetic Resonance Imaging (MRI).
DECLARATION

UNIVERSITI OF LIMERICK

ORIGINAL LITERARY WORK DECLARATION

Name of Candidate: WERN KAM
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Name of Degree: Doctor of Philosophy
Title of Dissertation/Thesis:
Plastic optical fibre based sensors for monitoring physiological parameters in clinical environments

Field of Study: ECE Engineering

I do solemnly and sincerely declare that the work described in this thesis is, except where otherwise stated, entirely that of the author and has not been submitted in any part for a degree at this or any other university.

Candidate’s Signature:   Date:

Wern Kam
ACKNOWLEDGEMENTS

I would like to express my deep appreciation and sincere gratitude towards my family members, my brother Kam Shern, my parents Chong Chee and Ai Gek for all the endless support, understanding and encouragement all along the research.

Many thanks to my supervisors, Prof. Elfed Lewis, Dr. Waleed S. Mohammed, Dr. Sinead O’Keeffe and Dr. Kieran O’Sullivan for their patient guidance throughout the project. They act as a role model for me and set good examples for me to learn from the beginning to the end of my research.

Gratitude also goes to Optical Fibre Sensor Research Center, University of Limerick for providing necessary instruments, apparatus and facilities for the research. Special thanks to the staff of University of Limerick, especially Jimmy Kelly, James Keanne, Jimmy Sullivan, John Bird, Bob and Donal which rendered assistance and support to make this research possible. My research activities in the University of Limerick were supported by Erasmus Mundus under LEADERS programme, which I really appreciated.

At the same time, I feel very thankful to all the research team mates, especially Yong Sheng and Ling-Xia who provided me useful information, support and opinion when I am facing problems. I cherish the moments we work and face the difficult period together. Finally, thanks to the volunteers who participated in the field test measurement of the sensor.
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<tr>
<td>2D</td>
<td>Two-dimensional</td>
<td></td>
</tr>
<tr>
<td>3D</td>
<td>Three-dimensional</td>
<td></td>
</tr>
<tr>
<td>ADC</td>
<td>Analog to Digital Converter</td>
<td></td>
</tr>
<tr>
<td>AI</td>
<td>Analog Input</td>
<td></td>
</tr>
<tr>
<td>AI-PCFI</td>
<td>Agarose Infiltrated Photonics Crystal Fibre Interferometer</td>
<td></td>
</tr>
<tr>
<td>AO</td>
<td>Analog Output</td>
<td></td>
</tr>
<tr>
<td>BMI</td>
<td>Body Mass Index</td>
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</tr>
<tr>
<td>CAD</td>
<td>Computer-aided Design</td>
<td></td>
</tr>
<tr>
<td>CCD</td>
<td>Charge-coupled Device</td>
<td></td>
</tr>
<tr>
<td>CO₂</td>
<td>Carbon Dioxide</td>
<td></td>
</tr>
<tr>
<td>COPD</td>
<td>Chronic Obstructive Pulmonary Disease</td>
<td></td>
</tr>
<tr>
<td>CT</td>
<td>Computed tomography</td>
<td></td>
</tr>
<tr>
<td>DAQ</td>
<td>Data Acquisition</td>
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<tr>
<td>DI</td>
<td>Digital Input</td>
<td></td>
</tr>
<tr>
<td>DO</td>
<td>Digital Output</td>
<td></td>
</tr>
<tr>
<td>EM</td>
<td>Electromagnetic</td>
<td></td>
</tr>
<tr>
<td>EMI</td>
<td>Electromagnetic Interference</td>
<td></td>
</tr>
<tr>
<td>λ</td>
<td>Wavelength</td>
<td></td>
</tr>
<tr>
<td>LBP</td>
<td>Lower Back Pain</td>
<td></td>
</tr>
<tr>
<td>LED</td>
<td>Light Emitting Diode</td>
<td></td>
</tr>
<tr>
<td>LPF</td>
<td>Low Pass Filter</td>
<td></td>
</tr>
<tr>
<td>LPG</td>
<td>Long Period Grating</td>
<td></td>
</tr>
<tr>
<td>MMF</td>
<td>Multimode Fibre</td>
<td></td>
</tr>
<tr>
<td>MRI</td>
<td>Magnetic Resonance Imaging</td>
<td></td>
</tr>
<tr>
<td>Abbreviation</td>
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</tr>
<tr>
<td>--------------</td>
<td>----------------------------------</td>
<td></td>
</tr>
<tr>
<td>$n_c$</td>
<td>Refractive Index of Core</td>
<td></td>
</tr>
<tr>
<td>NA</td>
<td>Numerical Aperture</td>
<td></td>
</tr>
<tr>
<td>O$_2$</td>
<td>Oxygen</td>
<td></td>
</tr>
<tr>
<td>OFS</td>
<td>Optical Fibre Sensor</td>
<td></td>
</tr>
<tr>
<td>Op-amp</td>
<td>Operational Amplifier</td>
<td></td>
</tr>
<tr>
<td>OSA</td>
<td>Optical Spectrum Analyser</td>
<td></td>
</tr>
<tr>
<td>PCB</td>
<td>Printed Circuit Board</td>
<td></td>
</tr>
<tr>
<td>PMMA</td>
<td>Poly(methyl methacrylate)</td>
<td></td>
</tr>
<tr>
<td>POF</td>
<td>Plastic Optical Fibre</td>
<td></td>
</tr>
<tr>
<td>RF</td>
<td>Radio Frequency</td>
<td></td>
</tr>
<tr>
<td>ROI</td>
<td>Region of Interest</td>
<td></td>
</tr>
<tr>
<td>ROM</td>
<td>Range of Motion</td>
<td></td>
</tr>
<tr>
<td>SMF</td>
<td>Single Mode Fibre</td>
<td></td>
</tr>
<tr>
<td>SNR</td>
<td>Signal to Noise Ratio</td>
<td></td>
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<tr>
<td>USB</td>
<td>Universal Serial Bus</td>
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INTRODUCTION

1.1 Human Spine System

Recently, development of novel sensor systems for health monitoring in clinical environment has gained increasing attention. Types of health monitoring applications include tracking of spine bending [1], limb motion tracking [2], blood pressure monitoring [3], respiration rate assessment [4] and body temperature measurement [5]. These sensors have evolved from predominantly manual measurement e.g. pressure cuff and manometer [6, 7] into autonomous monitoring using different sensing approaches e.g. fully automated blood pressure recording system [8] which is present in the majority of modern clinics.

There is an increasing need for a reliable and accurate human spine monitoring sensor to measure the movement or bending of the human spine. The human spine, or vertebral column plays an important role in skeletal support and protects the spinal cord. The function of vertebral column includes support and allows movement of head and trunk, allows nerves to exit the spinal cord and provides attachment for muscles via tendons. The system consists of five sections: cervical, thoracic, lumbar, sacrum and coccyx as illustrated in Figure 1.1. Each region consists of vertical segments which are: 7 vertebrae in the cervical region, 12 vertebrae in the thoracic region, 5 vertebrae in the lumbar spine followed by 5 fused sacral vertebrae and 4 fused coccygeal vertebrae [9].

As shown in Figure 1.1, a normal spine shape resembles a soft natural ‘S’ curvature when viewing from the side (lateral) with a slight lumbar lordosis (anterior curvature) and thoracic kyphosis (posterior curvature). This curve supports the spine in absorbing stress from movement and gravity. The spine forms a vertically straight structure when viewing from the back (posterior).
The lumbar spine section, between thorax and sacrum consists of five mobile vertebrae that graduate in size from L1 to L5. They are the largest vertebrae in the spinal column and are more robust compared to other spinal bones in order to support the weight of entire torso. Of all the spinal regions, the lumbar spine is the most intensively investigated due to the occurrence of low back pain (LBP) in adults [10]. LBP is the fifth most common reason for a physician visit [11]. It also has had a great impact in the modern industrialized world. LBP is one of the most common health problems and the second leading cause of work absenteeism in United States [11, 12]. The cost of medical care for treating back pain, compensation payments to workers and time lost due to back pain problems are estimated to cost as much as $50 billion a year [13]. These statistics emphasize the need for an accurate and low cost lumbar spine monitoring sensor to ensure that the major cause of LBP is identified.
In this work described in this dissertation, a 3-D printed sensor housing accommodating the optical fibre probes has been fabricated with the aim to be used in Physiotherapy clinical environment for LBP diagnosis and identification of the most appropriate treatment. The Plastic Optical Fibre (POF) sensor is used to measure the bending angle, specifically in the lumbar spine region. The sensor has been developed in order to be able to autonomously monitor bending of spine in both lateral and sagittal directions. The low cost, non-invasive, reliable coupled with excellent portability of the system developed in this work will aid the physician for LBP diagnosis within a Physiology clinic and other specialized environments and potentially minimize use of expensive imaging modalities such as Computer Tomography (CT) or Magnetic Resonance (MRI) scanning.

1.2 Human Respiration System

Vital signs are the basic measurement that indicate the status of human body. The four basic vital signs that are monitored by the health care are body temperature, pulse rate, respiration rate and blood pressure [14]. Human respiration is therefore one of the main vital signs for assessment of health condition. Respiration provides three main functions to the body namely: supply of sufficient Oxygen (O₂) to all cells in the body, maintaining the pH of the blood plasma and plays a vital role in regulating the body temperature [15]. Figure 1.2 shows a sketch of the physiology of the human respiratory system. The respiration cycle can be considered to commence when air enters into the mouth or nose which wets and warms the air. The air then travels through pharynx (back of the throat) to the trachea. The trachea is shaped like a tube of approximately five inches in length and this respiratory tract splits into two branches called as bronchi to enter the lungs. The primary bronchi branches into smaller bronchi. The smaller section of bronchi
splits into bronchioles that spread throughout the lungs. The final division ends with the alveoli where the gas exchange takes place at these miniature round air sacs [16].

During the inhalation process, the pressure inside the lungs decrease. The muscle of the diaphragm contracts and the ribcage is forced upwards and outwards to increase the space (volume) in the chest cavity. Air with oxygen that flows into the lung and enters the alveoli and from there passes into the surrounding blood vessels. At the same time Carbon Dioxide (CO$_2$) moves from the blood capillaries into the air sacs. During the exhalation, the diaphragm in the dome shape becomes relaxed and move upwards. Conversely, ribcage moves downward and inward. The chest cavity thus becomes smaller in volume and air with CO$_2$ is forced out of the nose or mouth [17].

![Diagram of the human respiratory system](attachment:respiratory_system.png)

**Figure 1.2**: Human respiratory system.

The respiratory signal, particularly the change in respiration rate can be used to predict potential serious illness such as cardiopulmonary arrest [18] and trauma leading potentially to death [19]. Studies have shown that the respiratory rate is a better vital parameter to discriminate between a patient in a stable condition and patient at risk compared to using blood pressure and pulse rate [20]. A study by Cretikos et al. [21] found that more than half of the patients that suffer cardiac arrest, unplanned ICU admission or unexpected deaths had a respiratory rate of more than 24 breaths/min. These
could be identified up to 24 hours prior to the serious event. Furthermore, Lindberg et al. [22] stated the importance of respiration information for example in the post-operative care unit where patient that has just undergone an operation has the risk being affected by analgesics that depress respiration. In neonatal intensive care units, it is important to monitor respiration in infants as breathing disorder is a common problem found in premature babies. All of these clinical reasons show that a reliable measure of the respiration cycle is necessary to provide indications related to general health or the physiological state of patient. Utilizing the respiratory rate as a signal in clinical diagnosis is therefore useful in lowering the incidence of related illness and thus reduces associated morbidity and mortality.

The second motivation of this work is to support the continuous, portable and low cost monitoring of respiration signal in detecting early signs of illness. The advantage of the fabricated POF sensor for spine monitoring has been explored for its use in respiratory monitoring purposes. The respiration signal is measured through monitoring the movement of upper body (chest wall and abdomen region) during the breathing process. The POF sensor is attached on the body to detect the expansion and contraction of the lungs as described above, and thus respiration rate can be derived with relative ease from the sensor signal. The sensor has the advantages of being small sized, light-weight and able to be mounted externally on different positions of the upper body (including the back) compared to the conventional respiratory sensors which are generally provided in a stretchable belt form. The portability and all plastic fabrication of the POF sensor also provides an alternative option for it to be used in the clinical environment or for home based monitoring as it can be made very robust. The respiration sensor is considered in the next section below.
1.3 **Objectives of Research**

The aim of this research is to develop functional robust and cost-efficient sensors based on a novel combination of 3D printing and optical fibre based technology. The sensor modulation mechanism is based on intensity modulation and is able to monitor spinal bending or general body movements including in both sagittal and frontal plane. The sensor is fabricated using 3-D printing and plastic optical fibre which opens the possibility for it to be operated autonomously (at a distance of a few meters at least) within a Physiology clinic and specialized environments for medical diagnosis such as X-ray or Magnetic Resonance Imaging (MRI).

The overarching objective is achieved through implementation of the following sub-objectives:

1. To develop a ‘hand-held’ user friendly system for lower back pain (LBP) diagnosis based on a novel optical fibre sensor which is capable of accurately measuring back movements of patients in both sagittal and lateral planes.

2. To design and fabricate a cost effective, small size, light-weight and easy to operate sensor probe that is able to adapt to different body size and is comfortable to wear.

3. To be able to use this sensor to provide accurate information to clinicians to evaluate the effectiveness of on-going physiotherapy and assessment of the spine condition.

4. To utilise the same technology to develop a respiration monitoring sensor that is capable for use in MRI and other high energy scanning environments.
1.4 Thesis Overview

This thesis comprises a comprehensive study resulting in the design and implementation of a novel and robust optical fibre sensor device for spine monitoring and respiratory assessment based on intensity modulation. The device is investigated and demonstrated on human subjects in order to optimize the performance of the sensor. The first chapter gives a brief introduction of human spine system and human respiratory system. The motivation behind this work as well as the research objectives are identified in this chapter.

Chapter two covers the background and details of the research topic, which includes the types of spine assessment, classification of spine monitoring sensors, as well as respiration monitoring sensors. An extensive and up to date review of the literature describing optical fibre sensors from previous approaches for spine and respiration monitoring sensor is included in this chapter, where appropriate the advantages of proposed skin-mounted sensor configuration of this investigation are highlighted.

The third chapter describes the methodology of the sensor which includes a detailed explanation of the underlying concept and configuration of the sensor using four POF fibres. The theory of the sensor’s operation is also elaborated in this chapter. The development of the sensor from the first experimental prototype fabricated using readily available material to the final development of 3-D printed and packaged sensor is described. The chapter also includes the design specifications of 3-D printed sensor, the signal conditioning circuit, software and hardware components used in the sensor configuration.

Chapter four includes details of the finalized CAD (computer aided design) of the sensor as a 3-D printed entity. Steps of sensor probe fabrication including the housing of the POF and connection between both light transmission tubes via a silicon tube and mold
is discussed. Details of the sensor probe and transmitting fibre assembly coupled with the PCB (LED driver and photodetector amplifier circuits) and detection hardware device are included. This chapter also outlines the methods for the sensor’s calibration using an in-house designed optical setup, laboratory measurement and preparation for field testing of the sensor on human subjects for both lower back bending and respiratory assessment.

Chapter five focuses on the details of the specific laboratory sensor calibration including the sensor characteristic such as bending response, operating range, sensitivity, accuracy and resolution. The results of theoretical investigation and results of preliminary spine bending measurement on human subjects are also described. For the preliminary testing, an image acquisition method is used as an alternative means for assessing lower spine bending and to provide the opportunity to compare the results with bending signal obtained from the POF sensor. The sensor is further evaluated on a larger group of test subjects for which the POF sensor performance is compared with a conventional Biometrics Ltd goniometer. Analysis of the sensor evaluation using a Statistical Package for the Social Sciences (SPSS) software is included.

The sixth chapter describes an investigation including the results of the POF sensor testing on respiratory assessment. The sensor is initially tested on different positions of the same subject to emphasize the capability of the sensor to operate at various detection points on the human body. Three identical sensor probe and systems are fabricated to test the breathing signal simultaneously in three different body positions (front and back). In the second part of the assessment, the POF sensor is simultaneously compared with a conventional respiratory monitoring device - NUL-236 Neulog belt sensor. Results of both sensors on normal breathing and deep breathing in sitting and lying positions are presented.
The last chapter summarizes the significant achievements and draws conclusions of using the POF sensor for both spine and respiratory monitoring. Potential future plans and modification that can be carried out to improve the performance are also included in this chapter.
CHAPTER 2 LITERATURE REVIEW

2.1 Introduction

This chapter includes a review of the current state of the art for types of spine assessment, spine monitoring and respiratory monitoring sensors. For spine assessment, instrumentation for static spine posture and dynamic spine measurement is reviewed. These health monitoring devices are commonly categorized into two sections, imaging devices and contact type devices. Contact type spine monitoring devices are further classified into the hand-held sensors and skin-mounted sensors. The contact type respiratory monitoring sensor is categorized into two sub-groups based on the method of sensing, which is the airflow sensing and movement of chest or abdominal deformation detection. Currently available commercial devices, research based devices and general methods involved in monitoring human spine condition and respiration is explored. The last section focuses on the currently available optical fibre bending sensors for health monitoring and respiratory monitoring.

2.2 Types of Spine Assessment

Spine postural assessment is an important step in the identification of the cause of back pain that is associated with spine ailment. Clinically, it involves a detailed examination of the human back such as detection of unusual spine curvature angle, range of spine motion, rotation and flexion of the spine to locate the source of pain. Spine assessment in a clinical setting can be categorized into two types, namely static spine posture and dynamic spine movement monitoring. Static spine assessment acquires the physical condition of the spine (i.e. the spine curvature angle and/or spine structural abnormalities) during different set stationary postures such as standing or sitting. Each
natural curvature of the spine is intended to evenly distribute the mechanical stress incurred from the rest of the body even when stationary.

Spinal curves that are significantly misaligned from natural static curves can cause discomfort and even disability [23]. Common examples of spinal deformities include Lordosis, Kyphosis and Scoliosis as illustrated in Figure 2.1. In this situation, the curvature in certain sections of the spine is misaligned or exaggerated. Lordosis, or swayback is the excessive anterior (forward) curvature at the lumbar spine. Kyphosis, or often named as hunch back is an excessive posterior (backward) curvature at the thoracic spine region. Scoliosis is the abnormal curving of the spine to the lateral (left or right side) occurring as ‘C-shaped’ or ‘S-shaped’ [24, 25]. These spinal deformities affect the functional aspects of patients, resulting in excess pressure on the spine, occasionally causing pain and may require surgery in the cases of dangerous curvature. It is important from a clinical point of view to pinpoint the static spine parameters to identify specific spinal disorder. The most common measurement of static spine parameter is conducted using radiography such as X-rays and MRI scanning. It is the gold standard for assessment of spinal disorder and allows accurate calculation of the Cobb’s angle from the perpendicular lines drawn on radiography [26]. Other alternative spine monitoring approaches are discussed in the next section.
Dynamic spine assessment is the examination of the spine condition when patient is subjected to continuous spine movement in flexion, extension, lateral bending and axial rotation. From the dynamic spine assessment, flexibility in bending and specific movement that causes pain during daily activities can be identified. The characteristics of dynamic spine parameters such as range of motion (ROM), spine bending angle in different planes or angular velocity are obtained to evaluate the improvement of motion following surgical treatment. Functional aspects of patients are more interconnected with dynamic spine status compared to static posture [27]. The level of LBP injury can also be classified using this approach as in most cases, LBP is not always physically apparent and the patient suffers the pain only when undergoing specific motions. Therefore, it is important to study the spine parameters from a dynamic physical examination to identify a specific movement or group of movements that causes discomfort or pain. In the past, physicians study the spine condition using only ‘naked eyes’ observation from different angular perspectives as the patient undertakes motion in the sagittal and lateral planes as illustrated in Figure 2.2. This physical examination requires specific expertise to conduct the test and there is no explicit quantitative scale to determine the output results and
therefore diagnosis can be somewhat subjective. A device to measure the characteristic of spine is often used by the physicians to aid them in diagnosing the spine condition. A comprehensive review of dynamic spine monitoring devices is presented in the next section.

Figure 2.2: Human motion planes and axis.

2.3 Spine Monitoring Sensors

Lower back pain (LBP) is a major symptom or health problem associated with poor spine condition. Lower back pain is currently the primary cause of disability worldwide [28]. There are many reasons that lead to lower back pain. LBP can be due to individual factors such as age and excess weight or psychological factors including anxiety, depression and mental stress [29]. Occupational related mechanical exposures such as lifting, pulling, pushing heavy weights and kneeling or squatting for a long time
[30] are other frequently reported risk factors giving rise to LBP. This causes a torn muscle/ ligament or tissue injuries that damage the discs, nerves roots and joints of spinal vertebrae. Other external factors such as poor posture and inadequate back support especially workers who sit in an office or standing for long hours can also place excess stress on the spine. Eventually, the bad posture will change the anatomical characteristic of the spine causing back pain [31].

To accurately identify the source of low back pain, a physical examination is necessary. In the clinic, accurate diagnosis or treatment of patients with LBP problems requires implementation of accurate spine monitoring tools. Selection of the appropriate sensor for physical examination or treatment is important to aid the physician in screening and identifying causes of spine impairment. There are various types of spine monitoring sensors for the assessment of static spine bending and dynamic spine bending. These sensors can be categorized into two main groups, namely sensors using imaging techniques and non-imaging techniques or the contact type sensors. For spine imaging sensors, the device achieves screening of the spine image without any skin-contact with the patient. This group of sensors include the conventional radiography techniques used in the clinic to screen the spine conditions e.g. X-Rays. In the case of the contact type sensor, the device makes direct contact with the skin surface of the patient in order to measure the spine conditions.

2.3.1 Spine Imaging Sensors

Spine monitoring using imaging devices is the most common option for low back pain patients in clinic to diagnose spine conditions. The development of spine imaging technology allows the non-invasive assessment of spinal cord structure and condition in detail. An example of an imaging technique that is widely used by physicians in routine clinical practice to monitor spine condition include radiography (X-ray Imaging) [32],
magnetic resonance imaging (MRI), computed tomography (CT) scanning, myelography [33] and scintigraphy [34].

X-ray is a wavelike form of electromagnetic energy that has higher energy than visible light and lower energy than gamma rays [35]. Modern clinical X-ray imaging machines are widely used to diagnose various bone ailments, e.g. fractures. Despite the fact that X-ray imaging is non-invasive and painless, there are some potential limitations compared with other radiography technique. First, it is impossible to display the structure from a single X-Ray image in 3-D dimension. Hence, the details from the X-ray’s image may lead to confusion when object is situated at depth. Secondly, it can be hard to distinguish between the soft tissues such as human organs and it can also be difficult to quantify the separate densities of individual substances as X-rays record the mean absorption by all the various tissues through which they pass [36].

In order to overcome the limitations of conventional X-ray technology, computed tomography (CT) imaging has been introduced. The CT scan was first developed by Godfrey Hounsfield [37] and began to appear in clinics from around 1974. The scanning was first implemented in clinical setting to examine a patient’s head [38] and was subsequently extended to the larger size later for whole body imaging. The CT scanning technique obtains cross-sectional images of body by irradiating X-ray beams from different angles. Multiple image slices are acquired simultaneously, reformatted and reconstructed to obtain high resolution of 3-D images. Therefore, it has the advantages of obtaining a more accurate display of the human anatomy from the point of view of possible formation of a 3-D radiography image. Both CT and conventional X-Ray imaging utilize low to mid energy (10s of KeV) X-Rays. X-Rays are classified as ionizing radiation and hence pose a small but not insignificant risk to the patient. Their repeated use is therefore not widely acceptable in the medical field, but are used only when considered absolutely necessary e.g. for accurate diagnosis of fractures.
Magnetic resonance imaging (MRI) represents another imaging suited for examining spinal cord injuries other than CT scanning and X-ray radiography. MRI is a medical imaging technique that uses the spinning protons (hydrogen ions) in the human body to produce image. It requires a strong magnetic field (1.5 Tesla or more) to align the hydrogen atoms either parallel or antiparallel to the field [34]. A radio frequency (RF) pulse is applied to change the spin direction of hydrogen nuclei which resonate. When the radiofrequency pulse is off, MRI detects the energy that release from hydrogen nuclei to its resting states. Based on the received radio wave, the MR image is created to investigate the tissues and anatomy [39]. The main feature of MRI compared to CT scan and X-ray is there are no known biological hazards such as radiation and thus does not damage the tissue during scanning. The limitation is the cost for MRI equipment and scanning is very expensive [40]. As the MRI machine uses magnetic field to operate, patients with ferrous metal implants or pacemakers can be dangerous for use in the scanners and thus not suitable for MRI scanning.

The spine imaging modalities discussed above are frequently used in the clinic as a gold standard in spine assessment. There also exist other available spine imaging sensors, for example the use of rasterstereography technique [41]. The system projects white light raster line (grid line) on the surface of the back and capture the contour lines overlaid on a photograph of the back using a camera oriented at a certain angle. The coordinates of the raster points are calculated using triangulation algorithms to construct the 3-D image of scanned object. The use of rasterstereography in back spine analysis has the advantages of having a relatively fast scanning time, being reliable and ionizing radiation-free [42]. It enables the patient to be subject to multiple scans in measuring important parameters of the spine before and after treatment. However, the normal rasterstereography scanner is suitable only for static spine posture assessment.
Another example of imaging sensor is the commercially available Vicon motion analysis system [43] for tracking and analyzing spine motion. A number of passive spherical reflective markers are mounted along the spine. Each marker is placed on a specific anatomical landmark depending on the required measurement parameters. Spine motion is then recorded by a minimum of three high speed infrared Vicon cameras and calculated using proprietary software. The sensor is able to monitor spine movement based on the markers’ trajectory. Such a system requires a room space with minimum light reflection to install the camera and requires careful placement of the markers at the spine to avoid error from skin movement artefacts.

In conclusion, the use of imaging sensors for spine assessment represent a popular options due to their accuracy and reliability. However, it is not possible to use these techniques regularly for assessment or in physiotherapy practice due to their high implementation cost [44] and they can be time consuming. The equipment associated with these methods are generally large, and often immovable and therefore require a special room to function effectively. Most importantly, there exist risks from radiation damage when using certain imaging modalities (i.e. CT scan, X-ray) [45]. Additionally, most radiography based techniques have the limitation that they can only monitor static posture parameters in patients with spine problems [27]. Subsequently, the use of a non-invasive contact type sensor provides a valuable alternative to assist physicians in spine related defect diagnosis and therapy.

2.3.2 Contact Type Sensors

Contact type devices for spine monitoring assessment include the hand-held and wearable or skin-mounted types. Hand-held type sensors require a trained examiner or physician to operate the instrument. The devices are used by manually placing the sensor on a specific spine region when patient performs certain static postures. The sensor is
used with operators maneuvering along different spine segments of the patient to record the data. Skin-mounted sensors are used by attaching the sensor probe on the back of the patient and the data is immediately recorded. Examples of different groups of these type of sensor are presented in the following section.

2.3.2.1 The Hand-held sensor

In early investigations, the device most widely used for spine assessment was a hand-held type sensor where the human operator was required to manually measure the spine position. Such devices were usually used for measuring spine angle when the subject was in a stationary posture. The first example of this type of device comprised a simple tape measure [46]. The tape measure in this case comprised flexible strips of tape with regular intervals marking on them. It was used to measure the amount of change in length of spine segments at different body postures. The skin was marked along the spine and the distance between markers measured using the tape measure following trunk bending. The technique is simple to achieve but cannot measure the exact angle of lumbar curvature as it only measures the amount of spine displacement in centimeters. Thus, only limited spinal parameters could be accessed using this method of measurement. A similar kind of hand-held device for spine curvature measurement was available in the flexi-curve ruler [47], which resembles the flexi-curves used in secondary education to hand-draw graphical curves before the advent of computer graphical drawing software. It is a plastic moldable device that could be bent in one plane. The ruler is able to retain the shape after alignment and molded to the curvature of spine. The spinous processes is first identified from the palpation method [48], which is a common examination technique performed by the physician to identify the anatomy of spine or particular location of bone segments/joints. The specific spinous processes is marked using removable stickers. The flexible ruler is placed on the bare back at the range of the markers and is pressed firmly
to the spine. The curvature of the ruler that presents the curvature of measured spine was then traced onto paper for further analysis. The flexi-curve ruler shown to be a reliable option in spinal posture measurement due to its simplicity to use, non-invasiveness and low cost. Nonetheless, the sensor is prone to error when tracing the structure of spine as the shape of ruler can be accidentally altered when detached from the spine [49].

The manual goniometer is a widely used device in clinical spinal assessment. There are several types of goniometer available for spine measurement such as the gravity goniometer, parallelogram goniometer and standard goniometer [50]. It is often used to measure the range of motion and angle at a joint or between two spine landmarks. The general goniometer has two base platforms that need to be placed on two particular points of the spine. The angle formed between two goniometer arms are directly read from the goniometer. The goniometer is preferably chosen over the tape measure as it can directly study the angular change between the curvatures at different posture. However, the reliability of goniometer relies on even pressure of goniometer’s base being applied to the patient’s skin. The manual goniometer was modified to an electro goniometer later to allow continuous measurement of lumbar spine position. Other examples of hand-held device include the inclinometer [51]. The inclinometer is an instrument that measures the slope or tilt angle of the body with respect to the normal (perpendicular to a flat floor) caused by gravity. It is used to measure the spine curvature or angle of motion by placing the feet of the inclinometers over the specific spinous processes [52]. Usually two inclinometers are used with one being placed at the top of the measured spine segments while the other inclinometer is located at the lower spine segment. Readings of angle of inclination are thus read directly from the inclinometers with the value from lower spine being subtracted from the value from the upper spine.
The sensors described above require the examiner to manually read the data from the instrument. This could lead to human error when taking the reading and recording data. A more accurate option for the hand-held spine monitoring sensor became available that combined with a computer program (PC) and was given the name Spinal Mouse\textsuperscript{TM} [53]. The sensor has two rubber rollers on a mobile support and is used by guiding manually along the spine. The surface line of spine that reflects the contour in sagittal and frontal planes are automatically recorded by three sensors (three plane axes). Data is transmitted through a Bluetooth connection to the computer program to calculate the clinical parameters. The Spinal Mouse has exhibited good reliability in measuring the standing sagittal curvatures of vertebral, ranges of motion [54] and the spinal curvature in the frontal plane [55]. However, currently this sensor is only limited to use in static spine measurement.

Another computerized hand-held sensor is the ZEBRIS CMS–HS\textsuperscript{TM} [56] ultrasound-based measuring system with WINSPINE proprietary measurement software. The system consists of a sensor head comprising three transmitters, triplets with three ultrasound microphones (placed on the patient’s pelvis) and a pointer stick with two ultrasound microphones [57]. The transmitters send ultrasound pulses at regular intervals to be recorded by the microphones. The coordinate of the spine are calculated by triangulation of the spatial coordinates from the microphone data. The overall shape of the spine is automatically recorded through guiding the pointer over the processus spinosus from cervical to sacral vertebrae by the examiner. This requires an experienced physician to identify the position of the spine using the palpation method [48] to use the sensor.

The examples discussed above describe the hand-held type of sensors which is a relatively simple device and much cheaper than the use of imaging. The use of this type of sensor requires manual intervention by experienced physicians. Hence, continuous
measuring of spine posture for long durations or daily activities are not possible due to availability constraints. It is also prone to measurement errors such as human error from data reading or uneven pressure applied on the spine that results in non-reliability of the data. To minimize such problems, wearable or skin-mounted sensor are often a preferable option.

### 2.3.2.2 Wearable or Skin-mounted Sensor

The group of skin-mounted sensors allows continuous monitoring of spine assessment with minimum human supervision. The sensor is incorporated in a wearable garment, belt or directly mounted on the spine section to monitor spine curvature. An example of the wearable sensor is the strain gauge and optical mouse sensor that are placed on a thin stainless steel blade for sagittal and lateral bending [58]. The metal blade is attached perpendicular to a wearable nylon belt with the strain gauges at the middle of blade and the optical sensor at the end. The belt is worn at the waist region with the blade at the center of spine from first sacral vertebrae. Bending of lumbar spine region in the sagittal plane is measured from the change of strain gauge’s resistance value while the optical mouse sensor measures the relative horizontal movement of the spine with respect to the fixed position of the optical sensor. Despite the apparent excellent potential of this device, its size is relatively large, cannot be concealed easily and is deemed by many patients to be uncomfortable to wear. The BodyGuard™ by Sels instruments [59] is another strain gauge based sensor used for spine bend monitoring. The BodyGuard sensor consists of a strain gauge probe and a portable electronics system to process the signals. Elongation of the strain gauge sensor from spine bending changes its internal resistance and hence the output signal (voltage) of the electronic unit. The voltage output is linearly related to the elongation length and this in turn reflects the relative distance between anatomical landmarks. The motion pattern of the spine can be monitored with minimal
impact on the user’s movement [60]. The BodyGuard™ sensor is a relatively simple instrument, light weight and is minimally invasive as a spine monitoring sensor. The drawback of the sensor is that it measures the elongation of strain gauge from the spinal motion and not the angular displacement. Also it can only operate in the sagittal plane and does not allow measurement of spine in the lateral plane.

Spinal posture analysis is also possible using an accelerometer based system. A tri-axial accelerometer (KXM52-Tri-axis, Kionix) was used to detect spinal posture change in sitting positions [61]. Accelerometers were mounted on three elastic straps and worn around the upper Thoracic region T1-T2, upper Lumbar region T11 and L1 and at the level of Sacrum at S1 as illustrated in Figure 2.3. The inclination change between two sensor modules located at the thoracic and lumbar or lumbar and sacrum was derived and defined as the posture change. The tri-axial accelerometer was demonstrated to be satisfactory for detection of posture changes in the sitting position, but relatively large errors were evident for postural measurement under dynamic conditions. The same author later combined the use of accelerometers and a gyroscope to track the trunk movement in the sagittal and coronal (lateral) planes [62]. The use of additional gyroscope in the system provides additional dynamic information from the orientation of the sensor, but also adds considerable complexity and cost to the system. The system was able to achieve a more reliable monitoring of dynamic postural change in daily activities but it cannot measure the static spine posture.
Figure 2.3: Accelerometer sensor modules mounted at the region between T1 and T2, between T11 and L1, and at the level of S1 respectively using elastic straps [61].

Nevins et al. also developed a posture monitoring system using accelerometers to monitor spine movement in the sagittal plane [63]. Six accelerometers (Analog device model ADXL202E) were attached along the subject’s spine using medical bandage tape. The accelerometer inclinations of six points were collected, converted to polar coordinates and rotated to obtain the angle of spine relative to the vertical. Stick figures were developed to model the subject’s spine movement when assuming a series of postures. The accelerometer is a popular option for human movements in the medical field, due to its small dimension and light weight. Drawbacks of electronic accelerometers include drift and fluctuation in dynamic response resulting in poor quality of inclination estimates for large acceleration values and over relatively long periods of usage [64]. The same problem is associated with the gyroscope but can be overcome using filters (e.g. Kalman filter). The use of such filtering techniques adds to the complexity of the system and results in uncertainty regarding the efficacy of the data obtained.
The gyroscope provides a separate alternative as a wearable sensor for dynamic spine assessment. The gyroscopes are popular for use in joint motion measurement due to their small size, light weight, being relatively inexpensive (electronic MEMS versions) and easy to apply. A real-time 3-D movement of the lumbar spine was measured using an inertial tracking system (IS 300Pro) that comprises three orthogonally aligned gyroscope [65]. The gyroscope that attached to the trunk measured the angular change of rotation in three axes that was integrated into a 3-D orientation. The limitation of the solid-state gyroscope used is the bias that leads to drift of the angle signal when the sensor is in a stationary condition. Therefore, gravitometers and magnetometers were added into the sensor to eliminate the gyroscope drift. However, these add to the complexity and cost of the system. Chikara et al. developed a wearable prototype sensor using two perpendicular mounted dual axis gyroscopes, Invensense IDG300A [66]. The sensor is based on inertial sensor with the Body Sensor Network (BSN) platform to allow wireless communication when measuring the data. The sensor is mounted on the skin at L1 and L5-S1 vertebrae to assess the quantitative motion of the lumbar spine. Only movement in the lateral flexion was analyzed and presented in the paper. As discussed previously, the sensor has the limitation of drift error in the static condition which requires additional sensor compensation techniques or an applied corrective algorithm to reduce the drift error.

For continuous measurement of spine movement, the skin-mounted type of sensor represents a better selection compared to hand-held and imaging sensor that were discussed in previous section. It requires minimum human supervision to operate and facilitates long term monitoring assessment (e.g. monitoring the spine in daily activities). This type of sensor is mostly non-invasive, portable and is less expensive compared to the imaging sensor, thus it is potentially suitable for clinical implementation as well as home monitoring. Each sensor has its advantages and limitations as discussed above and
they are therefore designed for different monitoring purposes. Therefore, the selection of
the type of sensor that can be used is based on specifications (e.g. static, dynamic, bending
directions) and the requirements of the specific applications.

2.3.3 Summary and Comparison of Different Sensor Types

This section includes a brief summary and comparison between different spine
monitoring groups that are previously presented. Among both sensor types, the spine
imaging sensor and particularly the radiography modalities are often used in a clinical
assessment due to their gold standard status in spine diagnosis. Despite their high
accuracy and sensitivity, imaging techniques such as X-ray and CT-scan uses ionizing
radiation which poses potential risk to health. Due to the hazard risk of these radiation
devices, it is often not recommended for routine diagnosis among patients with low back
pain. MRI serves as an alternative option in the clinical spine assessment that is free from
ionizing radiation. However the cost of MRI scanning is more than other imaging
modalities (e.g. CT scans and X-Ray) [67]. MRI scan can also be ruled out for patients
with interior ferromagnetic material such as metal implants or cardiac pacemakers due to
the strong magnetic field attraction. For other field use of spine imaging sensor systems,
rasterstereography and the Vicon™ motion analysis system utilize cameras to capture the
contour profile or markers’ trajectory to monitor the spine. The rasterstereography
provides a relatively fast scanning time (40 miliseconds for a 3D scan of contour data)
than other imaging techniques [68]. This scanning technique is ionizing radiation-free but
is limited to only static spine posture measurement. The disadvantages of Vicon™ motion
analysis approach is that it requires point-by-point placement of markers at different spine
segments, rendering a potentially high risk of error from skin motion artefacts.
The BodyGuard™ system [60], other strain gauge sensor [58] and tape measure [69] are examples of reported contact type sensors being based on spine elongation at different body postures. However, the reading provides only the magnitude displacement of the skin surface (location where the sensor is mounted) and thus do not offer any information regarding angular displacement. Furthermore, such sensors that measure the longitudinal gap variation are limited to only measurement in sagittal plane. There are also systems based on Cartesian coordinate movement detection including the Spinal Mouse™ and Zebris Pointer™, which require the system to employ point-by-point reading (i.e. they need to be placed along the spine line at each segment), rendering rather complex data collection and processing. Other types of contact type sensor utilize accelerometers, inertia sensor or gyroscopes to measure the movement of spine. The sensors can provide accurate results for dynamic assessment but they suffer from fluctuation and output drift under dynamic conditions [70]. These limitations can be overcome using filters [64] or integration of additional devices such as gravimeters and magnetometers [71] to eliminate the drift. However, the size, complexity and cost added to the system also significantly increases.

Even though the hand-held sensor is a simpler and cheaper option for spine monitoring assessment, it requires human intervention to operate and hence is prone to human error from measurement taking and reading (e.g. inclinometer, tape measure and flexi-curve ruler) or unevenly applied pressure (e.g. manual goniometer). Automated hand-held sensor systems such as Spinal mouse and Zebris Medical pointer (referred to earlier in this section) provide better reliability in recording the data. Nonetheless, both these sensors require an experienced physician to guide the sensor over all the processus spinosus for measurement. Continuous or long term measuring of spine posture is not possible. Therefore, the skin-mounted contact type sensor is a preferable selection for both continuous measurement as well as static measurement assessment. It requires
minimum human supervision to operate and facilitates long time monitoring especially in clinical implementation (to detect early LBP symptom). It is non-invasive, poses no potential risk of radiation, portable and relatively cheaper compared to other systems in this class e.g. the imaging sensor.

Considering the limitations and unique specifications of each reported sensor, the selection of the particular sensor is highly dependent on the purpose of measurement and the required parameters. Although there are many varieties of spine monitoring options available (commercial and self-fabrication), none of the reported sensors offer a specification that meet all criteria as follows:

- Small and lightweight sensor probe to be mounted on different body sizes (adults or children).
- Non-invasive, portable and easy to implement using standard medical double-sided tape.
- Able to provide continuous measurement of spine assessment.
- Able to measure spine bending in both sagittal and lateral planes.
- The sensor probe is non-ferrous and can be used in special environment such as MRI or X-ray based scanning.

The POF sensor fabricated in this investigation is based on angular displacement (curvature) between the input and output fibres and is able to fulfill the criteria as listed above. Additionally, the same sensor can operate as a respiratory monitoring sensor. The respiratory sensor also has the above mentioned advantages and can be simply operated by attaching at different attachment points at the upper body (including front and back body). Consequently, the proposed POF sensor offers an excellent solution for the monitoring of physiological parameters (spine monitoring and respiratory monitoring) in clinical environments. The currently available respiratory monitoring sensor in different sensing categories is discussed in the next section (Section 2.4).
2.4 **Respiratory Monitoring Sensors**

Respiration rate or frequency of breathing provides an important physiological vital sign that has been shown to be a reliable indicator of the presence of illness or physiological state of a patient. Breathing problems in patients can usually be associated with lung function problems. Examples of lung disease include Chronic Obstructive Pulmonary Disease (COPD) where less air enters in and out of the airways. It is a progressive lung diseases which includes asthma and chronic bronchitis (inflammation of the airways). Symptoms seen in patients with COPD include shortness of breath, increased breathlessness, coughing and wheezing. It is a significant cause of mortality. Even though COPD is not fully reversible, it is treatable and can be prevented from occurring [72]. The right treatment of pulmonary rehabilitation also helps the patient with COPD to breathe better.

Monitoring the respiratory patterns can help in anomaly detection and prevent such chronic respiratory diseases. Other breathing abnormalities symptoms include occasional cessation of breathing and irregular breathing rate. Example of diseases that can be diagnosed from the irregular breathing pattern include Apnea, Tachypnea and Hypopnea [73, 74]. Apnea is the temporary cessation of breathing (especially during sleep). Tachypnea is the abnormal rapid respiration and Hypopnea refers to the abnormal deep breathing. All these breathing symptoms can be detected and analyzed from monitoring the respiration signal. Respiratory rate also deviates according to the physical activities such as running, walking or sleeping. Changes in respiration rate due to O$_2$ demand or CO$_2$ production reflects the health status. For example, an increase in acid level in the body leads to the increase in the rate and depth of breathing in order to eliminate more CO$_2$ and reduce the acidic level.
In the early days, respiratory signal measurement in a clinical setting was based on manual counting. The physician visually observed patient’s chest movement or detected the air flow from the nose to estimate the breathing rate based on breathing movement. However, it was impossible to sustain this measurement for long periods and the results were often unreliable and not accurate [75]. Failure to continuously monitor the respiration signal in an accurate manner might cause severe after effects that may hinder the recovery of patients. Hence, there is a clear need for reliable and accurate automated (e.g. computerized) respiratory monitoring equipment to overcome such limitations. Current breath detection sensors use a variety of techniques such as humidity [76, 77], temperature [78] and movement of the chest due to respiration [79] to detect the breathing pattern. Generally respiration sensors can be categorized into the contact type (wearable, skin-mounted sensor) and non-contact type. Both types of sensor have their own advantages and disadvantages depending on the required application.

2.4.1 Non-contact Respiratory Monitoring Sensor

Non-contact respiration sensors have included the use of infrared video cameras [80], charge coupled devices (CCD) video cameras [81] or microwave radar antenna [82] to track the breathing signal. The sensing mechanism is based on capturing and processing the image from breathing movement or transmitting the signal to and from the body surface and analyzing the received data. The device measures the respiration signal without making any skin-contact with the subject or mounting the sensor on the body. The first reported example of non-contact breathing device used a high precision, single-point infrared temperature sensor [80]. A camera was used in positioning the infrared sensor to target the subject’s face and nose. The infrared sensor read the temperature changes in the sub-nasal skin surface corresponding to inspiration and expiration. The
temperature data was continuously recorded from the infrared sensor and a sinusoidal curve-fitting function was used to estimate the breathing frequency. The test results demonstrated that the system effectively captured breathing rate with an error of less than two breaths per minute. Currently the main limitation of the method is in the positioning of the sensor (anomalous frequent nose detection problems) and being able to automatically compute the breathing rate.

A thermal based imaging respiratory sensor was also developed using the FLIR A40 thermal camera [83, 84]. The camera was fixed on a tripod and placed one meter in front of the test subject. The respiration signal was detected from the changes of skin temperature caused by respiration. The region of interest (ROI) in the test was set to be circular skin area centered on the tip of the nose. A tracking algorithm was designed to locate the ROI when the subject was requested to perform small, random and regular head movements. The sensor successfully detected the respiration process through recording the thermal video images at ROI. The drawback of the sensor was the failure of ROI tracking when the subject opened his/her mouth during the recording or if the ROI went too far towards the edge of the recording region. The same sensor system was also tested on children as subjects [85]. The image was recorded for two minutes when the children slept comfortably on a bed. The results from thermal imaging were closely correlated with a separate conventional contact respiratory sensor. Further improvement is required for the occurrence of errors with signal detection when the subject performed mouth breathing, large head movement or when the subject wore glasses.

The sensors discussed above all used thermal detection where a change of temperature during breathing was detected near the nose or face region. Optical based respiratory sensors offer an alternative non-contact approach where a camera and illuminated reflective markers are used. One example of the sensor used two sets of CCD cameras with an Infrared light source to capture the respiratory motion [86]. The subject
was requested to wear a stretch T-shirt with nine sphere shape retro-reflective markers placed on upper chest, lower chest and abdomen. The markers serve to reflect the incident light source back to the recording camera. The image coordinates of reflective markers were captured using the camera in the presence of the infrared illumination. The respiration signal was thus obtained from the displacement of the markers due to body surface motion. Nakajima et al. also developed a real-time monitoring of respiratory rate for subjects in bed using a CCD camera suspended from the ceiling and a PC equipped with a high-speed image processor [81]. Respiration rate was measured by observing the blanket or chest movement of the patient from the camera’s projection. Since the sensor detects the respiratory signal using an infrared optical image of the chest region, it was essential to maintain uniform lighting conditions throughout the recording period. A similar sensor setup was developed using a single visible light camera to track the motion of color fiducials on the patients’ abdomen [87]. Three colour fiducial cubes were mounted on a velcro belt and worn by the patient around the abdomen. The colour of cube was chosen before being attached to the belt based on the best colour for tracking in the environment (e.g. due to lighting conditions, contrast etc). The camera was placed on an adjustable arm and tracked the cube’s locations (movement) in the image when subject breathed.

In summary, the non-contact respiratory devices do not physically constrain the patients and provide more comfortability because the sensor is not placed in contact with the body. However, the device usually requires monitoring by a human operator and it is not possible for monitoring breathing in daily activities as the patient would need to stay within the sensor or camera detectable region. Also, lighting conditions in the room for the reflection measurements are particularly important to obtain accurate and reliable data. Additionally, some of the sensors require a special room for the camera installation and thus it is non-portable and the cost of equipment and data processing are very high.
Contact type respiratory monitoring sensor could potentially provide a good alternative for low-cost, portable and daily sensing.

2.4.2 Contact type Respiratory Monitoring Sensor

For the contact type sensor, the device is attached, wearable or directly in contact with the patients body. This type of sensor can be classified into two sub groups based on the sensing technique. The first type of sensor measures the airflow properties such as temperature, humidity, sound or movement of the air as it travels into and out of the lungs. The sensors are usually placed around the nose or trachea region and take the form of a nasal-oral mask, nasal mask, trachea intubation or nasal cannula. Some of these configurations are considered to be quite invasive. The second type of contact sensor monitors the breathing signal by measuring changes in body movement that occurs during respiration. This type of sensor is usually in the form of a belt or wearable textile form. The sensor is usually mounted around the chest or abdomen area to detect the changes of body circumference or pressure changes due to respiration.

2.4.2.1 Airflow Sensing

The respiration signal can be detected from the airflow as the exhalation gas has higher temperature, humidity (more water vapour) and contains more CO\textsubscript{2} gas. Changes in temperature or humidity between inhaled and exhaled gas is detected and converted into a respiratory signal. It is also possible to detect the signal from the sound caused by the airflow turbulence during respiration. In the latter scenario, a sensor is placed over the throat to estimate the respiration rate from the recorded breathing sound and converts the sound into an electrical signal.
The most common tool available to a physician to measure the respiratory rate in a typical hospital setting is the stethoscope. A stethoscope is used by placing a sound collector (diaphragm or bell) directly on the skin of patient to auscultate the breath sound. The use of a stethoscope is an essential part of the clinical respiratory assessment which allows practitioners to inspect the patient’s cardiac and respiration condition [88]. From the auscultation, causes of abnormal or decreasing of breath sound due to lung disease and airway damage can be deduced. Only expert practitioners with sufficient training are able to detect the abnormal breath and heart sound with stethoscope. Other medical systems that are used in hospitals for monitoring respiration rate include the Acoustic Respiration Rate transducer (RRa) by Masimo inc. [89]. The adhesive respiratory acoustic sensor that is connected to the RRa monitor is placed on the neck of the patient (near the trachea) to detect the vibration associated with respiration. The sensor uses an integrated acoustic transducer to detect the acoustic signal from the breathing activity. This sensor is usually used on a patient that is sleeping to continuously measure their respiration rate. Sanchez Morillo et al. [90] developed a contact respiration sensor based on an accelerometer. The accelerometer was placed around the neck (between thyroid cartilage and the superior third of the breastbone) to obtain the acceleration produced from cardiac signals and acoustic vibration of respiratory movements. The cardio respiratory signal obtained from the sensor is useful for diagnosis of abnormal breathing phenomena during sleep.

The three sensors reviewed above monitor the respiration signal based on breath sound or acoustic vibration from ventilation activity. Capnography is an example of ventilation monitoring sensor that operates based on detection of the breathing gas. It is used particularly when the patient is receiving anesthesia or in intensive care [91]. Capnography detects the partial pressure of CO₂ during respiration to provide the respiratory condition of a patient [92]. The sensor uses an Infrared light beam that passes
through a sample of the gas to an IR detector. An increasing concentration of CO₂ gas more strongly absorbs the IR light and results in the decrease of detected IR intensity. The capnography can be placed directly into a breathing circuit (mainstream capnographs) or by drawing a sample of the gas to a separate sensor to detect the concentration of CO₂ (sidestream capnographs). It is a non-invasive technique and is able to continuously display the concentration of inhalation and exhalation of CO₂ over time. A disadvantage of this method is prior calibration is needed with a previously analyzed standard gas mixture [93].

The Linshom Respiratory Monitoring Device (LRMD) [94] is an example of a temperature based contact-type respiratory monitoring instrument. The LRMD sensor utilizes three medical-grade thermistors (1.5mm diameter) mounted on a face mask or nasal prongs to detect the temperature of the gas flow. Two of the thermistors are positioned near the mouth and nose region and the third acts as a reference and is located outside the face mask to measure the ambient temperature. The sensor works by detecting the warm temperature of exhalation during respiration activity and has been used to demonstrate a reliable measurement. Another similar low cost temperature sensor was developed by Basra et al. to detect the respiration signal [95]. In this case the sensor probe is clipped at the nostril with an analog temperature transducer placed at the tip of the nasal region. Two analog temperature sensor ICs (AD590) are used to measure the thermocouple voltage resulting from ambient temperature and the temperature change from respiration. The temperature difference is converted into a voltage signal to obtain the respiration cycle. The sensor is suitable for detecting sleep disorder and abnormal breathing but is not able to obtain any signal when the subject is breathing through the mouth.

One problem associated with the contact type measurement using airflow sensing is the discomfort to the patient when wearing the sensor [84]. Patients that wear the
sensor’s mask or intubated have problems with talking during the measurement of respiration. Moreover, some sensors require the change of mouthpiece / disposable mask after each use due to hygiene issues. Another type of contact sensor is discussed in the next section which utilizes the movement (expansion and contraction) of the chest wall to detect the respiration signal.

### 2.4.2.2 Movement of Chest, Abdominal and Volume Detection

For this group of sensor the movement of the chest, abdomen region and variation of thorax volume when a subject breathes is detected to obtain the respiratory signal. The rib cage and abdomen section of chest wall are strongly related with the breathing activities especially in volume displacement and motion [96]. Therefore, the sensor can be strategically placed at the abdomen and chest wall to detect the movement using a wearable textile or belt.

The first reported example of respiration monitoring sensors utilized the so-called wearable technologies, where the sensor is integrated into a textile form. The textile based sensor was used to monitor individual health signal for daily activities and was capable of detecting any early abnormal conditions. L.Guo et al. developed a wearable garment for monitoring the breathing signal that consisted of two elasticated conductive coated straps (from ELASTOSIL LR 3162) and these were sewn together using conductive threads that functioned as data transmission wires [97]. The conductive coating sensor (piezoresistive sensor) was placed on the chest and abdomen to detect the breathing signal when the chest/abdomen extended. The sensor contains electronic components with a Bluetooth transmission device at the back of the garment to transmit the data to PC for further analysis. The textile substrate used has good elasticity to ensure that the fabric kept track with the mechanical chest motion. An important factor to consider when using
wearable textile sensors is that the size and shape of the individual patient may not exactly fit with the garment.

Respiratory inductive plethysmograph (RIP) is an example of a belt type sensor for monitoring ventilation and breathing patterns [98, 99]. It consists of an inductive sensors and two sinusoidal wire coils placed on a tight fitting elastic bands. The elastic belt transducer is worn around the chest and abdomen to measure the breathing signal from changes of chest circumference during breathing. By measuring the rib cage and abdomen displacement using dual RIP bands, tidal volume can also be calculated from respiratory movements. The drawback of using such sensor is that the patient often finds that the chest band causes discomfort when worn over a long periods [100]. Other similar alternative versions of belt type sensors includes the use of an extensometer to measure change in the chest wall circumference [101]. The extensometer consists of an ultrasonic transmitter that measures displacement in circumference that is covered by a rubber tubing. Using two extensometers (one at thoracic and another at abdominal), tidal volume can be estimated. Hung and Phan et al. separately developed a respiratory monitoring sensor using an accelerometer that was attached to a flexible belt [102, 103]. The sensor belt with biaxial accelerometer was strapped on the chest to detect the periodic movement of thorax from breathing. Changes in the accelerometer’s inclination arising from the breathing movement was processed to obtain the respiratory waveform signal. Since the accelerometer is very sensitive to any body movement, it is necessary for the subject to stay still when taking the measurements.

Gaidhani et al. developed a respiratory motion sensor that was designed for use while performing sports or exercises and is based on an inertial sensor [104]. The sensor utilized two IMUs (inertial measurement units) that were placed on the chest and posterior side (back) of the chest. One sensor majorly detects the body motion while another sensor measured both body motion and respiratory motion. A mathematical algorithm was
proposed to eliminate the influence of body motion on the breathing signal. For on-bed respiratory monitoring, MC Huang et al. developed a non-invasive e-textile bedsheet with transducer incorporated to detect the breathing signal [105]. The pressure sensitive bedsheet converts the pressure exerted by the resting human body (in particular the chest area) to the bedsheet and subsequently into an output voltage signal. The respiration signal waveform was extracted from the quantifying pressure distribution sequence. The sensor demonstrated an accurate measuring of respiratory signal but the measurements only applied to bed-bound patients. The monitoring system also failed to identify an accurate breathing signal when the subject was moving or changing posture. This is due to the large non-respiratory signal which was present when the subject changed sleep position. The authors claim that this problem will be solved in future work.

In summary, sensors which respiration measured using direct contact with patient tend to be more accurate but provide less comfort. Thus, it is important for the contact type sensor to be small-size, light-weight and comfortable to wear regardless of the size of the patients. In the research of this investigation, the fabricated sensor is small, non-invasive, and flexible and therefore provides better comfort compared to other wearable respiration sensors. Moreover, the sensor developed in this investigation is not restricted to be used only on the front chest and abdomen region, which provides more freedom for the user (practitioner) depending on the conditions under which the use is to take place.

2.5 Optical Fibre Bending Sensor for Health Monitoring

An optical fibre is a cylindrical dielectric waveguide that consist of center core surrounded by an annular cladding with a slightly different (smaller) refractive index value than the core. Light propagates along the fibre through multiple total internal reflection at the core cladding boundary. Optical fibre came to the fore as it has been widely used as telecommunication medium to transmit information due to its bandwidth
transmission capability. Optical fibre applications have extended into many other application areas including lasers, sensors, and high power transmission. The general structure of optical fibre sensor system comprises a light source, the sensor itself and a detector as represented in Figure 2.4. The selection of the light source (e.g. monochromatic or broadband light source) and detector (e.g. photodetector, Optical Spectrum Analyzer) depends on the wavelength region selected and constraints of the sensing technique e.g. technical and/or cost.

![Figure 2.4: Block diagram of optical fibre sensor system.](image)

Compared to other types of sensors (e.g. piezoelectric, thermocouple, inductive, capacitive), optical fibre sensors have the advantages of small size, light weight, potentially low cost and immune to external electromagnetic interference (EMI) [106, 107]. An example of a wearable POF sensor for monitoring seated spine posture was designed and evaluated by Dunne et al. [108, 109]. The sensors were fabricated according to the technique modified from POF sensor developed by Kuang et al. [110]. A POF was abraded on one side to allow light leakage from the polished region. One end of fibre was connected to an LED light source with a detector (light-dependent resistor and data acquisition) located at the opposite end. Bending of the sensor affects the amount of light leakage through the cladding boundary and thus changes the output intensity. The POF was stitched to an elastic garment using a loose zig-zag stich to allow the sensor to move vertically along the garment surface. This sensor was suitable for daily use such as outdoors but was not ideal for testing in clinical environments. Additionally, the use of
the garment might sacrifice the user’s comfort and appearance. Calibration of the system for each subject was also necessary to minimized variables from poorly-fitting garment (sensor) due to the difference back length of the subjects.

The same sensing technique using side-polished fibre and Bluetooth technology wireless system was applied to design a knee sagittal monitoring sensor [111]. The POF sensor was integrated into a knee support and this was used to monitor the joint movement from the transmittance changes during bending. The sensitivity of the sensor was enhanced using the side-polished fibre as it resulted in an increase of the light attenuation when undergoing macro bending. The wireless POF based sensor demonstrated a valid and reliable performance and allows freedom to its users. The limitation of using side polished fibre is the limited mechanical strength of the fibre and aging potentially reducing the lifetime of sensor. Additionally it was noted that bending of the side polished sensor at large angles or overstrain permanently damages the fibre sensor.

Another example of an optical fibre spine monitoring sensor was fabricated using 8 paired fibre-optic loops with 60 mm spaced apart and attached to a ribbon of elastic sprung steel [112]. A reference sensor was attached at the sacrum S1 position and the data of each fibre sensor with respect to the base reference was collected. The sensor measures spinal motion through the change of transmitted light intensity. The authors claim a good performance from the sensor but the use of steel ribbon might cause discomfort to the patient and prohibits the sensor to be used in X-ray or MRI scanning environments. Zawawi et al. [113] developed an intensity modulated optical fibre sensor for spine monitoring. The sensor measured the bending of the spine from the change of light intensity using two source fibres and one reference fibre that were bundled together. An aluminum foil with a small window was placed at the end tips between source fibre bundle and output fibre to allow part of the source light to be transmitted to the output fibre and partly reflected back into the reference fibre. The sensor measures the bending
of the spine angle based on changes of detected intensity (voltage) ratio between an output fibre and reference fibre. The device was demonstrated to be accurate and applicable in a clinical setting. However, the long term stability of device was found to be inadequate for repeated use.

A further example of a POF based bending sensor includes the fabrication of POF based goniometer [114]. The sensor works based on the principles of chromatic modulation, where the spectral distribution power of light was modulated at the bending axis of goniometer made from an angled glass wedge modulator. Changes in the spectral power distribution due to sensor’s bending was measured over a range of bandwidth using two or more specific wavelength response of photodetectors. Output from the individual photodetectors were combined to obtain a relationship between intensity over the formed (colour) signal and the angular position goniometer. The POF goniometer was able to measure angular position of a patient’s limb (joint) in an MR scanner environment. A disadvantage of this sensor was that the operating range is limited to 0-90° and has a reduced sensitivity between 0-5° and 85-90°. The accuracy also depends on the precise aligning between pivotal axes of sensor with the pivotal axis of a joint. It was suitable for measuring large joint e.g. knee, but in the form reported [114] was much too large to be incorporated as a spine bending monitor.

2.6 Optical Fibre Sensor for Respiratory Monitoring

Apart from the spine or gait monitoring sensor, optical fibre is also ideally suited to be used in respiratory monitoring. The good sensitivity, small size, water resistance and low weight of fibre allows miniaturization of the sensor probe for enhanced comfort in the case of long-term monitoring [115]. Its immunity to electromagnetic field also
allows the sensor to be used in hospital environments [116] and in conjunction with MRI and CT scanning environment [117].

An example of OFS for respiratory monitoring was included in a plastic mask equipped with an optical fibre that detects the breathing pattern through the relative change in humidity between inhalation and exhalation [118]. The sensor was developed using Agarose infiltrated photonic crystal fibre interferometer (AI-PCFI) which is a relative humidity sensor. The setup comprises a light source, a fibre optic coupler, the AI-PCFI and an optical detector. AI-PCFI consists of a photonic crystal fibre (PCF) fused to a single mode fibre. The other free end of the PCF is infiltrated with Agarose. The refractive index of Agarose changes with humidity and in turn alters the modal propagation constant of cladding mode. A resulting phase change between the interfering core and cladding modes causes a shift in interference pattern. The breathing pattern can be observed from the change of spectral shift caused by changes in humidity due to exhalation. Li et al. also developed a mask type breathing sensor based on an optical fibre interferometer [119]. A single mode- multimode- singlemode (SMS) fibre structure was attached on a thin plastic film in an oxygen mask. Air flow from the mouth during breathing introduced a pressure to the plastic film and this bent the SMS structure which was sensitive to the macro bending. The light power variation at the output of SMS fibre structure directly shows the breathing signal. Both of the sensors described above feature advantages of compact size, ease of fabrication and a fast response time. Disadvantages of the mask type sensor include its price and non-reusability. The mask sensor needs to be low cost and disposable due to the contamination of the sensor head from breath condensation after each use.

An OFS respiratory monitoring transducer mounted on a polymer rubber elastic band was demonstrated by Augousti et al. [120]. The sensor was developed based on the
macro-bending loss effect in optical fibres. The optical fibre was modified/bent into a figure-of-eight coil knot to monitor the circumference of the chest and abdomen. Based on elongation changes on the chest movement, the radius of curvature of the knot changes and this varied the detected signal intensity. The signal demonstrates satisfactory result on breathing pattern and rate. Nonetheless, the sensor has the tendency to overstate the amplitude of the breathing signal when measuring a large shift of tidal volume during the calibration of the sensor. Another fibre sensor based on macro-bending effects for respiratory monitoring was developed using a fibre that was stitched onto a normal T-shirt [121]. The fibre was stitched into a sinusoidal shape with small radii of curvature. Light transmitted along the fibre is attenuated based on the radius curvature of fibre which was affected by the body motion. A respiratory signal was obtained by analyzing the bending of the fibres arising from the breathing movement of the wearer’s upper body. The sensor shows an apparent breathing signal for slow and normal breathing. However, some interference was present in the signal caused by non-respiratory motion and the output voltage (intensity) value drift caused by different breathing modalities or body postures. The authors have indicated that there is scope to improve this situation using a noise filter algorithms and compensation for the drift of output voltage in future investigations.

A textile breathing frequency sensor was also developed based on the changes of distance between input waveguide and output receiver caused by changes of chest movement [122]. The sensor head consists of two optical fibres (transmitting and receiving waveguides) that were positioned in one horizontal line. Each fibre was surrounded by slidable sleeves and mutually connected to a textile spring. One fibre was connected to an LED source while the other was connected to a distal photodiode. The sensor has a band that wrapped tightly around the chest and can expand a maximum of 2 cm distance when stretched on a human thorax. The author claims that the sensor is simple
to construct and can be easily implemented into clothing textile. M. Krehel et al. also fabricated a textile based respiratory monitoring sensor using four custom made optical fibres [123]. The fabricated Geniomer fibre are multimode optical fibres that are fabricated from a different percentage of soft silicon and hard polyurethane [124]. The fibres with high flexibility and mechanical strength were integrated into a polyester textile for respiration monitoring. The sensing textile was sewn into a chest strap, with rubber straps and buckles to allow attachment onto the chest of subject. The Geniomer fibre with different diameter was folded into a half oval form and tested with different fibre lengths. The sensor shows a good correlation when compared with a commercial respiratory monitoring sensor.

For the measurement of body motion, T.D.P. Allsop et al. developed an embedded progressive-three-layered fibre long-period gratings (LPG) for respiratory monitoring [125]. The progressive three-layered single-mode fibre was embedded into a flexible platform as a curvature sensor. The bend sensor was tested on a resuscitation training manikin and showed good sensitivity to be part of a respiratory monitoring system. However, the working principle of the sensor utilized LPG that is temperature sensitive and might lead to a small error when the sensor is in contact with the skin when there is a skin temperature variations. The authors suggest the use of a referenced Bragg grating temperature sensor to reduce the error from the temperature variations. However, it is clear that the addition of a reference grating would further increase the complexity and cost of the system. The authors later incorporated a series of in-line curvature sensors on a garment (stitched to a Lycra vest) and shows reasonable agreement in the volumetric tidal changes when simultaneously compared with a spirometer [126]. Another fibre Bragg grating (FBG) based smart textile for continuous monitoring of respiratory movements was developed by J. Dejonckheere et al. [127]. Due to the sensitivity to strain, the sensor is able to detect elongation from the shift of the Bragg wavelength. The setup
consists of a broadband source and an Optical Spectrum Analyzer (OSA) to track the wavelength shift when a mechanical strain was applied. The FBG was stitched on an elastic fabric and was tested on a simulator that performs movement similar to a real breathing activity. The sensor demonstrates a high sensitivity and stability performance. The preliminary tests on healthy volunteers have also shown the feasibility of the sensor being used to obtain typical respiratory signal from both thoracic and abdominal movements.

In conclusion, optical fibre sensors are particularly attractive for use in many medical environments. Apart from the aforesaid advantages of optical fibre, the sensor can be easily replaced, is robust, provides water resistance and possesses intrinsic safety. Therefore, it has been selected for use in the research of this investigation for monitoring both spine bending and the respiratory signal. In the current work, an intensity modulation technique is used to measure the spine bending and breathing signal through measuring the change of an output intensity ratio at the ends of three different fibres. This approach was preferred compared to other interrogation technique due to its simplicity and potentially low cost [128]. The sensing setup utilizes a low cost LED, standard POF and a simple optical detector being a Silicon photodiode and amplifier. The modulation method is implemented through the misalignment loss (fibre angular displacements) between input and output fibre at joint axis [129]. It measures the change of light intensity at the output fibre that is varied (attenuated) due to the change of bending angle at the sensing region. Even though the intensity modulation technique is commonly susceptible to source intensity fluctuations which in turn lead to potentially false readings, in this investigation the variation or loss in source intensity and environment is automatically compensated without additional referencing system i.e. the sensor is self-referencing. Further details of the POF sensor design and configurations are included in the next chapter.
CHAPTER 3 THEORETICAL BACKGROUND AND SENSOR DESIGN

3.1 Introduction

This chapter includes the underlying theory, design, concept and the working principles of the POF sensor of this investigation. The sensor works based on intensity modulation utilizing four (one input and three output) multimode POFs. The sensor estimates the bending angle from the power transmission ratio between the three receiving fibres tilted at an angle relative to the input fibre. An analysis of the power transmission loss between two optical fibres with lateral and angular misalignment is investigated in this section. A power coupling equation is obtained which is used to simulate the output ratio at different bending angles and the resulting values are further used for comparison with the experimental results of chapter 5. Various prototypes and designs of the sensor probe have been developed to fit the applications and to optimize the device performance. The design of all electronic interface circuits, including an LED driver circuit (light source) and a photodetector conditioning circuit are discussed in this chapter. The detection hardware and software (Data acquisition system, LabView™ and an Arduino based platform) used to convert the output signal to a digital form, capture it and perform analysis are presented in detail.

3.2 Power Coupling between Two Multimode Fibres with Lateral and Angular Misalignment

Extrinsic loss often occurs when two end face of optical fibres are geometrically misaligned during splicing or temporary joining. Some intensity sensors utilize the transmission losses from angular misalignment between two fibres to measure the bending angle [130]. An analysis of power coupling between two multimode fibres with
angular misalignment has previously been presented by Opielka et al. [131]. Gao et al. presented a similar angular misalignment analysis but with different numerical apertures (NA) between the input and output fibres [132]. In this section, the power coupling between multimode fibres with combined lateral and angular misalignment is calculated based on a theoretical consideration and geometrical optics of the power coupling between the step-index multimode fibres used in the sensor of this investigation. The result of the theoretical predictions is beneficial for calculating the transmission loss or power coupling between fibres with lateral and angular misalignment at the fibre coupling joint.

The fibre sensor in this investigation works based on the light intensity modulation from the tilting angle variation of three output fibres that are laterally offset to align to the center axis of the input fibre. The tilt angle at the coupling region causes an alignment mismatch between the input and output fibres and hence the optical power is redistributed between the receiving fibres. The power coupling equation analyzed is used to calculate the theoretical result and to evaluate the experimental data. The characteristics of the sensor configurations are included throughout the theoretical calculation process.

### 3.2.1 Definition and Assumption

The power coupling between the input and output fibres was calculated based on the graphical explanation as illustrated in Figure 3.1. The figure shows two fibres where fibre 1 is the source transmitting fibre and fibre 2 is the receiving fibre. Both fibres meet at plane S with a lateral misalignment displacement, d and tilted at an angle, α. The theoretical analysis is made based on several assumptions as outlined below:
1. All fibres considered in the analysis and used in the experiment are identical step-index multimode fibres with same numerical aperture (NA) and core diameter. In this investigations, the multimode fibre used has the NA of 0.51 and core diameter of 0.98 mm.

2. A uniform mode distribution (UMD) is assumed in the multimode fibre.

3. The effect of leaky modes are neglected.

4. The coupling is ideal except for the loss caused by the lateral and angular misalignment.

![Graphical representation of the lateral and angular misalignment between two fibres.](image)

Figure 3.1: Graphical representation of the lateral and angular misalignment between two fibres.

### 3.2.2 Calculations and Equation Results

The power coupling efficiency of fibre 2 attributed to lateral and angular misalignment can be calculated by integrating over the solid angle, $\Omega$ and area overlap in the plane $S$. Figure 3.2 shows the overlapping cones of NA from both fibres where the shaded area is integrated using the defined parameters and geometry. Generally, the power transmitted within fibre 2 can be written as:

$$P_2 = \int_S \int_{\Omega_2} L \cos \theta d\Omega_2 dS_2 \quad (3.1)$$

where $L$ is the radiance based on power distribution in the plane $S$ and $\theta$ is the radiating area element within the overlap area from fibre 1. For the case of uniform power
distribution, $L$ is constant in the overlapping region. Considering the inner integration of equation 3.1, the power density, $P_2'$ is initially evaluated as:

$$P_2' = \int_{\Omega_2} L \cos \theta \ d\Omega_2$$  \hspace{1cm} (3.2)

The definition of solid angle, $\Omega$ at the overlapping area is represented by:

$$\Omega = \frac{A_0}{R^2}$$  \hspace{1cm} (3.3)

Referring to Figure 3.2, the overlapping region of the solid angle, $A_o$ at the NA cones is integrated for each equal quadrant following the parameter shown in Figure 3.2(b):

$$\Omega_o = \frac{4}{R^2} \int_{\beta=0}^{\beta_{\text{max}}} \int_{\rho_{\text{min}}}^{\rho_{\text{max}}} dA_o$$  \hspace{1cm} (3.4)

and the area element $dA_o$ is described as:

$$dA_o = R^2 \sin \theta \ d\theta \ d\beta$$  \hspace{1cm} (3.5)

Figure 3.2: Overlapping cones of numerical aperture at (a) side view and (b) plan view.
For the case of combined transverse offset and angular misalignment coupling, the limit of the inner integration within the overlap area of one quadrant is calculated as illustrated in Figure 3.2(b) such as:

\[ \beta_{\text{max}} = \arccos \left( \frac{L/2}{\rho} \right) \]  

\[ L = \sqrt{(y+t)^2 + x^2} \]  

\[ L = \sqrt{y^2 + (x+t)^2} \]  

where \( t = R \sin \alpha \)  

where value \( x \) is the lateral displacement in x-axis from the input fibre F1, value \( y \) is the displacement distance in y-axis relative to the input fibre and value \( t \) is the shifting of F2 cone at plane S caused by angular misalignment in y-axis. For the case of the angular misalignment in y-axis, equation 3.7 is applied in the integral limit while equation 3.8 is applied for the angular misalignment in x-axis. The limit \( \beta_{\text{max}} \) can be simplify as:

\[ \beta_{\text{max}} = \arccos \left( \frac{\sin \gamma/2}{\sin \theta_c} \right) \]  

where \( \gamma \) corresponds to the angle between the radiation extended at the lateral and angular misalignment and the normal of emitting surface as presented in Figure 3.2(a).

\[ \rho_{\text{min}} = R \frac{\sin \gamma/2}{\cos \beta} \]  

\[ \sin \gamma = \frac{L}{R} \]  

\[ \rho_{\text{max}} = R \sin \theta_c \]  

\( \theta_c \) is the maximum acceptance angle for both identical fibres and corresponds to the angle \( \theta \) in Figure 3.2(a). The limit \( \beta_{\text{max}} \) is the maximum angle of one shaded quadrant. \( \rho_{\text{min}} \) and \( \rho_{\text{max}} \) represent the maximum and minimum radius of the shaded area. Substituting the integration limit and the overlap region of solid angle into equation 3.2, one obtains:
\[ P_2' = 4 \int_{\beta=0}^{\beta_{\text{max}}} \int_{\rho_{\text{min}}}^{\rho_{\text{max}}} L \cos \theta \sin \theta \, d\theta \, d\beta \]  
\text{(3.14)}

Substituting \( \sin \theta = \rho/R \), the inner integration in power density equation is replaced with the relations of:

\[ \sin \theta_{\text{min}} = \frac{\sin Y/2}{\cos \beta} \]  
\text{(3.15)}

\[ \sin \theta_{\text{max}} = \sin \theta_c \]  
\text{(3.16)}

Solving the equation 3.14 by replacing the limits of integral above will get the equation:

\[ P_2' = 2L(\sin^2 \theta_c \beta_{\text{max}} - \sin^2 \frac{Y}{2} \tan \beta_{\text{max}}) \]  
\text{(3.17)}

The transmitted power in fibre 2 is found by integrating over the area element at endface of the input fibre as follows:

\[ P_2 = \int_0^{r_{\text{max}}} \int_0^{2\pi} P_2' \, d\theta \, r \, dr \]  
\text{(3.18)}

The \( r_{\text{max}} \) value represents the boundary radius where the acceptance cone of both fibres no longer overlap due to lateral and angular misalignment. For the solution of the step index fibre, the \( r_{\text{max}} \) value is the radius of the multimode fibre, \( a \). Hence, the power coupling in fibre 2 including the transmission loss from lateral and angular misalignment can be written as:

\[ P_2 = 2\pi a^2 L (NA^2 \beta_{\text{max}} - \sin^2 \frac{Y}{2} \tan \beta_{\text{max}}) \]  
\text{(3.19)}

Simplifying equation 3.10 into:

\[ \beta_{\text{max}} = \arctan \left( \frac{\sin^2 \theta_c}{\sin^2 \frac{Y}{2} - 1} \right)^{1/2} \]  
\text{(3.20)}

The transmitted power in fibre 2 at the overlapping region is found as:

\[ P_2 = 2\pi a^2 L (NA^2 \arctan \left( \frac{\sin^2 \theta_c}{\sin^2 \frac{Y}{2} - 1} \right)^{1/2} - \sin^2 \frac{Y}{2} \left( \frac{\sin^2 \theta_c}{\sin^2 \frac{Y}{2} - 1} \right)^{1/2}) \]  
\text{(3.21)}
The total power in transmitting fibre 1 is:

\[ P_1 = \pi^2 a^2 L N A^2 \]  

(3.22)

Finally, the coupling efficiency, \( \eta \) is calculated with the ratio between power received in fibre 2 over the total power of transmitting fibre.

\[ \eta = \frac{P_2}{P_1} \]  

(3.23)

\[ \eta = \frac{2}{\pi N A^2} (N A^2 \beta_{\text{max}} - \sin^2 \frac{\gamma}{2} \tan \beta_{\text{max}}) \]  

(3.24)

In the theoretical representation of the POF sensor in this investigation, the power transmission for each output fibre is calculated using equation 3.21. The output ratio between three lateral offset output fibres is performed using this equation. The results have been compared with the experimental data for a range of angular misalignment values and will be discussed further in chapter 5.

### 3.3 Sensor Configuration and Concept

The sensor’s working principle is based on an intensity modulation technique where the coupled light intensity changes when the sensor is deformed (bent). The power intensity loss from the angular misalignment varies with bending angle. A sensor based on similar concept has been previously proposed by Zawawi et al. of this research centre using three fibres at the output and one input fibre connected to the light source [133]. The use of three optical fibres to receive the light signal allow the measurement of bending angle in sagittal and lateral directions as defined in Section 3.3.1 of this thesis. The proposed method enables self intensity compensation without the requirement for a separate referencing fibre in the sensor arrangement. The sensing mechanism also allows compensation for variations in light source intensity and losses of the input fibre.
3.3.1 Spine Bending Monitoring

Three fibres are bundled together at the output to determine the bending of the human lower spine for lateral and sagittal directions as demonstrated in Figure 3.3. Lateral bending, also known as frontal plane motion, is the side bend from the waist to either left or right directions. Sagittal bending is the movement in sagittal plane, where the human body bends to either forwards (flexion) or backwards (extension). The fibre configuration in this investigation is designed to detect the bending angle of the spine movement in these two bending directions.

Figure 3.3: Lateral and sagittal bending activities of human body.

Figure 3.4 shows the fibre alignment configuration and the offset coordinate for the input and output fibres in the middle position and after bending.
The lateral offset coordinate of the three output fibres calculated from the center axis of the input fibre coordinate is:

\[
I_n(x, y) = \begin{cases} 
I_1(-r, -\frac{r}{2\cos30}) \\
I_2(r, -\frac{r}{2\cos30}) \\
I_3(0, \frac{r}{\cos30})
\end{cases}
\] (3.25)

where I1, I2 and I3 refers to the three output fibres as presented in Figure 3.4 and \( r \) is the radius of each identical fibre which is 0.5 mm in our case. When the sensor is aligned in the middle position, the output fibres ideally receive equal optical power transmitted from the input fibre (Figure 3.4(a)). During lateral left bending (angular misalignment in the x-axis) in Figure 3.4(b), the input fibre is shifted towards the fibre I1 direction which increases the light coupled to I1 and decreases that to I2. In contrast, light coupled to I2 increases and decreases that of I1 for the lateral right bending (Figure 3.4(c)). Using this
intensity modulation method, the bending response for lateral bending is estimated and calculated from the following output ratio equation $R(\theta_x)$:

$$R(\theta_x) = \frac{I_1(x,y,z) - I_2(x,y,z)}{I_1(x,y,z) + I_2(x,y,z)}$$  \tag{3.26}$$

For the sagittal bending measurement (angular misalignment in the $y$-axis), the output intensity ratio from the three output fibres is calculated. During flexion bending to the front, the input fibre shifts upwards as shown in Figure 3.4(d). Light coupled to the fibre $I_3$ increases and decreases the coupling to both fibre $I_1$ and $I_2$. The process is reversed for the case of sagittal extension as illustrated in Figure 3.4(e). The sagittal bending response is calculated from the ratio of three output fibres $R(\theta_y)$ as follows:

$$R(\theta_y) = \frac{\frac{1}{2}I_1(x,y,z) + \frac{1}{2}I_2(x,y,z) - I_3(x,y,z)}{\frac{1}{2}I_1(x,y,z) + \frac{1}{2}I_2(x,y,z) + I_3(x,y,z)}$$  \tag{3.27}$$

In the theoretical treatment in this investigation of the bending sensor, equation 3.21 is used to calculate the fibre transmission loss of each output fibre. The output ratio equation of lateral bending in equation 3.26 can be established as:

$$R(\theta_x) = \frac{(NA^2\beta_{1\text{max}} - \sin^2\frac{\gamma_1}{2}\tan\beta_{1\text{max}}) - (NA^2\beta_{2\text{max}} - \sin^2\frac{\gamma_2}{2}\tan\beta_{2\text{max}})}{(NA^2\beta_{1\text{max}} - \sin^2\frac{\gamma_1}{2}\tan\beta_{1\text{max}}) + (NA^2\beta_{2\text{max}} - \sin^2\frac{\gamma_2}{2}\tan\beta_{2\text{max}})}$$  \tag{3.28}$$

and the sagittal bending ratio in equation 3.27 can be represented as:

$$R(\theta_y) = \frac{(NA^2\beta_{1\text{max}} - \sin^2\frac{\gamma_1}{2}\tan\beta_{1\text{max}}) + (NA^2\beta_{2\text{max}} - \sin^2\frac{\gamma_2}{2}\tan\beta_{2\text{max}})}{(NA^2\beta_{1\text{max}} - \sin^2\frac{\gamma_1}{2}\tan\beta_{1\text{max}}) + (NA^2\beta_{2\text{max}} - \sin^2\frac{\gamma_2}{2}\tan\beta_{2\text{max}})}$$  \tag{3.29}$$

where the limit values $\beta$ and $\gamma$ in both equations are depending on the angular bending and offset coordinate of each output fibre. The ratio varies corresponding to the angular
misalignment in the x-axis for lateral bending and y-axis for sagittal bending. Since all the input and output fibres in the calculation and experiment are identical, it can be shown from a theoretical calculation that both of the power ratio equations are independent of the light source intensity and loss (optical attenuation) of the system. If the type of fibre is fixed with all outputs received from the same source, the sensitivity of the output ratio is only dependent on lateral and angular offset between the input and output fibres. Using the ratio equation to estimate the bending angle, no reference signal is required as all the output fibres automatically provide a self-referenced signal. Noise from the light source or environment is therefore greatly reduced or eliminated considering that all three output fibres are closely bundled together and the input light signal is coupled from a single source. Moreover, the overall size of the sensor can be reduced as the requirement for a separate reference fibre in the sensor has been eliminated.

3.3.2 Respiratory Monitoring

The same working principle is applied for the respiratory monitoring sensor. For this application, the sensor is attached on the upper torso to detect movement of the chest or abdomen during respiration. The input fibre is aligned to the center of three output fibres before any measurement is taken. When the sensor is attached on the upper body, breathing movement affects the position of the input fibre due to the expansion of human’s rib cage. The movement of the human’s chest causes a certain degree of small bending to the sensor. This causes angular misalignment between the input and output fibres. Light coupled to the output fibres varies when the input fibre shifts upwards or downwards as shown in Figure 3.5. Thus, respiration signal is obtained using the output intensity ratio of three output fibres $R(\theta_3)$ as discussed previously in equation 3.27 above.
Figure 3.5: Fibre configuration of body movement during the respiratory activity.

For the respiratory monitoring sensor, only one bending movement is required to detect the peak to peak signal. The output intensity from the sagittal ratio is recorded for the respiration signal since the movement of the chest in this direction provides a larger signal amplitude (ratio) compared to the lateral case. The respiratory POF sensor can also be fabricated utilizing only two output fibres in a vertical position (up and down) with the input fibre aligned to the center point of both output fibres at the resting position. In this case, equation 3.26 can be applied and recorded as the respiratory signal. However, the same sensor can only measure bending movement in one direction and consequently could not be used for monitoring spine bending in the lateral plane.

3.4 Prototype Development and Device Design Specifications

In this section, the design and prototype development process of the POF sensor is presented. The fabricated sensor probe is used to measure the bending angle of the lower back spine and to measure the respiration signal. Hence, the sensor has been designed and modified in such a way as to fit the objectives of the project:
• It is able to provide continuous measurement of human spine bending in both lateral and sagittal planes.

• It is suitable for other health monitoring application such as respiratory monitoring when the sensor is placed at around the lung region.

• The sensor probe is small, flexible and lightweight as to adapt to the various sizes of body. It is low cost and easy to operate so that the sensor can be used in a clinical environment or home monitoring.

• The sensor is portable and can be used in different environments, for example when used in conjunction with MRI or X-ray scanning.

Numerous sensor prototypes have been designed, fabricated and modified to improve the sensor for practical application as shown in Figure 3.6. The first prototype of the sensor in Figure 3.6(a) was fabricated using a readily supplied material such as plastic assembly blocks and PTFE rod as the fibre holder. The input holder tube was drilled with a hole of 1 mm diameter and the output holder tube was drilled with three connected hole (closely grouped together) of 1 mm diameter each. Both fibre blocks were subsequently aligned and connected using a spring at the flexible part of the sensor. The downside of this sensor proved to be the deformation and energy loss of the spring after repeated stretching over time, which requires the reinstallation of new spring at some point. The second modality of the sensor was fabricated with the use of a joint coupling steel at the flexible region that allows the sensor to bend freely in the lateral and sagittal directions. The metal universal joint has a hole of 4 mm diameter in the center of the joint to fit to the PTFE rod that holds a POF. The use of the metal universal joint allows the sensor to bend only in lateral and sagittal directions. The joint is able to hold both tubes in the aligned axes but could cause discomfort to the patient when requested perform various posture. Furthermore the presence of the steel joint in this sensor also rule it out for use in the MRI and X-ray scanning environments. The sensor performance was
enhanced by eliminating the use of spring and steel joint at the flexible point and was replaced with a silicon mould to improve the overall flexibility and robustness of the sensor (refer to Figure 3.6(c)). The use of the silicon mould allows the sensor to be all-plastic in structure after the removal of steel spring and joint.

To build a lighter and less bulky sensor, the sensor has been fabricated in various prototype forms and tested on the back of the human body (being myself and limited volunteers in the laboratory). The selected design that fits the objectives above was generated in a three Dimensional Computer Aided Design (3-D CAD) model (using FreeCAD Software). The design was subsequently 3-D printed to be attached on the patient’s back. The use of 3-D printing to build the sensor probe reduces the overall dimension of the sensor, the complexity of fabrication and facilitates easy attachment onto the various size of patients. Moreover, printing of the sensor allows the accurate alignment between the input and output tubes and reduces the offset error that affects the flexion and the lateral bending response. The body of the sensor is also more durable being highly flexible, strong and robust to hold the optical fibres in position during the measurement.

The final realization of the 3-D printed sensor shown in Figure 3.6(f) removes the electronic part (LED) from the sensors housing locating it to a remotely located launch (into the fibre) position. The input fibre tube with an LED holder in Figure 3.6(e) is modified into a fibre holder to keep the LED separated from the sensor probe. The sensor probe is thus in an all-plastic structure and this enables the sensor to be used in harsh or special environment as previously mentioned e.g. X-ray and MRI scanning.
Figure 3.6: The evolution of the different fabricated prototypes of the sensor to improve flexibility and practicability.

FreeCAD (computer aided design) software is used to draw the 3-D sensor models before printing. The cross-section diagram of the selected sensor design generated as a solid body diagram by the CAD software is depicted in Figure 3.7.
Figure 3.7: Picture shows the horizontal cross-section diagram of the POF sensor’s CAD design.

The device consists of two separated tubes to house the optical fibre: an input fibre tube on the right and the output fibre tube on the left. The design of the sensor using two tubes as casings is important to make sure the fibre itself does not bend during the measurement which would adversely affect the results of bending angle at the sensor’s hinge point. The input fibre tube holds the single input POF that is connected to the LED source at the launching end. The output holder tube consists of three holes each with a 1mm diameter bore. This tube therefore accommodates the three output POFs such that each can be connected to a distal photodiode detector. Both cylindrical tubes have an outer diameter of 6 mm. This allows the minimum wall thickness of 2.5 mm for the input fibre tube (accommodating a 1.0 mm diameter hole) and 1.9 mm for the output fibre tube (accommodating a 2.2 mm diameter hole).

Both housing tubes are integrated in separate rectangular sensor pods, which form the attachment point to the skin. The distance between each tube and pod is designed to be as small as possible (around 1.5 mm) to provide a more accurate representation of the spinal curvature when the sensor is close to the spine region. The flat size of the sensor (smaller height) also allows undetected and comfortable wearing under clothing, without
hindering the actual movement of the joint. The pods of both tubes are separated by a 30 mm displacement. The thickness of the pods was designed to be 2 mm for the base of the sensor to be robust. The base pod of the output fibre tube has a dimension of $25 \times 20 \times 3$ mm while the base of the input tube is $25 \times 15 \times 3$ mm. The base pod of the sensor was designed to be this dimension so that it can be firmly attached to the patient’s skin during the testing but not outsized so that it is comfortable to wear for each individual patient. Both sensor’s bases can simply be attached on the body using only biocompatible adhesive tape.

A small ring bulge of 1 mm radius was added to the design of the input tube as pointed in Figure 3.7. The bulge provides greater tension to the silicon tube by stretching it. This facilitates the silicon tube to fix firmly when connecting both tubes together at the joint. A U-bend tube with centerline diameter of 14.0 mm was added to the design model at the end of the input fibre tube. The U-bend tube has a wall thickness of 1.5 mm with a 1.0 mm hole in the center axis to house the input fibre that is connected to the LED at the transmitting end. The input fibre was bend at 7.0 mm radius when inserted into the printed casing. The macro-bending of the fibre causes some light attenuation. However, the loss was compensated by using a higher power LED source and increased drive current. The details of the LED driver circuit are included later in Section 3.5 of this thesis. During the measurement, the input light coupled is not affected by the bending activities as the input fibre is secured in the prototype casing. The fibre bending designated at this radius produced a smaller sensor size for a better comfortability without significant additional signal power loss and allows the LED and detector (and the associated electronics) to be located in the same enclosure. The design also maintains the all-plastic sensor composition as previously developed.
The structure of fibre hinge and the fibre assembly are both important factors that influence the performance of the sensor. The input fibre tube has been designed with a Philips screw head like pattern comprising four 3 mm height raised ridge sections. The output tube has been designed to be the opposite of the Philips form being a 4 mm recess depth such that the raised cross element easily fits the input tube but not allowing any backlash movement. The design therefore prevents slippage during bending, also prevents axial rotation and hence ensures that the sensor bends only in sagittal and lateral directions. It also keeps both sensor tubes aligned in the center position with a minimum gap of 1mm between the tubes. The four edges of ‘Philips screw head’ was also slightly filleted or chamfered to allow the smooth bending of the sensor when fitted into the output tube during assembly. A clear diagram of the structure of the sensor’s hinge is shown in Figure 3.8. The sensor tube is therefore well aligned, firmly connected and fixed in position using a silicon tube and silicon mold to further secure both tubes together and provide additional mechanical strength to the sensor assembly.

![Figure 3.8 : Structure of the sensor hinge and assembly.](image)

The sensor probe was designed to measure the bending angle of the lumbar spine region in both sagittal and frontal planes. It can also be used for respiratory assessment.
by attaching the sensor probe on the upper body. The device is able to provide real time feedback and recording of data for both the spine and respiratory monitoring.

3.5 Electronic Circuit Design

An electronic circuit board designed for the input light source and output signal detection is discussed in this section. The output signals from photodetectors were filtered and amplified before transmitting to a data acquisition unit. The circuit consists of three parts: an LED driver circuit, a thermistor circuit and three photodetectors conditioning circuit.

The LED is driven from a constant current source as shown in Figure 3.9(a) so that the intensity of the output light is maintained constant throughout the measurement. To drive the LED current, a 3.6 V Zener diode is connected to the base of the transistor to maintain its voltage (Clamped) at a constant value. The emitter current can thus be maintained constant and calculated from the voltage drop across the 160 Ω resistor and accounting for the 0.7 Volt drop of the BE junction of the forward biased transistor. Using this method, the current flowing through the LED is controlled to maintain the current as the desired forward current. Although there are three output fibres to receive the modulated signal, only one LED is used in the sensor so that any error and uncertainty due to the light source drift and environmental effects on it (e.g. temperature, humidity) are eliminated from the output ratio. This also meets the requirement for a simple and low cost design of the sensor system. Using equations 3.26 and 3.27 to estimate the bending angle, the sensor is automatically compensated for source variation and losses in the input and output fibres [133].
A thermistor circuit was added into the design of electronic board to measure its temperature in case of any adverse effect on the photodetector and associated amplifier circuits. Figure 3.9(b) shows the thermistor measurement using a constant voltage source in a voltage divider circuit. A 10 kΩ thermistor is used in this circuit. The voltage output signal, \( V_{\text{out}} \) depends on the thermistor resistance. Temperature is obtained by converting the thermistor resistance to the parameter that is provided in the thermistor datasheet. The temperature obtained during the assessment acts as an indicator only and there is no need to record it throughout the measurement.

The photodetector with amplifier and filter circuit was implemented to amplify the output signal to a clearly measurable level in the software and to improve the signal to noise ratio (SNR) of the signal. Three standard Silicon photodiodes (SFH250V) were used to detect the output signals of the POFs. Figure 3.9(c) shows the circuit design for a non-inverting amplifier and low pass filter (LPF) circuit. When the input fibre coupled the light signal to the three output fibres, each Silicon photodiode produced a photocurrent, \( I_f \) of the incoming optical power. The output photocurrent is converted into the voltage, \( V_{\text{out}} \) using equation:

\[
V_{\text{out}} = I_f \times 5.6k
\]

(3.30)

The converted output voltage, \( V_{\text{out}} \) from each of the photodiode circuits has a small magnitude due to significant signal attenuation loss, i.e. transmission loss, fibre bending loss and loss from fibre gap between the input and output fibres. Therefore, a non-inverting amplifier circuit is required to increase the signal level to a measurable signal using LabVIEW software. The converted output voltage of the photodetector circuit, \( V_{\text{out}} \) was amplified using a non-inverting amplifier circuit. The voltage gain, \( A_v \) was set to be 16 through selection of the appropriate resistor values as calculated in equation 3.31.
below. The output voltage range following amplification should remain within the maximum analog voltage limit of the data acquisition hardware used.

\[
A_v = 1 + \frac{150k\Omega}{10k\Omega} = 16
\]  

(3.31)

In the actual measurement, there exists system noise which is generated in the electrical circuit. An active filter (low pass) was added immediately following the amplifier circuit in order to minimize the noise and hence improve the SNR. In this circuit, a Sallen-Key topology has been implemented to filter the noise introduced in the output signal. A second order low pass filter was applied using active and passive components and has a unity gain output. Adequate gain is provide in this circuit from the non-inverting op-amp based circuit described above. The LPF was designed with both resistors and capacitors having the same value (standard Sallen-Key design protocol). The cut off frequency of the LPF was set to be 1.5 Hz which was adequate to transmit the slowly varying signal frequencies encountered in this investigation. The same filter is also applied on the respiratory monitoring system. The respiratory signal is relatively slow as a normal human respiration rate for adults is under 20 breaths per minute [134] and thus signal details were not missed using the low pass filter with a 1.5 Hz cut off frequency. The LF353 dual JFET input operational amplifier (op-amp) was implemented in the circuit. This single device houses two op-amps in the single IC which were used for both the amplifier and filter, which reduce the overall size of the circuit and require only low supply current of 3.6 mA and hence meeting the low cost requirement discussed earlier as a basic requirement of this sensor system. The designed electronic circuit was fully simulated using P-spice software to access the system performance prior to undertaking the full circuit building process.
Figure 3.9: (a) LED driver circuit, (b) Thermistor circuit and (c) Photodetector circuit for filtering and amplifying signal.

3.6 Data Acquisition Hardware and Software

The data capture hardware used in the system (i.e. Data acquisition hardware device) is described. LabView™ software programming was implemented for the in-house developed software of this investigation being fully compatible with the DAQ to capture and record the signal output of the sensor hardware. A fully portable and ultra-low cost sensor system using Arduino platform is also described in this section.
3.6.1 Data Acquisition Hardware and LabVIEW

The DAQ card used with the PC of this investigation was the national instruments DAQ (data acquisition) NI USB-6008 unit to provide both data capture and analog to digital conversion. The DAQ unit consists of 8 single-ended analog input (AI) channels, two analog output (AO) and 12 channels that can be configured individually to digital input (DI) or digital output (DO). The analog to digital converter (ADC) resolution of the sensor is thus 12 bits for differential input and 11 bits for single-ended input, which corresponds to 4.88mV per step for a 10V range. The maximum sample rate for single channel is 10kS/s which far exceeds the required sampling rate in this investigation being at most 5 sample per second based on a LPF cut off of 1.5 Hz. The data acquired was analyzed using an in-house programme developed using the LabVIEW™ software suite.

Each output voltage was acquired from the DAQ and the compensated ratio signal was processed and displayed using the LabVIEW™ software. A Laboratory Virtual Instrument Engineering Workbench (LabVIEW) also called virtual instruments (VIs), is a software programming language similar to MATLAB (Mathworks, Inc.) but is implemented entirely in a graphical programming language. It consists of two windows, a front panel and a block diagram. The front panel functions as a user interface where the control buttons and indicator graphs are displayed in the window. Graphical code for data acquired and processing is kept in the block diagram window. In this project, LabVIEW was used to study the temperature reading from thermistor, to read the voltage from three output fibres and to calculate the signal’s output ratio. The LabVIEW code and user platform to record the sensor’s signal will be elaborated in Section 4.3.1 of the next chapter.
3.6.2 Arduino Platform

The sensor system has been made fully portable by utilizing a low-cost Arduino platform as the hardware to read the output signal from the photodetector circuit. This sensor system is intended for circumstances where the sensor is needed to be operated in a fully remote environment and in an independent manner e.g. outdoors or whenever its use is required without a computer e.g. as a ‘hand-held’ clinical monitor. An Arduino compatible Mega 2560 Microcontroller board along with a 3.2 inch TFT LCD touch screen was connected to the circuit board to capture the data. The board using an ATmega 2560 processor consists of 54 digital input or output pins and 16 analog inputs. The 16 analog to digital converter (ADC) channels allows data capture from the sensor, each provides 10 bits of resolution. It measures the input voltage in the range of 5 Volts to ground by default, which yields the resolution of 4.88mV per step (1024 different value). This is important to obtain accurate results in measuring the voltammetry test as the resolution should be as small as possible to present the closest results of an analog state. Each I/O pin provides or receives a maximum current of 20mA. This microcontroller is chosen due to its high reliability, versatility (in programming) and low cost. It can also be powered using an external power source (ie. 5 Volts USB power pack or other regulated power supply) to the pin $V_{\text{IN}}$, which allows the device to be portable.

3.6.3 MATLAB Programme

MATLAB (Mathworks, Inc.), short for MATrix LABoratory is the programming language that allows user to plot functions and data, developed and implement algorithms, data analysis, create user interface and interface with programs written in other language. A typical MATLAB programming interface is composed of several windows. The directory window allows instant access of files in the Matlab working directory or path.
The command window allows running of Matlab commands in a script or function for execution. The workspace window allows management or to view the current values, status or names of all variables currently in use. In this project, the MATLAB programming language was used to calculate and compare the theoretical calculation results of the sensor response with the experimental data. The software was also utilized to process the signal captured from both spine bending and respiratory monitoring, i.e. signal smoothing, signal normalization and data comparison and quantifying.

3.7 Summary

This chapter included a theoretical analysis of the sensor in this investigation that works based on coupling loss between input and output fibres with lateral and angular misalignment. The sensor configuration and working principle using three output fibres that centrally align to an input fibre has been described in detail. The prototype of the sensor development and design has been studied to determine a suitable sensor probe to be 3-D printed for spine and respiratory monitoring. Electronic circuit design for the light source (LED) and photodetector conditioning circuit have also been described in detail. The data acquisition hardware and software were described in terms of their means of implementation outlined via the DAQ, LabVIEW software, Arduino platform and Matlab. Further details of the programmes etc. are given in the appendices of this thesis owing to space limitations in the main section.
4.1 **Introduction**

In this chapter, sensor realization including the fabrication process and assembly of the fully integrated sensor is described. The sensor probe design that was described in the previous chapter was 3-D printed and fitted with four POFs. The successful assembly of the sensor probe and the system is demonstrated. The complete sensor system was calibrated using an optical setup comprising a precise rotational stage before testing on human subjects. Experimental sensor bending results from the calibration process were compared with the theoretical calculations of the previous chapter.

In this investigation, the same POF sensor is used in the spine bending assessment as well as the respiratory monitoring assessment. It was necessary to test the sensor on a limited set of volunteers to investigate the viability and practicality of the proposed sensor. This chapter is confined to defining the methodologies and protocols adopted to the various experimental calibration and testing procedures. Therefore no results are presented in this chapter, but these are fully described in Chapter 5 and Chapter 6 of this thesis for the spine monitoring and respiratory monitoring assessment respectively.

4.2 **Sensor Printing and Fabrication**

A Stratasys Connex 500 printer located at the department of Design and Manufacturing Technology at University of Limerick was used to implement the 3-D printing of the designed sensor probe. The printer is capable of printing multiple materials with a high spatial resolution of 32 µm. The sensor probe was printed using a rigid and opaque polymer material named ‘Vero white’ which prints strong and durable prototypes. The printing material was mixed with a small amount of rubber-like material, ‘Tango
black’ during the printing to increase resilience and robustness. The final prototype as discussed in the previous chapter was 3-D printed and the outcome is illustrated photographically in Figure 4.1. The supporting material inside the gap and fibre hole was removed and cleaned after the sensor probe was printed.

![Figure 4.1: Photograph of the 3-D printed sensor probe assembly.](image)

Four multimode step index Polymethyl-Methacrylate (PMMA) fibres (one input and three outputs) were included in the sensor fabrication. Large core multimode fibres (MMF) have been implemented in the system as the intensity modulated sensor required a large light signal for the best signal to noise performance [135]. The fibre chosen for use in this study is manufactured by Asahi KASEI (product number TCU-1000(L)). The PMMA fibre has an outer diameter of 1.00 mm with a core diameter of 0.98 mm and is protected with a jacket of 1.00 mm thickness. The refractive indices of the core and cladding are 1.492 and 1.402, respectively. The MMF has a NA of 0.51 and acceptance angle of 30°.

To complete the sensor probe fabrication, four POFs of the above specifications were cut into lengths of two meters each (length from the sensor probe to the light receiving circuit). Both sides of the fibre end were hand polished using a polishing puck and film for the best light transmission. The protective rubber jacket of the fibre was
removed at one end (to match the length of the tube) before being inserted into the printed casings. One of the fibres was inserted into the input fibre tube and was fixed using standard two-part epoxy at the end of the U-bend to ensure that the fibre is fixed in the casing. Three polished fibres were attached together using the same epoxy and inserted into the output tube. Subsequently, the bottom end of the output casing was fixed using the same epoxy to the three output fibres to ensure that the fibres did not move or creep when the sensor is bent. Next, opposite end of both input and output fibres were connected to the LED and photodetector mount that is located at the circuit board.

In this sensor application, a plastic fibre optic transmitter diode SFH756V (supplied by Broadcom Ltd, US) was used as the light source. The bright red light-emitting diode (LED) has a peak wavelength of 660 nm. The LED package features a connector with a 2.2mm opening to tightly secure the POF fibre end, and thus ensure no light leakage during the measurement. As LEDs tend to distribute their power over a large area and are not a coherent light source, it can be safely used with the relatively high power without posing danger to human eyesight in the case of exposure. Hence, even though the light source output is in the visible part of the spectrum, they do not pose any risk for eye safety.

The LED was connected to one end of the transmitting POF that was aligned to the centre axis of three POFs at the other end. The light signals that coupled to the three POFs were detected using Silicon photodiodes SFH250V (supplied by Broadcom Ltd, US). The input tube was embedded and optically aligned to the center of the output fibre tube after all the fibres were secured into both printed casings. The output voltage signals from each of the three output fibres were read during the alignment process to ensure that the input tube was in the correct aligned center point of the three output fibres as discussed in section 3.3.1. Subsequently, both tubes were locked together in this position using a rubber silicon tube of 1 mm thickness that allows the flexible movement and some
stretching space at the hinge during bending. The sensor hinge is coated with another layer of silicon mould gel that solidifies when left to dry for a few hours. The purpose of the silicon mould is to improve the robustness of flexible region and it assists in holding both tubes together for added strength and mechanical robustness of the complete sensor unit. A photograph of the POF sensor probe that is ready to be attached directly on the patient is shown in Figure 4.2.

![Figure 4.2: Photograph of the sensor probe after assembly.](image)

4.3 **Sensor Assembly**

The electronic circuit described in the previous chapter was constructed on a breadboard and soldered on a Veroboard for preliminary testing purposes. A printed circuit board (PCB) layout as shown in Figure 4.3(a) was designed using PCB-POOL Target 3001! Software. A 10 cm × 13.5 cm dimension of PCB was created to mount the circuitry in the final enclosure. The electronic circuit was powered using 12V and 5V power supplies. The board is arranged with the power supply terminal at the top and three photodetectors at the bottom. Five output connection terminals are located at the right of the PCB which provide the connection to the output signal of three fibres, thermistor voltage and a ground to the DAQ or Arduino platform. A completed PCB with LED and photodetector circuit components is shown in Figure 4.3(b).
4.3.1 Sensor probe and DAQ system

The output signals from the photodetectors and thermistor circuit on the PCB were connected to the DAQ channel for analog to digital conversion. The DAQ analog input terminals AI0, AI1, AI2 and AI3 were connected for the measurement from the three photodetectors and one thermistor. The channels share a common ground and the signal range with reference single ended coupling (for the DAQ) is limited to 10V. The PCB was mounted on a board using four insulating spacers with the DAQ placed at the bottom layer for a better portability. The DAQ was connected to a PC and the captured data was...
analyzed and displayed using the LabVIEW program designed and implemented as part of this study.

The LabVIEW program was developed on the basis of a block diagram structure which was divided into three segments to form a flat sequence structure described as follows:-

The first part receives the voltage measured from the three output fibres (labeled as fibre A, fibre B and fibre C in the LABVIEW code) as well as the voltage from the thermistor circuit and all were obtained from the DAQ output. Four DAQ ‘create channel’ programs were used to continuously read the signal from each output channel. The laboratory sampling rate \( (f_s) \) was set to be 10 Hz and the amount of samples read per channel per second was set to be 10. This provides 10 data points per channel for an acquisition time of one second.

The next segment is where each output signal was clustered into a data arrays. The compensated output ratio signals, i.e. \( R(\theta_x) \) and \( R(\theta_y) \) in equations 3.26 and 3.27 were then calculated using dataset arrays (for each output fibre) to record the sagittal and the lateral bending outputs. For the lateral equation, output from fibre I2 is subtracted by fibre I1 and the result is divided by the summation of both outputs as in equation 3.26. For the sagittal output ratio, the numerator value of equation 3.27 is first calculated using the summation of fibre I1 and I2 and subtracts double of output fibre value, I3. The numerator is then divided by the denominator calculated from the sum of fibre I1, I2 and double of I3. The output voltage obtained from the thermistor was converted into degree Celsius in this section. This was done by substituting the equation and variable from the data sheet into the ‘convert thermistor’ reading block diagram.

All graphs and display block diagrams were implemented in the last segment of the sequence structure.
The front panel consists of a display of the time varying output voltage signal in the display graph to be viewed and controlled through the graphical user interface. The right of the front panel consists of three graphs that are arranged in a single display table as shown in Figure 4.4.

![Figure 4.4: Front panel window (output voltage and ratio response).](image)

The ratio chart for lateral bending is located on the upper left of the display table. The ratio chart for sagittal bending is located on the upper right of the table and output voltage of three fibres are placed at the bottom positions. Data was recorded and saved into the excel file to be analyzed later. The left hand section of the front panel window (top and bottom tab) is shown in Figure 4.5.
Figure 4.5: Front panel window (control tab and temperature display table).

On the bottom of the display panel (Figure 4.5), a table shows the numeric indicator of the temperature data measured from the thermistor voltage after conversion into degree Celsius. The front panel also contains the tab control table with control buttons to adjust the sampling rate, number of samples or for changing the DAQ channel that connects to the output pin. The overall block diagram and front panel for the recording of the sensor’s data is listed in Appendix A: Labview Block Diagram and Front Panel.

The schematic diagram of the sensor layout including source and detector is shown in Figure 4.6. The DAQ hardware was connected to the PC and the PCB was powered using a benchtop power supply. The fibre tail of the sensor probe was connected to the LED source and photodetector with a separation between the sensor probe and the LED and detection unit of 2 m. The LabVIEW program was run to read and capture the data. A photograph of complete sensor system after assembly is illustrated in Figure 4.7.
4.3.2 Portable Sensor System using the Arduino Platform

A new PCB of 11.5cm x 5.5cm dimension was fabricated for the portable device using an Arduino based platform. The Arduino board can only read a maximum of 5V input from the ADC channel, and therefore the PCB design needed to be slightly modified to fit the platform. Firstly, the value of the resistor applied in the non-inverting amplifier
circuit in Figure 3.9(c) was changed to a smaller resistor value of 68 kΩ. This is to decrease the overall gain so that the voltage of each output fibre after amplification is within the range 0 to 5V. The op-amp model used in this circuit for the amplifier and filter was the LF358 low-power dual-operational amplifier which can operates using a single power supply over a wide range of voltages. The Arduino Mega 2560 and a LCD touch screen were used as described in the previous chapter. The three output signals from the output of the photo-amplifiers were connected to the analog pin A13, A14 and A15 of the Arduino board. The Arduino Mega was mounted on top of the PCB using four self-tapping standoff screws. The LCD touch screen was mounted on top of the Arduino Mega board. Both the printed circuit board and Arduino were powered using a single 5V regular power supply or portable power bank. A plastic box with a cap of dimension 7 cm x 12.8 cm x 5 cm was designed and 3-D printed to house and protect the sensor circuit. The various layers of the sensor system together with the 3-D printed box is shown in Figure 4.8. The whole sensor is thus highly compact, small, lightweight and portable fit to be carried and used anywhere.
The Arduino Mega was programmed using proprietary Arduino software. The output signal from the photodetector circuit that was connected to the Analog channel was read using the analogRead() function. The lateral and sagittal output ratio of the signals (in equation 3.26 and 3.27) were calculated in the software and directly displayed on the LCD screen. The signal is displayed on the screen with a circle dot using the DrawPoints() function. Each column of the screen is updated with the new data at each loop using the increment (++) function. The signal of the sensor is thus displayed in a time-domain graph by multiplying ratio value to the height of the screen in pixel. As the maximum and minimum output ratio is in the range between -1 to 1, the output value of the sensor is offset to be in the middle axis of the screen with the lateral and sagittal ratio signal at a 30 pixel distance apart to avoid both signals being stacked together at the value of zero. The yellow colour drawpoint represents the lateral ratio and the green drawpoint represents the sagittal ratio.
Figure 4.9 demonstrates the sensor system with an LCD touch screen using the Arduino platform. Several buttons in a rectangular box shape were added to the top of the LCD screen to change the display signal when pressure is applied to the screen. The first blue ‘bend’ button displays both the lateral and sagittal ratio signal as discussed above. This is used during the bending spine assessment to allow monitoring of both bending signals simultaneously. The grey ‘RESP’ button represents signal obtained during the respiratory monitoring assessment. When the grey ‘RESP’ button is pressed, only the sagittal ratio signal in green is displayed on the screen. The ‘Save’ button allows the raw data of the output ratio to be written and stored to the comma separated value format (*.csv) in a SD card to be read from PC later if required. Once the ‘save’ button is pressed, a file in csv format is open in the SD card to enable write access to the file. The green ‘start’ button next to the ‘save’ button initiates the writing of the lateral and sagittal values into the file until the same button is pressed for the second time. The complete coding for the Arduino software to read, display and save the data is included in Appendix B: Programming Code for Arduino platform. Data collected can be stored in the SD card inside the Arduino platform, connected to a PC using the USB cable or the signal transmitted wirelessly using a Bluetooth module.

Figure 4.9 : Portable electronic circuit with the Arduino board and LCD touchscreen enclosed in a 3-D printed box.
Sensor Angle Calibration

The POF sensor was calibrated after the fibres were assembled inside the 3-D printed sensor housing structure and connected to the electronic system. The purpose of this calibration process was to determine the operating limits of the sensor system and obtain a sensor response relationship between the output ratio and the bending angle. In the ideal case of the fabricated sensor, the output voltage of three fibres should receive ideally equal amount of light intensity when the sensor is at the zero aligned position. However, the voltage measured from the output of the photoamplifiers of the three output fibres may not be equal in the practical case. This problem can be resolved by calibrating the sensor for the first time before use to investigate the relationship between the bending angle and the output ratio of the sensor.

An optical setup system as depicted in Figure 4.10 was built especially for the calibration purpose. The setup consists of two translational stages, two optical posts, two holders with clips and a precise rotational stage. Figure 4.10(a) illustrates the configuration for the lateral bending measurement. The input sensor’s base (side A in the Figure 4.10(a)) was fixed horizontally to the translational stage with two optical posts that were also mounted on the stage. Each optical post consists of a clip to secure the sensor’s base on its left and right sides. The output sensor’s base (side B) was secured onto the rotational stage of ± 0.5° accuracy. The pivot axis of the sensor was aligned to the center of rotation axis. The rotational stage was mounted on another translational stage. This is to allocate some movement freedom to ease both the alignment process and setup of the sensor.
Figure 4.10: Image and schematic diagram of the experiment setup for sensor characterization on (a) lateral bending and (b) sagittal bending.

For the sagittal bending calibration, the POF sensor was mounted vertically on the optical setup as shown in Figure 4.10(b). Two clips were vertically placed on a single optical post to secure the input sensor’s base. A home-made wooden block was fabricated and attached on the rotational stage using a screw. The wood block was used to hold the output sensor’s base in the vertical direction. The translational stage on both sides were utilized to align the sensor to the center position after the sensor was installed on the setup.

For both configurations, the output fibre’s intensity was recorded and the output ratio was calculated (in the LabVIEW™ programme described in Section 4.3.1) in near real-time whilst rotating the stage. The stage was rotated with an increment angle of 2° every 30 seconds starting from 0°. The rotation continues until reaching the saturation point at which no further changes in the output ratio in response to the angle change could be observed. At this point the rotation angle was slowly decreased with a 2° interval until
returned to its original 0° position. A finer calibration was also implemented by
decreasing the angle interval to 0.5° using a fine tuning adjustment on the rotational stage
up to the range of ±4°. The steps were repeated in the opposite rotation direction until the
saturation point was reached. Each calibration process was repeated three times to assess
repeatability and hysteresis. The fabricated POF sensor is calibrated using the same
method before testing on a human subject. The sensor’s bending response from this
calibration was compared with the theoretical analysis results. Results from both
experimental and theoretical were plotted using Matlab software and are discussed in the
next chapter.

4.5 Lower Back Spine Monitoring Assessment

The fabricated sensor in this investigation aims to aid and allow clinical staff to
accurately identify the underlying cause of low back pain and helps in monitoring the
ongoing rehabilitation progress of the patients from spinal injury. This is done by
monitoring the lumbar spine of the patients undergoing simple standard movement tests
in the clinic. The device is to be placed at the lumbar spine region, which is the most
commonly reported pain area among patients with LBP [136].

The testing and evaluation of the POF sensor on the human spine is discussed in
this section. The sensor was first tested on two subjects to check the initial response of
the sensor when placed on human subject and undergoing some simple motion exercises.
The bending angle at the spine region was measured when the patient was requested to
present different postures or positions. Images of the bending assessment were recorded
and the result from the sensor was compared with the image acquisition. Subsequently,
the sensor was evaluated in a more comprehensive trial involving 18 patients. Results
from the sensor were simultaneously compared with the commercial spine monitoring
device - Biometrics goniometer. The protocols describing these tests are described in the following section and the results obtained are reported in Chapter 5 of this thesis.

4.5.1 Preliminary Test of the POF sensor on the Lower Back of the Participants

For the preliminary test, the POF sensor was placed at the specific lumbar spine segment as illustrated in Figure 4.11. In this test, the sensor was placed at the lower spine region around L3-L4 section. Both upper lumbar spine (from T12-L3) and lower lumbar spine (L3-S1) are known to function independently. Nonetheless, the lower lumbar spine was selected for this measurement on advice from our partner Physiotherapists in UL and due to it being reported as the most symptomatic region in LBP [137, 138]. The bending angle of this spine region was measured when the subject performed different posture or position.

Figure 4.11: The POF sensor attached on the lumbar spine region of human subject.

4.5.1.1 The Protocol for the Clinical Bending Exercises and Measurement on the Lower Back of the Participants

To improve the accuracy of the measurements, there are several precautions that need to be taken into account before the measurement process. First, a standard surgical alcohol wipe is applied on the back of the body for cleaning. Next, the location of the
lumbar spine segment is identified using a manual palpation technique [139] before mounting the sensor. This is a common procedure in manual therapy examinations to determine the locations of particular spine segments ‘by hand’. The identified L3-L4 lumbar spine segment was marked and the sensor was mounted directly using biomedical double sided-tape at the sensor’s base. To avoid the dislocation of the sensor during testing, it may be necessary to remove excessive hair around the mounting area if present.

Two healthy subjects (a female and a male) without back injury and spinal deformity were recruited to participate in the study. Ethical approval for this study was granted by Faculty of Science and Engineering Ethics Committee, University of Limerick and the information of the application and consent letter is included in Appendix D of this thesis. The test was carried out to investigate the capability of the POF sensor in measuring spine bending movement in both sagittal and lateral planes. For the sagittal plane assessment, the subject assumed several static postures as listed below and the position was maintained for 15 seconds at each trial;

- Upright standing with feet shoulder-width apart
- Forward Bending (Figure 4.12 (a)) in small bending degree
- Forward Bending in a larger bending degree
- Return to the standing position
- Extension (Figure 4.12 (b)) to the back
- Return to the standing position
For the lateral plane assessment, the subject assumed the several static posture as listed below and the position maintained for 15 seconds at each trial;

- Upright standing with feet at shoulder-width apart
- Full bilateral trunk left side flexion (Figure 4.13(a))
- Return to the standing position
- Full bilateral trunk side right flexion (Figure 4.13(b))
- Return to the standing position

The entire assessment was repeated and the data was measured and recorded at each trial. The spine movement signal was compared with the image acquisition method.
4.5.1.2 Image Acquisition

The POF sensor’s output signal was correlated to a visual observation of the lower back curvature. An ad-hoc image acquisition scheme was applied to study the relationship between the appearance of back curvature and the angle obtained from the sensor. To examine both lateral and sagittal bending situations, three green marks were placed at different positions at the body and sensor to assist the visualization of spinous processes on the lower back during the measurement session. A standard digital imaging webcam was used to record the data, in order to provide validation of the POF sensor measurement and efficacy of the data recorded using the POF sensor of this investigation. The camera was positioned to the side and back of the participant’s thoracic to lumbo-pelvic region to capture the sensor’s motion for sagittal and lateral bending. This is the standard practice in physiotherapy clinics and thus quantifiable data is extracted from the recorded examination process for different posture by tracing the location of the three spots.

In the case of the sagittal bending assessment, it is very challenging to acquire the exact bending angle where the sensor is attached through imaging due to the curvature of spine. Hence, the measured data captured by the sensor is correlated with the overall inclination of lower back bending and spine curvature of each individual. Two green spots were located one either side of the sensor’s base pods while another spot was located on the middle of the back (for our participants this was about 2cm above the tip of the sensor). Figure 4.14(a) depicts the locations of the three green spots placed on the body and the sensor’s pods.
Figure 4.14: Placement of three green reflective markers on the sensor and body in relation to the sensor module. The posture change was defined as the inclination change at (a) angle $\theta_1$ and $\theta_2$ for sagittal bending and (b) angle $\theta_1$ for lateral bending.

The left side of Figure 4.14(a) shows two angles, $\theta_1$ and $\theta_2$. The angle $\theta_1$ for sagittal bending refers to the overall bending of the back from a vertical line. It is measured between a vertical line passing through the point $P_3$ ($P_3X$) to the line connecting points $P_1$ and $P_3$ ($P_1P_3$). The locations of the green spots were obtained from the pixel of the image captured. The x and y coordinates of the green spot’s center point were initially recorded for every change of posture. Angle $\theta_1$ was calculated using the tangent function of a right angle triangle assuming a triangle connecting points $P_3$, $P_1$ and a point where a horizontal line from $P_1$ crossed the vertical line $P_3X$. Hence, angle $\theta_1$ can be calculated from equation below:

$$\theta_1 = \tan^{-1}\left(\frac{x_1-x_3}{y_1-y_3}\right)$$ \hspace{1cm} (4.1)

where coordinate of $P_1=(x_1,y_1)$, $P_2=(x_2,y_2)$ and $P_3=(x_3,y_3)$

Angle $\theta_2$ in the sagittal assessment represents the localized change of the lower spine curvature at different postures. This is the angle between the two lines $P_2P_3$ and $P_2P_1$. The curvature of the lumber spine arc varies between individuals. The angle $\theta_2$ is calculated
from the angle between the vector $\mathbf{P}_2\mathbf{P}_3$ and vector $\mathbf{P}_2\mathbf{P}_1$. The angle $\theta_2$ can be calculated using the equation below:

$$\theta_2 = \cos^{-1}\left(\frac{\hat{a} \cdot \hat{b}}{|\hat{a}| |\hat{b}|}\right)$$  (4.2)

where $\mathbf{a}$ is the vector $\mathbf{P}_2\mathbf{P}_3$ and $\mathbf{b}$ represents the vector $\mathbf{P}_2\mathbf{P}_1$.

Figure 4.14(b) shows the three reflective markers placed on the sensor during the lateral bending assessment. Two green spots were placed on each side of the upper sensor and one at the middle end of the lower sensor. The overall lateral bending angle $\theta_1$ was measured between a vertical line passing through $\mathbf{P}_1 (\mathbf{P}_1X)$ and a line connecting $\mathbf{P}_1$ to the middle of $\mathbf{P}_2\mathbf{P}_3$. This angle was calculated using the tangent function of a right angle triangle. Coordinates of the green spots’ center axis were obtained and the middle point $\mathbf{P}_m$ between line $\mathbf{P}_2\mathbf{P}_3$ was calculated. The angle $\theta_1$ can be calculated using the following equations:

$$(x_m, y_m) = \left(\frac{x_2+x_3}{2}, \frac{y_2+y_3}{2}\right)$$  (4.3)

$$\theta_1 = \tan^{-1}\left(\frac{x_m-x_1}{y_m-y_1}\right)$$  (4.4)

where coordinate of point $\mathbf{P}_1=(x_1,y_1)$, $\mathbf{P}_2=(x_2,y_2)$, $\mathbf{P}_3=(x_3,y_3)$ and $\mathbf{P}_m=(x_m,y_m)$

For lateral bending, video footage was captured during the whole bending assessment to compare the trend of bending angle for both sensor and imaging. In both bending examination processes, a web camera with resolution 1280 x 720 was used to record the image.

### 4.5.2 Validation of the POF sensor

The preliminary test on human subjects demonstrates the capability of the POF sensor to monitor the lumbar spine angle when the subjects undergo some typical simple motion exercises. The POF sensor was evaluated in a more comprehensive trial on a
larger group of subjects as part of a collaborative study in conjunction with physiotherapists based in UL. The sensor was directly compared with a commercial spine monitoring sensor - Biometrics Ltd goniometer for a series of spine bending assessments. The instrumentation, preparation and experiment protocol for the sensor validation is fully discussed in this section.

4.5.2.1 Biometrics Ltd Goniometer

A Biometrics DataLink Data Acquisition System along with a twin axis SG150/B goniometer (Biometrics Ltd., Newport, UK) [59] shown photographically in Figure 4.15 was used in this validation test. The goniometer was mounted on the lumbar spine region to measure the spine angle when the subject assumed several diagnostic test postures. The strain gauge based electro goniometer has a measurement accuracy of $\pm 2^\circ$ over the range of $\pm 90^\circ$. The goniometer was calibrated according to the manufacturer’s instructions before being mounted on the participants. The goniometer measures the angle between two plastic end blocks that were connected using a spring. The spring comprises of thin resistor wires, flexible steel wires and a series of strain gauge. When there is a change of angle between the blocks, deformation of strain gauge will affect the current passing through the resistor wires. Changes in strain along the length of the wire is measured and converted into angle. The design of the goniometer is only to measure the angular displacement between the endblocks. Hence, rotation of one endblock relative to another cannot be measured.
Figure 4.15: (a) Biometrics DataLink Data Acquisition System and (b) Biometrics goniometer SG 150/B for spine angle measurement.

Figure 4.15(b) shows the dimensions of the Biometrics goniometer SG 150/B used in the validation test. One proximal endblock of size 54 x 18 mm was attached to the lower lumbar spine of body near the sacral region. Another endblock of size 145 x 18 mm was attached to the middle back of the spine at segment T12 to L1, depending on the height of subject. There are two separate output channels that are connected to the goniometer, one measures the relative change of lumbar spine in sagittal bending (flexion/extension) and another measures lateral flexion to left or right directions. The goniometer was connected to a computer via the data acquisition (subject) unit DataLINK DLK900. The subject unit has 8 analogue channels and 5 digital channels to collect data and to convert the sensor’s input to digital signal. The subject unit was connected to a Base unit and to a PC using a USB (universal serial bus) port. The ADC converter is 13 bit that provides a ±4000 counts resolution for the goniometer. The measured data was converted into angle value in degrees using the formula \((x \times 180)/4000\) where \(x\) is the signal data from the goniometer. The sampling rate of the goniometer is set to be the same
as the POF sensor which is 10 Hz, where there are 10 samples read per channel for an acquisition time of one second. The goniometer was calibrated to the zero value after the sensor was attached onto a subject standing in an upright posture.

4.5.2.2 Patient Assessment Protocol for Evaluation

A larger group of subjects were also tested with the POF sensor and the Biometrics goniometer to investigate the correlation and agreement between both sensors in measuring lumbar spine bending. Eighteen healthy painless adults (12 females and 6 males) were recruited from within a university community to complete the study. Participants had a mean age of $30.4 \pm 10.8$ years, height of $167.7 \pm 7.7$ cm, mass of $69.9 \pm 11.3$ kg and body mass index (BMI) of $24.56 \pm 3.17$ kg/m$^2$. The selection criteria for this testing was adults between 18 to 65 years old with no history of lower back pain in the previous six months. Subjects are excluded if they were pregnant, had any history of spine surgery or any tumors of limbs that may worsen by undergoing test procedures. All subjects were given instruction about the sensor testing and practice before recording the data. Ethical approval was granted by Faculty of Science and Engineering Ethics Committee, University of Limerick under reference code Ref 2016_12_01 and this is included in Appendix D of this thesis.

Both the POF sensor and Biometrics goniometer were attached on the lumbar spine section of subject using biomedical double-sided tape as illustrated in Figure 4.16. The POF sensor was placed in the same position as was the case in the preliminary test at the lower lumbar spine region identified through the manual palpation method. The Biometrics sensor was placed side by side to the POF sensor to allow simultaneous comparisons. Data from both sensors were recorded at the same time when the subject performed different bending posture.
During the bending assessment, it was difficult to ensure each volunteer had bent to pre-allocated bending angles due to individual movement variation. Hence, a manual inclinometer (Figure 4.17) was used as a reference guide for specific static posture when the subject bent. The manual inclinometer was placed on the back of the patient between both sensors to study the overall inclination angle of the subject as a posture reference.

The use of the inclinometer in this test does not represent the actual bending angle of the lumbar spine, but merely acts as a reference guide for the recorded posture. For the
sagittal plane assessment, the subject assumed the following static postures and the position was maintained for 10 seconds each:

- Upright standing posture with feet at shoulder-width apart
- Forward bending until the inclinometer reads a change of 5° from the initial posture
- Return to the standing posture
- Repeat the forward bending process for another three forward bending steps, for which the inclinometer reads a change of 10°, 15° and 20° from the initial posture
- Return to the standing posture
- Extension bending (backward) until the inclinometer measures -10° changes
- Return to the standing posture
- Extension bending until the inclinometer measures -20° changes (or until uncomfortable)

For the lateral plane assessment, subjects assumed the following static posture and the position was maintained for approximately 10 seconds each:

- Upright standing posture with feet at shoulder-width apart
- Lateral trunk left side flexion at two different angles 10° and 20° (inclinometer)
- Return to the standing posture
- Lateral trunk right side flexion at two different angles 10° and 20° (inclinometer)
- Return to the standing posture

During both assessments, the upright standing posture was taken at zero reference position, at which point both the POF sensor and Biometrics goniometer signals were zeroed in this posture. Throughout the trial and in all measurement cases it was not necessary to detach any of the sensors as both sensors were capable of monitoring spine bending in two directions. At the end of the study, the comfortability of using the sensor was evaluated by each subject. The subject was requested to rate the comfortability from 0-5 when using the sensor where 0 represents no discomfort and 5 is extreme discomfort.
Data recorded from both sensors were saved in a .csv file and processed using Matlab™ software. Evaluation data from all subjects were further analyzed using IBM SPSS (Statistical Package for the Social Sciences) Statistical software, version 24.0 for windows. The results of this trial are discussed in the next chapter.

4.6 Respiration Monitoring Assessment

In the respiratory monitoring assessment, the same POF based sensor was attached on the upper body and was utilized to monitor the breathing signal of patients. Figure 4.18 depicts the whole setup configuration to detect breathing signals. For the respiratory assessment, the POF sensor sampling rate was set to be 10 Hz, which is considered sufficient to accurately measure the respiration frequency. In the preliminary test, the sensor was attached on the upper costal (chest) as illustrated in the diagram to check the initial response and study the feasibility of the sensor for being used as a respiratory monitoring sensor when placed on a human subject. Throughout the ventilation activities, the diaphragm moves upwards and downwards to change the pressure in the lungs. Simultaneously, the ribcage moves upwards and downwards. The sensor detects the movement of the chest wall during respiration activities to obtain the respiration signal.
4.6.1 Comparison of Sensing Positions

In the next study, the sensor was attached at different positions on the human body in order to investigate the versatility and capability of the sensor to be used at different sensing positions. The participant was asked to sit and relax while breathing naturally (no patient exertion) for one minute while the sensor was attached at four different positions separately. The sensor was attached at three different positions to the front of the body as illustrated in Figure 4.19: upper costal (chest), lower costal (around the diaphragmatic position), middle belly region and one on the back (below the left shoulder blade).
Figure 4.19: Positions of the sensor placed on the subject. (1) Upper costal; (2) Lower costal; (3) Middle abdomen (belly); and on the back of body (4) below shoulder blade.

The overall performance of the sensor placed at different positions was further evaluated by attaching three sensors simultaneously on different positions. In this investigation, different types of breathing signal were simultaneously recorded in order to evaluate the efficacy of the breathing signal recorded at different positions of the body. Three sensor probes and electronic devices in Figure 4.20 were fabricated for this test. Sensor 1 and sensor 2 utilized NI data acquisition system and LabVIEW software while sensor 3 employed an Arduino platform to read and record the signal. Three sensor probes were placed on the upper chest, middle belly and on the back of the body in the position 1, position 3 and position 4 as indicated in Figure 4.19.

This test was also designed to evaluate the breathing signal from three breathing patterns: normal breathing; deep breathing (diaphragmatic breathing) and breath holding. During normal breathing, the subject relaxed and breathed normally during which there should be no clear predominance of thoracic, abdominal or costal expansion. For deep breathing or diaphragmatic breathing, usually abdominal and lateral costal expand more to allow additional gas exchange in the lungs [140]. The subject was requested to inhale a maximum possible amount of air. For breath holding, the subject was requested to stop
breathing for several seconds for which there should be no or minimum movement from chest and abdomen.

The experiment was repeated three times, with each time switching the attachment points of the sensor as listed in Table 4.1. The sitting subject was instructed to breath according to five sections as follows:-

- First, the subject breathed normally for around one minute.
- Next, the subject was requested to hold their breath for a few seconds (or until uncomfortable).
- These were followed by a natural breathing pattern for one minute.
- Next, the subject was requested to hold their breath for a few seconds (or until uncomfortable) for the second time.
- The last section ended with the subject deep breathing (diaphragmatic breath) activity for one minute.

Data from the three sensors were recorded simultaneously for direct comparison. The consistency of repeated fabrication of the sensor (reproducibility) and repeatability of the sensor’s performance is demonstrated in this section.
Figure 4.20: Three POF sensors to measure the breathing signal simultaneously at three different attaching points.

Table 4.1: Positions of three sensors measuring the respiratory signal.

<table>
<thead>
<tr>
<th>Layout</th>
<th>Sensor 1</th>
<th>Sensor 2</th>
<th>Sensor 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Layout 1</td>
<td>Back (below shoulder blade)</td>
<td>Chest (Upper costal)</td>
<td>Belly (Middle Abdomen)</td>
</tr>
<tr>
<td>Layout 2</td>
<td>Belly (Middle Abdomen)</td>
<td>Back (below shoulder blade)</td>
<td>Chest (Upper costal)</td>
</tr>
<tr>
<td>Layout 3</td>
<td>Chest (Upper costal)</td>
<td>Belly (Middle Abdomen)</td>
<td>Back (below shoulder blade)</td>
</tr>
</tbody>
</table>

4.6.2 Comparison with a Commercial Device on Different Subjects

The practicability and accuracy of the POF sensor are further evaluated with direct comparison to a conventional respiratory monitoring sensor in this study. Measurement of the POF sensor was conducted simultaneously with a widely used existing commercial
device – a Neulog respiration monitoring sensor. The first test was designed to verify the normal breathing signal of four different subjects in correlation with the Neulog sensor. This is to investigate the relationship between the two sets of signals and confirmed the accuracy of the POF sensor through direct comparison made to a recognized commercial respiratory sensor. The second test compared the normal breathing and deep breathing between both sensors on two subjects. The purpose of this assessment was to verify the accuracy of the breathing signal when the subject switched from normal breathing to deep breathing. This assessment was repeated with the subject performing breathing activities in a sitting posture and supine position (subject lying down with the face and torso facing upward). The changes of posture (e.g. sitting/standing and supine position) can modify the contribution of ribcage and abdomen regions in breathing [141]. Hence, the sensor’s accuracy in detecting the breathing signal when the subject in the supine position was investigated from the respiratory signal.

4.6.2.1 Neulog Respiration Monitor Belt Sensor

A commercial device - Neulog respiration monitor belt sensor NUL-236 also shown in Figure 4.21 [142] was used to verify the respiratory signal from the POF sensor. The Neulog sensor works based on piezo resistive effect where the resistance of a silicone element mounted between metal foils changes dimension according to exerted pressure. The transducer is electrically connected into a Wheatstone bridge circuit as a strain gauge so that the output voltage varied based on the measured absolute pressure. The respiration monitor belt is attached directly to the sensor through a flexible rubber tube and a hand pump with a pressure release valve. The belt contains a flexible partial pressure membrane that measures the pressure applied on the belt when the user inhales. Pressure that is applied to the belt’s air bladder is detected by the internal sensing unit and
converted to a voltage reading in an arbitrary units. Respiration signal is monitored through the reading of pressure changes.

To use the Neulog sensor, the NUL-236 belt unit is connected to a USB-200 module which provides power from a PC and communication between the PC and the sensor. The respiration belt is wrapped tightly around the user’s lower ribs and diaphragm area. Velcro pads should face outwards and the rubber tubing that connects to the belt is located above the user’s navel and facing downwards. The air pressure release valve is tightened and the hand-pump is used to fill the belt’s bladder until it forms a snug but not uncomfortable fit. The sensor outputs the respiration wave function using an arbitrary analog unit (Arb). The range and operation modes of the sensor is between 0 to 20,000 arbitrary units. The ADC resolution of the sensor is 15 bit with a maximum sample rate of 100 S/sec.

![Neulog respiration monitor belt logger sensor NUL-236.](image)

Figure 4.21: Neulog respiration monitor belt logger sensor NUL-236.

### 4.6.2.2 Respiration Monitoring Assessment Protocol for Comparison with the Neulog Belt Sensor

In this investigation, four healthy adults (two males and two females) aged between 26 to 35 years old were recruited for the validation test. Selection criteria for the
breathing test was healthy adults between the ages of 18 to 65 with no lung disease. Approval of the study protocol was obtained from the Faculty of Science and Engineering Ethics Committee (Ref 2016_12_01) and is included in the Appendix D. Measurement was conducted with both the POF sensor and Neulog belt sensor simultaneously attached on the body as shown in Figure 4.22 to establish agreement of the measured breathing signals. The POF sensor was placed on the upper torso (around the chest) in this assessment. It is impossible to compare the POF sensor at different positions with respect to a commercial device as the respiration monitor belt has to be wrapped tightly around the diaphragm and lower ribs area. The volunteers were asked to relax and breathe normally for one minute during the test to demonstrate the feasibility of using the sensor on different subjects. During the measurement, both sensors’ data were recorded simultaneously to prove the reliability of the POF sensor and the agreement with the obtained breathing signal. The feasibility of the sensor being used on the different subjects was also demonstrated in this test.

The second examination was carried out on two healthy volunteers (one male and one female) to measure the signal in normal breathing and deep breathing activities. The subject was requested to sit down, relax and breathe naturally for one minute and has deep breathing following exertion for another one minute. This is followed by repeating the same breathing steps but with the subject lying down flat on a bed in a supine position (face and torso facing up). The respiration signal was recorded simultaneously for both the POF sensor and Neulog sensor. All the data obtained from breathing signal was processed using MatLAB™ software.
4.7 Summary

This chapter included a description of the 3-D printing and fabrication of the sensor probe. The sensor system assembly including the hardware component and software designed that is implemented in the system are fully elaborated. The complete POF sensor was calibrated with the bending angle in sagittal and lateral planes using an optical setup utilizing a precise rotational stage. The sensor’s performance was tested and evaluated on monitoring the spine bending at the lumbar spine region. Protocols of sensor testing on human subjects in spine bending and comparison with image acquisition and conventional Biometrics goniometer are elaborated. This chapter also discussed the respiratory monitoring assessment where the POF sensor was tested on human respiratory signal and comparison of the signals obtained with a commercial device - Neulog respiratory monitoring sensor.
CHAPTER 5 SPINE MONITORING ASSESSMENT DISCUSSION

5.1 Introduction

This chapter presents the results of the spine monitoring assessment using the POF sensor of this investigation obtained by attaching the sensor on the lumbar spine region of human subjects. Results from the calibration, testing and evaluation of the spine monitoring sensor is identified in this chapter. In the first section, characterization of the sensor using the optical setup described in Chapter 4, section 4.4 is investigated. The purpose of the sensor characterization is to obtain the relationship between the sensor’s output ratio and the spine bending angle. The sensor response from the calibration are further evaluated with the results calculated from theoretical analysis derived in Section 3.2. Secondly, the sensor is mounted on the lower back spine to observe the response when the subject perform different bending postures. It is important for the sensor to operate correctly at all times and provide a good and consistent response when operated on human subjects. The sensor also has to be comfortable to wear when the subject practices the bending assessment. The performance of the sensor and comparison with an ad-hoc imaging scheme are discussed in this chapter. Next, the feasibility of the sensor is further evaluated by testing on a larger group of recruited subjects (18 subjects). In this investigation, a commercial sensor – a Biometrics goniometer is used as a second reference measurement to validate the performance of the POF sensor. The relationship between the results of both sensors is fully discussed.

5.2 Bending Angle Calibration with Optical Component

Figure 5.1 illustrates the voltage obtained from each of the three output fibres channel (corresponding to received intensity) for bending at an increment step angle of
during calibration. The light coupled to the output fibre varies according to the tilting angle between the input and output fibres as previously described in Chapter 3. Figure 5.1(a) shows the output voltage from fibre I1 (intensity of the light coupled to fibre I1) increases while the intensity of fibre I2 and I3 decreases when the sensor bends to lateral left as defined in the previous section. At a bending angle of 12°, the saturation point is reached at which point fibre I3 receives a negligible amount of light coupled from the input fibre. Conversely, for lateral right bending, the light coupled to fibre I2 increases while fibre I1 and I3 decrease for larger bending angles. When the sensor is subjected to sagittal extension bending, the light intensity coupled to fibre I1 and I2 increases when the bending angle increases at step angle of 2°. In Figure 5.1(c), it is noticeable that at an angle around 10° - 12°, fibre I1 and I2 decreases which corresponds to the saturation point of the sensor. On the other hand, intensity of fibre I3 increases with a decrease in fibre I1 and I2 for sagittal bending in the opposite direction.

![Graphs showing output voltage for different bending angles](Image)

**Figure 5.1**: Output voltage of three output fibres at (a) lateral left, (b) lateral right, (c) sagittal extension and (d) sagittal flexion bending angle during calibration.
Output voltages from the three receiving fibres were converted into the output ratio utilizing the formula shown in equations 3.26 and 3.27 as discussed in the previous Section 3.3.1. The normalized output ratio was obtained directly using the LabView program described in Section 4.3.1 and was employed to estimate the bending angle. This provides automatic intensity compensation mechanism without the need for the use of a separate reference fibre.

Figure 5.2 shows the graph of the output ratio versus the bending angle in both lateral and sagittal bending cases. The output ratio increases when the sensor rotates to the left at the step angle of 2° (Figure 5.2(a)). At a bending angle of 12°, the output ratio increases up to approximately one at the saturation point. When the sensor’s bending angle decreases, the output ratio decreases from the peak and returns to the original data at the zero bending position. Conversely, the output ratio decreases to the negative value when the sensor is rotated to the lateral right direction as illustrated in Figure 5.2(b). The ratio reaches an approximate negative one value at the saturation point. For sagittal bending, the output ratio slowly increases to the value of one when the sensor bends in a larger angle for the extension direction. On the other hand, the output ratio decreases when the sensor bends in the sagittal flexion direction.

From the graph of Figure 5.2, it can also be observed that the sensor exhibits a larger variation of the output ratio (higher increments or decrements) at lower bending angles. This renders a higher sensitivity for the lower bending angle range from 0° to 4° (e.g. sensitivity, S1 = 0.1564/1° for lateral left bending) compared to a bending angle of more than 4° (S2=0.0759/1°). The variation of the output ratio for a fixed angle interval decreases (lower sensitivity) when the sensor bends at a higher bending angle and stops varying when the saturation point is reached. This is caused by the higher attenuation loss that results in less light being coupled to the output fibres when the bending angle is high.
At the saturation point, no further changes occur in the output ratio as the sensor is over bent and only minimum or no light is coupled to the output fibres.

Figure 5.2: The output ratio of three fibres against the bending angle for (a) lateral left, (b) lateral right, (c) sagittal extension and (d) sagittal flexion bending.

The output ratio of each bending angle was obtained from Figure 5.2. The average value from the repeated calibration for each angle was plotted and the results are shown in Figure 5.3. Figure 5.3 illustrates the relationship of the output ratio versus bending angle for both lateral and sagittal planes.
The POF sensor exhibits a non-linear relationship over the full range of angles, where saturation is reached at bending angles around ±12° in both bending directions. The three output fibres ideally receive equal amount of light when both tubes are in the aligned center position. Thus, the output ratio is near to zero value at the bending angle of 0°. An almost linear response is observed at the bending angle between -4° to 4°, with the linear correlation coefficient, R² of 0.9971 for lateral bending and 0.9928 for sagittal bending when the data was subject to a linear regression analysis. The larger bending angle leads to greater non-linearity and lower slope (sensitivity) in the bending data. This is attributed
to the increasing degree of misalignment between the input and output fibres that give rise to an exacerbated loss in coupling between the transmission and receiving fibres at the large bending angle values. The sensor’s operating range is limited to the range of ±12° as further bending of the sensor will result in a lower sensitivity.

The sensor’s response for bending angle between -12° to +12° has been fitted to a third order polynomial equation and is shown as the solid line in Figure 5.3. The polynomial equations obtained for the lateral and the sagittal bending assessment are as follows:

\[ y = -0.0004x^3 + 0.0009x^2 + 0.1392x - 0.0835 \]  \hspace{1cm} (5.1)

\[ y = -0.0003x^3 + 0.00009x^2 + 0.1143x + 0.0637 \]  \hspace{1cm} (5.2)

The lateral and sagittal polynomial fit in equations 5.1 and 5.2 exhibit a good correlation coefficient, \( R^2 \) of 0.9970 and 0.9929 respectively. These polynomials were used during the actual measurements to interpret the sensor output as a bending angle. For the sagittal bending assessment in Figure 5.3(b), the polynomial fit is not ideal for angles lower than -6°. This is due to the asymmetrical configuration of the sensor. During the sagittal bending, a higher intensity of light coupled to the two fibres during bending in the positive direction whereas more intensity of light coupled to only one fibre when bending is in the negative direction.

From the sensor calibration, specifications of the sensor such as the sensitivity, accuracy and resolution of the sensor were calculated. As the sensor has a non-linear response for both sagittal and lateral planes bending, sensitivity of the sensor varies with different angle values. Therefore, the sensitivity of the sensor is calculated based on five different sections of the characteristic angle response (bending angle) and is summarized in Table 5.1.
Table 5.1 : Sensitivity of the sensor at different bending angle range for lateral and sagittal planes.

<table>
<thead>
<tr>
<th>Bending Angle Range</th>
<th>Lateral Plane</th>
<th>Sagittal Plane</th>
</tr>
</thead>
<tbody>
<tr>
<td>8° to 12°</td>
<td>0.0377/1°</td>
<td>0.0348/1°</td>
</tr>
<tr>
<td>4° to 8°</td>
<td>0.0818/1°</td>
<td>0.0707/1°</td>
</tr>
<tr>
<td>-4° to 4°</td>
<td>0.1454/1°</td>
<td>0.1267/1°</td>
</tr>
<tr>
<td>-8° to -4°</td>
<td>0.0680/1°</td>
<td>0.0540/1°</td>
</tr>
<tr>
<td>-12° to -8°</td>
<td>0.0131/1°</td>
<td>0.0411/1°</td>
</tr>
</tbody>
</table>

It can be observed that the sensor exhibits highest sensitivity (larger variation) at the lower bending angles between -4° to 4° for both bending planes. The sensitivity gradually decreases at larger bending angle and exhibits a minimum sensitivity at the bending angle of more than ±8°. The operating range of the sensor is thus within the range of ±12° as previously discussed due to the poor sensitivity at saturation point that is occurs at high bending angle values.

The lateral and sagittal output ratio from three output fibres were monitored for a continuous one hour period without bending to investigate the output ratio drift of the sensor. The test was conducted in order to investigate the signal’s consistency and to determine the sensor accuracy. The output drift of the sensor was calculated based on the long term output drift test results presented in Figure 5.4. The drift was calculated from the maximum fluctuating value (max-min) during the one hour test. From Figure 5.4(a), the varying value in the lateral ratio is between -0.05154 to -0.06087, which gives the output drift value, $E_{od}$ of 0.00933. For the sagittal ratio, the fluctuating value is in the range of -0.03174 to -0.03885, which gives the output drift value, $E_{od}$ of 0.00711.
Figure 5.4: One hour output drift for (a) lateral and (b) sagittal ratio signals.

The error due to drift for each bending angle section presented in degree angle is calculated using the following equations:

\[ E_{\text{OD}}(°) = \frac{E_{\text{OD}}}{\text{Sensitivity}} \]  

(5.3)

Where \( E_{\text{od}} \) is the output drift value obtained from Figure 5.4 and sensitivity for each bending angle section is obtained from Table 5.1. The output drift of the sensor in angle degree for each bending range in lateral and sagittal planes was summarized in Table 5.2.
Table 5.2: Output drift, $E_{OD}$ (in angle degree) of the sensor at different bending angle range for lateral and sagittal planes.

<table>
<thead>
<tr>
<th>Bending Angle Range</th>
<th>Output drift, $E_{OD}$ (in angle degree)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Lateral Plane</td>
</tr>
<tr>
<td>8° to 12°</td>
<td>0.2475°</td>
</tr>
<tr>
<td>4° to 8°</td>
<td>0.1141°</td>
</tr>
<tr>
<td>-4° to 4°</td>
<td>0.0642°</td>
</tr>
<tr>
<td>-8° to -4°</td>
<td>0.1372°</td>
</tr>
<tr>
<td>-12° to -8°</td>
<td>0.7124°</td>
</tr>
</tbody>
</table>

From Table 5.2, the sensor exhibits a lower output drift in angle degree of 0.0642° in the lateral plane and 0.0561° in the sagittal plane for a low bending angle range between -4° to 4°. The output drift of the sensor calculated in the worst case scenario (at minimum sensitivity) is 0.7124° for lateral plane and 0.2042° for sagittal plane.

As the sensor bending response is non-linear, the accuracy of the sensor is also dependent on the bending angle range. The accuracy of the developed POF sensor is calculated as follows:

$$\text{Accuracy} = \frac{\text{Bending Angle} - E_{OD}(\text{in angle})}{\text{Bending Angle}} \times 100\%$$

(5.4)

Where bending angle is the angle value obtained from the sensor during bending. The percentage accuracy for each bending angle section is thus summarized in Table 5.3.
Table 5.3 : Accuracy of the sensor at different bending angle range for lateral and sagittal planes.

<table>
<thead>
<tr>
<th>Bending Angle Range</th>
<th>Lateral Plane</th>
<th>Sagittal Plane</th>
</tr>
</thead>
<tbody>
<tr>
<td>8° to 12°</td>
<td>97.53%</td>
<td>97.96%</td>
</tr>
<tr>
<td>4° to 8°</td>
<td>98.09%</td>
<td>98.32%</td>
</tr>
<tr>
<td>-4° to 4°</td>
<td>98.39%</td>
<td>98.59%</td>
</tr>
<tr>
<td>-8° to -4°</td>
<td>97.71%</td>
<td>97.81%</td>
</tr>
<tr>
<td>-12° to -8°</td>
<td>92.88%</td>
<td>98.27%</td>
</tr>
</tbody>
</table>

From Table 5.3, the sensor exhibits the accuracy of 92.88% in the lateral plane and 97.81% in worst case scenario. The resolution of the sensor was calculated using the following equations:

\[
Resolution = \frac{\text{Noise amplitude}}{\text{Signal amplitude (full swing)}} \times \text{Angle Range (°)} \quad (5.5)
\]

Where noise amplitude represents the minimum detectable output ratio change, signal amplitude for a full swing is the difference of the output ratio value (max – min) and angle range represents the total operating range (degrees) of the bending sensor. The resolution of the sensor is thus calculated to be:

\[
Resolution = \frac{0.003}{1 - (-1)} \times 24° = 0.036° \quad (5.6)
\]

The proposed sensor therefore provides good sensor sensitivity, accuracy and resolution for the measurement of spine bending monitoring application. The performance of some other skin-mounted bending sensors is shown in Table 5.4 as a comparison reference.
Table 5.4: Performance of other skin-mounted bending sensor.

<table>
<thead>
<tr>
<th>Sensor</th>
<th>Measurement plane</th>
<th>Accuracy</th>
<th>Resolution</th>
<th>Working Range</th>
</tr>
</thead>
<tbody>
<tr>
<td>Side polished fibre sensor [109]</td>
<td>One</td>
<td>3.67°</td>
<td>1°</td>
<td>± 25°</td>
</tr>
<tr>
<td>Spineangel® Device [143]</td>
<td>One</td>
<td>&lt; 2.33°</td>
<td>1°</td>
<td>± 90°</td>
</tr>
<tr>
<td>Biometrics Goniometer [59]</td>
<td>Two (Sagittal and lateral)</td>
<td>2°</td>
<td>1°</td>
<td>± 90°</td>
</tr>
<tr>
<td>POF Sensor (in this investigation)</td>
<td>Two (Sagittal and lateral)</td>
<td>92.88%</td>
<td>0.036°</td>
<td>± 12°</td>
</tr>
</tbody>
</table>

The sensor performance provided in Table 5.4 showed that the POF sensor has a better accuracy and resolution than some of the skin-mounted sensor devices. Furthermore, the proposed POF sensor can be applied in both sagittal and lateral measurement, and at the same time is compatible for use in EMF environment. The actual performance of the sensor operated on human spine bending was tested and shown in section 5.4.

5.3 Theoretical Investigation and Evaluation

The output ratio versus bending angle response was evaluated using the theoretical analysis for power transmission between two fibres as derived in equations 3.28 and 3.29 in the previous chapter. The lateral offset of each output fibre relative to the input fibre was considered in the equation by substituting the coordinates of each output fibre in equation 3.25. The power ratio was plotted over the angular bending angle using Matlab software in the operating range of ±12° for lateral plane and sagittal plane.
Figure 5.5 shows a comparison of the theoretical simulation and experimental results for angular bending in the lateral and sagittal planes.

![Lateral bending ratio](image1)

![Sagittal Bending Ratio](image2)

Figure 5.5: Comparison of the output ratio between the theoretical and experimental results for angular bending in (a) lateral plane and (b) sagittal plane.

For the theoretical simulation, the output ratio was calculated using the parameter of the step-index multimode fibre with a NA of 0.51 as per the specification of the fibre used in the experiment. As shown in Figure 5.5, the trend of both bending measurements exhibits excellent agreement when fitting with the theoretical values using a separation gap of 1.34 mm between the input and output fibres. In the case of sagittal bending, the
maximum deviation between the theoretical and experimental results is at the output ratio value of 0.12 that occurs at 9° of flexion bending. For lateral bending, the maximum deviation value was 0.0877 and occurs at the bending angle of -2°. If the output ratio is offset to the range from zero to two, maximum deviation between the theoretical and experimental results is calculated using equation as follows:

$$\text{Differ} = \frac{\text{Ratio (theory)} - \text{Ratio(exp)}}{\text{Ratio(theory)}} \times 100\%$$  \hspace{1cm} (5.7)

The maximum deviation between the theoretical and experimental results are calculated to be 11.97% difference (at -2° tilting angle) for the lateral plane and 6.04% (at 9° tilting angle) for the sagittal plane.

In the experiment, the designated gap between the input and output fibres is set to be around 1.10 mm to allow some space for fibre bending. However, the separation gap between both tubes might extend a further small amount during the alignment process between both fibre tubes and during the bending. As discussed in the previous section, the bending ratio response shows a normalized output ratio of ±1 in the lateral bending and sagittal bending for the angle range of ±12°. The bending angle beyond this working limit results in saturation of the sensor output. A slight variation in the saturation point is observed in the output ratio of sagittal bending between positive and negative tilting angle due to the asymmetrical response of the sensor as previously discussed. In short, the sensor shows a good sensitivity when subjected to the angular bending within the operating region.

Based on the theoretical estimation, the sensitivity and operating range of the sensor can be adjusted from the separation gap between the fibres for other application purposes. Figure 5.6 denotes the theoretical estimation of the separation gap effect on
the bending angle response. The output ratio was plotted against the bending angle at four
different normalized gap ratios. The value of normalized gap ratio, $\varepsilon$ was calculated from
the separation gap divided by the value of 1.34mm, which is the theoretical estimated gap
that fits the response of the sensor fabricated in this investigation. For angular bending in
both directions, a larger separation gap (higher normalized gap ratio) results in a larger
sensing range. However, the sensitivity of the sensor is reduced in compensation with a
larger working range. For example, in the lateral bending ratio, the normalized separation
gap $\varepsilon$ of 1.5 constitutes a maximum operating range of $\pm 29^\circ$ but with a lower sensitivity
response of 0.0466/1°. On the other hand, the output ratio of the sensor with normalized
separation gap $\varepsilon$ of 0.9 has a lower working range of $\pm 5^\circ$ but with a higher sensitivity
response of 0.2767/1°. From the theoretical estimation, the bending response of the sensor
can be adjusted in future depends on the operating range or sensitivity required for the
applications.
Figure 5.6: Theoretical estimation of the fibre gap effect on the output ratio response for (a) lateral bending and (b) sagittal bending.

5.4 Dynamic (Time Resolved) Patient Assessment for Lower Back Bending

The sensor was tested on a human subject on lower spine bending after the calibration of output ratio with respect to the bending angle. Figure 5.7 illustrates the sample results recorded for the lateral and sagittal bending ratio versus time when the subject performed different static posture with the sensor attached at the lower back as shown in Figure 4.11.
In the case of the lateral bending assessment, the user was initially in a normal upright standing posture (middle) and the ratio value at this rest position was recorded as -0.17. The output ratio signal increased rapidly when the subject bent over towards the lateral left and stayed relatively constant as this posture was held for 15 seconds. The ratio signal reduced to the initial position when the subject slowly returned to the original standing posture. In contrast, the output ratio values decreased when the subject performed the lateral right flexion. A small shift from the zero position can be noticed when the subject is not bending. This may be attributed to a manual positioning error of the user when placing the sensor on the body at rest posture but this posture not being
precisely straight. In future tests, this problem was resolved by offsetting the sensor to zero value after placing the sensor on human subject in a straight posture.

For the sagittal assessment result as presented in Figure 5.7(b), the subject starts from the upright standing position (middle posture) and slightly bends forward. The output ratio decreased to a negative value from the original data when the subject performs sagittal flexion (forward). The signal decreased further when the subject bends forward at a higher bending degree. The signal returned to the initial value when the subject returned to the initial upright standing posture. When extending backward, the ratio signal moved to the positive values as shown in Figure 5.7(b). In the sagittal bending assessment, the small shift from the zero position can be observed from the data when the subject is not bending. This can be attributed to the fact that the spine of human resembles an ‘S’ shape even when the subject is not bending. The natural lower spine curvature in humans should observe as a forward curvature named lordosis (forward curve) at the lumbar spine and kyphosis (backward curve) at the sacral area. The results demonstrate the same signal trend when the subject repeats the bending exercise on both lateral and sagittal planes.

The output ratios were converted into angle (degrees) using the Matlab software to compare with the values extracted from the image acquisition and Biometrics sensor used as a reference device in later tests. Equations 5.1 and 5.2 obtained from the previous calibration were first included in the Matlab program with the x value being the sensor’s operating range from -12 to +12 (degree). Each output signal value was converted into the angle value using a ‘for’ loop statement. The code written in the loop for conversion is shown as below:

\[ y_{\text{diff}} = \text{abs}(y - y_{\text{out}}(i)) \]  
\[ [\text{value } idx] = \text{min}(y_{\text{diff}}) \]
\[
\text{angle} = [\text{angle} ; x(idx)];
\]  

(5.10)

where \( y \) is the output ratio from the calibration equation 5.1 for lateral bending and 5.2 for sagittal bending, \( x \) corresponds to the angle value (degree) from the equations 5.1 and 5.2 and \( y_{\text{out}} \) is the measurement results in the output ratio. Equations 5.8 and 5.9 above determine the nearest value of the output ratio from the measurement signal that is matched with the calibration equation (Equations 5.1 and 5.2). The ‘Min’ function in equation 5.9 is used to find the indices of the smallest difference between the output ratio of calibration equation and signal results. The closest difference is then used to locate the angle through its corresponding index number, \( \text{idx} \) in equation 5.10.

5.5 Comparison with the Image Acquisition Technique

The output ratios obtained from the sensor as presented in Figure 5.7 were converted into bending angles using the method as discussed above. The measurement angles were compared with the values extracted from the image acquisition technique described in section 4.5.1.2. The relationship between the bending angles measured from the POF sensor and image acquisition angles \( \theta_1 \) and \( \theta_2 \) (as defined in Figure 4.14(a)) when two independent subjects perform the bending exercise outlined in the previous section are shown in Figure 5.8. The results demonstrate a good fit with a second order polynomial response when subject A (female) and subject B (male) bend at a sagittal angle \( \theta_1 \) in the range -15° to 7°. Both subjects have a similar trend of relationship when comparing the overall inclination angle \( \theta_1 \) from the image acquisition with the sensor’s signal. However, the slight variation of bending angle relationship between two subjects is dependent on differences in the movement of individual subjects. Results of both subjects for the bending angle corresponding to imaging angle \( \theta_2 \) also exhibit a second
order polynomial response as shown in Figure 5.8(b). The imaging angle $\theta_2$ represents the large arch angle of the lumbar spine at different static postures. Hence the variation of angle relationship between both subjects can be attributed to different arch angles of the individual’s lumbar spine.

Figure 5.8: Relationship between measured bending angle from the POF sensor and the angle captured from the imaging system (a) angle $\theta_1$ and (b) angle $\theta_2$ for the sagittal bending assessment.

Figure 5.9 shows the time resolved output from the POF sensor as well as the imaging angle $\theta_1$ (obtained from Figure 4.14(b)) during the standard lateral bending assessment for both subjects. Both sets of data were plotted together on the same x and y
axes and were referenced to the zero angle representing the subject standing in the straight position. This was performed to reduce human error when placing the sensor on a human subject at the lumbar spine. For lateral bending, it is easier to compare the sensor and image angles as three spots were placed on the sensor. When the subject performed different static postures, the movement of the sensor could be more easily captured for analyzing as the spine appears as a vertical straight if viewed from the directly behind the subject. Figure 5.9 shows that there is a difference of angle value when the subject bends to the lateral left and right. This can be attributed to the locations of the three green markers that are placed on the sensor for image acquisition. The green markers were placed at a larger separation than the gap between the input tube and output tube of the POF sensor. Therefore the imaging angles obtained are larger (exaggerated) compared to the angle measured from the sensor.
Figure 5.9: The time resolved lateral output angle response of the POF sensor and imaging system for the core of (a) subject A and (b) subject B.

Figure 5.10 is a scatter plot of the POF output versus the imaging angle for all values tested on subject A (Figure 5.10(a)) and subject B (Figure 5.10(b)). Figure 5.10 illustrates the relationship between the responses of the POF sensor compared with the imaging angle for both subjects. A positive angle is obtained when subject performs the lateral left bending and a negative value corresponds to the right bending. The trend of bending for both subjects is in agreement with the imaging angle. However, both subjects produce slightly different relationships due to the individual variation of bending characteristics and body (spine curvature) shapes of the subjects.
Figure 5.10: The relationship between the lateral bending measurement data obtained from the sensor versus imaging angle $\theta_1$ for (a) subject A and (b) subject B.

The results demonstrate the feasibility of the sensor for measurement and producing reproducible sets of result for spinal monitoring in both sagittal and lateral directions. In the next part of the research, the POF sensor was further evaluated and verified by testing the sensor on a larger group of volunteers in a physiotherapy clinic setting. The bending angle at the lumbar spine was directly compared with a commercial spine monitoring device, a goniometer from Biometrics Ltd.

5.6 Sensor Evaluation and Comparison with Biometrics Ltd Goniometer

The POF sensor was evaluated on 18 healthy subjects following the assessment protocol as discussed in section 4.5.2.2. Both the POF sensor and Biometrics goniometer
sensor’s signals were recorded simultaneously and converted into an angle value as discussed in section 5.4 for the POF sensor and section 4.5.2.1 for the Biometrics goniometer. The signals were referred to the angle 0° when the subject stands at the upright standing posture. Figure 5.11 illustrates the sagittal bending signal of both the POF sensor and Biometrics goniometer instrument from one of the subjects. The results of both sensors show good agreement in the bending trend for the sagittal plane assessment. The positive angle in the graph represents the flexion bending while the negative angle represents the extension. The square peak starting at the time, t = 10 s demonstrates the increasing angle signal when the subject was instructed to bend forward until the overall body inclination of 5° was reached (measured using a manual inclinometer as described in section 4.5.2.2). The plateau in the square signal shows the bending angle when the subject holds this posture for 10 s. The signal decreased at time 20 s when the subject returns to the relaxed standing posture. At the time around 30 s, a larger increase of square peaks was observed in the signal due a larger flexion bending angle. The higher angle signal starting at time 50 s and 70 s denotes the flexion bending assessment at an inclination angle of 15° and 20°. At time 90 s, the subjects performed the extension bending (backwards bending) which coincided with the change of the respective signals to a negative angle value from the original position (reverse square peak). The signals of both sensors showed a larger drop at the time 110 s when subjects were asked to bend at a higher degree of extension (20° as recorded on the manual inclinometer or until uncomfortable).
Figure 5.11: Sagittal bending signal measured from the POF sensor and Biometrics goniometer in the sagittal plane assessment.

The bending signals recorded in the lateral plane assessment for both sensors are presented in Figure 5.12. A negative bending angle denotes a lateral left flexion while a positive angle represents lateral right flexion bending. Both sensors measured a drop in the interval between 10 s to 20 s when the subject was asked to bend to the left direction and hold the posture for a further 10 s. The signal drops further starting at the time \( t = 30 \) s when the subject bent at a higher degree to the left direction. At the time \( t = 50 \) s and \( t = 70 \) s, the subject performed the lateral right flexion from the upright standing posture. The response coincided with an increase of the respective signals at time 50 s and 70 s for two different degrees of bending.

The consistently higher angle value obtained from the Biometrics goniometer compared to the POF sensor for the bending assessments can be attributed to several reasons.
Firstly, both sensors were placed on a different lumbar spine region although they were placed near to each other.

Secondly, the POF sensor measures the bending of lumbar spine in the L3-L4 region while the Biometrics goniometer measures the whole lumbar spine section (T12 to S1). This is due to the overall sensor length in the case of the Biometrics goniometer being around 250mm (endblock to endblock) which is much longer than the length of entire POF sensor which is only 70mm.

Consequently, the bending angle obtained from the Biometrics goniometer measurement includes a larger lumbar spine range and different lumbar spine locations. The same condition is applied for the sagittal bending measurement.

![Figure 5.12: Lateral bending signal measured from the POF sensor and Biometrics goniometer in the lateral plane assessment.](image)

The bending angle at specific static posture (the sagittal and lateral planes assessment) was obtained from the flat regions of the square peaks of the signal in Figure 5.11 and Figure 5.12. The signals measured as plateau peaks from both sensors were
smoothed and the average angle at specific posture was obtained using the Matlab program described in Section 3.6.3. Figure 5.13(a) illustrates a representative scatter plot obtained from the plateau peaks in Figure 5.11 when the subject performed the sagittal bending assessment. Six bending angles from different static postures (four flexion and two extension bending) were obtained from the sagittal bending assessment. The same scatter plot was obtained from the lateral bending assessment as illustrated in Figure 5.13(b). For the lateral assessment, angle from four plateau peaks which include two left and two right flexion was acquired for each subject. The simultaneously captured results from the Biometrics goniometer and POF sensor could therefore be compared directly with each other.
Out of 18 participants, three results (two females and one male) were removed from the evaluation due to them being considered as outliers resulting from measurement error. The most likely reason for the signal to be out of trend is that during the trial one or both of the sensors slid along the skin (skin movement) when the patient was instructed to move to a different posture. The slight dislocation of the sensor during the measurement can be due to the subject’s excessive perspiration or use of body oils/lotion, causing the adhesive tape to malfunction (lose adhesion). Human error when placing the sensor at an inaccurate spine section might also cause the measurement error.
The angle measurement from the POF sensor was therefore compared directly and correlated with the Biometrics goniometer for 15 participants. The scatter plot data (as illustrated in Figure 5.13) from all test subjects were combined in a single graph for each sagittal and lateral bending assessment to investigate the correlation between the two sensors. Figure 5.14 shows the relationship between the angle measured from the Biometrics goniometer and POF sensor in the sagittal and lateral planes. The results from both bending planes demonstrate a linear relationship that intercepts at zero. For the sagittal bending, the Biometrics sensor exhibits a linear relationship and the equation corresponding to the POF sensor is as follows:

$$y = 6.409x$$ \hspace{1cm} (5.11)

where $y$ is the angle measured from the Biometrics goniometer and $x$ is the angle obtained from the POF sensor. The lateral bending assessment corresponds to the following linear relationship:

$$y = 12.603x$$ \hspace{1cm} (5.12)

where $y$ is the angle measured from the Biometrics goniometer and $x$ is the angle obtained from the POF sensor. Both of the sagittal and lateral assessment results demonstrate a good linear correlation coefficient, $R^2$ value of 0.8696 and 0.8599, respectively.

For the sagittal bending case shown in Figure 5.14(a), the positive angle denotes the flexion/forward bending while the negative angle illustrates the extension bending. More points were accumulated at a positive angle as the protocol involved four different flexion angles and two extension angles obtained from each of the subjects. From Figure 5.14(b), the positive angle values obtained from lateral bending represents the lateral right flexion while negative angle represents the lateral left flexion. The data points of all subjects distributed on both the positive and negative side of the graph when subjects bend at 10° and 20° (overall inclination angle obtained from the inclinometer) to the right and left lateral flexion. The results demonstrate that the POF sensor is capable of
monitoring and accurately measuring the lumbar spine bending in sagittal and lateral planes. The variation of angle value in the graph is probably indicative of slightly different placement of both sensors’ location (placed adjacent) for each subject. The subject was requested to rate the comfortability from 0-5 when using the sensor where 0 represents no discomfort and 5 is extreme discomfort. The 15 subjects rated an average comfortability of 0.6 when wearing the POF sensor for testing at the spine region.

Figure 5.14: Measured bending angle from the POF sensor with respect to Biometrics goniometer when subjects performed different bending postures in the (a) sagittal plane and (b) lateral plane.
5.7 Statistical Data Analysis

For a better comparison and evaluation of the sensor, the data was further analyzed using SPSS statistics software. Figure 5.15 and Figure 5.16 represent the mean angle measured from the 15 subjects at each bending posture. The mean bending angle represented as the bar chart shows a similar trend (consistent increase) in angle for both sensors when the subject was instructed to bend to a higher bending degree. The standard deviation represented as the scatter bar (error bar) in the chart represents the range of lumbar spine angle (movement pattern) that varied across the population at each measurement angle. Both sets of results exhibit a high consistency in the variation of lumbar spine angle measured, albeit demonstrating some variance from patient to patient.

Figure 5.15: Average sagittal bending angle measured from the (a) POF sensor and (b) Biometrics goniometer at different bending posture.
A Shapiro-Wilk test was used to test the normality of the data in which normally distributed data is the null hypothesis. The data were found to have a p-value more than 0.05 for each bending position and bending plane and hence the data distribution is normal. The correlation between the POF sensor and Biometrics goniometer was assessed separately for the lateral and sagittal bending assessment using the Pearson correlation coefficient ($r_p$) and coefficient of determination ($r^2$). To accept the strong validity of the POF sensor, the correlation coefficient value has to be larger than 0.5 ($r_p > 0.5$) in order to be classified as strong correlation [144]. A high correlation [145] is interpreted from the Pearson correlation measurements where the $r_p$ value is 0.933 for the sagittal bending angle and 0.930 for the lateral bending angle. The coefficient of determination, $r^2$ is also high which is 0.870 for the sagittal plane and 0.865 for the lateral plane.

As previously discussed, the POF sensor measured a lower bending angle as the sensor was fabricated to measure a smaller spine angle (L3-L4) compared with the Biometrics goniometer (T12 to S1). The high correlation demonstrated between both sensors indicates that the POF sensor which measures the smaller section of the lumbar spine can be referred to the bending angle of the whole lumbolumbar measured by the Biometrics goniometer. Assuming the POF sensor is able to estimate the bending of

Figure 5.16: Average lateral bending angle measured from the (a) POF sensor and (b) Biometrics goniometer at different bending posture.
whole lumbar spine section (same spine section measured by Biometrics goniometer), agreement between two sensors are compared. The linear correlation equation 5.11 and 5.12 were used as a multiplying factor to convert (calibrate) the angle from the POF sensor to a comparable angle with the Biometrics sensor. Level of agreement between sensors in this case is graphically established using Blant-Altman analysis [146]. Blant-Altman plot is carried out by plotting the differences of both angles on the y-axis and mean of both paired angles on the x-axis.

Figure 5.17 and Figure 5.18 display the Blant-Altman plot for the sagittal bending and the lateral bending assessment angle, respectively. The three horizontal lines represent the mean value and the approximate 95% confidence intervals, which is the mean ±1.96 SD (standard deviation). The difference is normally distributed (p>0.05) so the assumptions of Blant-Altman plot is satisfied. The spread of points are relatively consistence (no proportional bias) for both of the sagittal and lateral plane assessment. The plot also demonstrates a good level of agreement with mean difference ± SD of −0.1875 ± 3.658° for sagittal bending and 0.7972 ± 4.0849° for lateral bending. The mean difference is small between the Biometrics goniometer and POF sensor. Even though the sensor shows a good agreement with the Biometrics goniometer when it is calibrated using the linear regression correlation equation, it must be noted that the agreement comparison were in no way intended to replace the measurement for whole lumbar spine section. It merely to establish that the sensor could be used to measure a segment of lumbar spine bending (L3-L4) and shows the comparison made to a recognized commercial spine monitoring sensor.
In short, the concurrent validity of the POF sensor demonstrates that it is a suitable tools to aid the physician in monitoring the spine movement over time. The sensor has numerous attractive features to be used in clinical and special environment. It is low cost,
simple to use, portable and its all plastic features allow the use of the sensor with no electromagnetic interference issue. The output of the device will facilitate valuable feedback to the physician on the patients that undergoing physiotherapy. The sensor able to provide a near real-time measurement for bending of spine in two different planes, sagittal and lateral. The sensor provides the extra benefits compared with other spine sensor such as BodyGuard™ sensor [60] and Spine DMS™ [147] that only capable to measure spine bending in one plane.

5.8 Conclusions

A 3-D printed POF sensor for measuring the bending angle of the human lumbar spine region has been designed and successfully fabricated. The output ratio of the sensor has been calibrated and found to exhibit a non-linear relationship corresponds to the bending angle with a working region of ±12° over the sagittal and lateral planes. The performance of the POF sensor is summarized in Table 5.5.

Table 5.5 : Performance of the POF bending sensor.

<table>
<thead>
<tr>
<th>Sensor Properties</th>
<th>Measurement plane</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Lateral</td>
</tr>
<tr>
<td>Operating Range</td>
<td>-12° &lt; θₙ &lt; 12°</td>
</tr>
<tr>
<td>Max. Sensitivity (-4° to 4°)</td>
<td>0.1454/1°</td>
</tr>
<tr>
<td>Min. Sensitivity</td>
<td>0.0131/1°</td>
</tr>
<tr>
<td>Accuracy</td>
<td>92.88%</td>
</tr>
<tr>
<td>Resolution</td>
<td>0.0360°</td>
</tr>
</tbody>
</table>
The bending response of the sensor has shown good agreement when evaluated and compared with the theoretical calculation results. Testing of the POF sensor on human subjects has been conducted to demonstrate the capability of the sensor to monitor the bending of lumbar spine region. Experimental results were recorded and compared with angular deformation values obtained from a simultaneous ad-hoc image capture technique using a standard web camera. The results demonstrated the repeatability of the sensor for monitoring spinal bending. A clinical assessment of the POF sensor based on a population of 18 healthy subjects and comparison with a reference instrument, a commercial Biometrics goniometer for spine monitoring has been successfully undertaken in conjunction with the Physiotherapy department at UL. Fifteen valid results from this trial have shown a linear correlation between the POF sensor and Biometrics goniometer. The results also prove that the sensor is capable of accurate measurement of lumbar spine bending in both lateral and sagittal directions in the clinical environment. The portable and compact design of the sensor allows it to be used in a wide range of clinical settings with the additional advantage of being non-invasive, low cost and robust. The all plastic composition design further opens the possibility for sensor to be used in conjunction with MRI as well as X-ray scanning environment.
6.1 Introduction

The POF sensor for spine bending test described in the previous chapter has also been used to measure the human respiration signal. The sensor was placed on the upper body to track the movement of the body during respiration activity. Initial tests have been carried out by placing the sensor at four different locations of the body including the front and back positions as mentioned in section 4.6.1. This is to demonstrate the feasibility of the sensor to detect the breathing signal at different locations. Following these test, a more comprehensive assessment was performed by measuring the respiration signal simultaneously at three different locations on the body using three separate but identical POF sensors. Measurements of the breathing signal from normal breathing, breath holding and deep breathing activities at different attachment positions have been made. The consistency of the sensor’s manufacture and repeatability of the performance for each sensor has also been investigated from the simultaneous test of the three sensors in this examination. Finally, the POF sensor was compared with a commercially available respiratory monitoring sensor – NeuLog respiration monitor belt logger sensor NUL-236. Both sensors were attached on the body simultaneously to measure the respiration signal on four healthy volunteers. The respiration data from normal breathing and deep breathing patterns and in sitting and supine positions were obtained and are compared in the last part of the chapter.

6.2 Sensor Dynamic Response

As demonstrated in Chapter 5, the sensor is able to measure the signal in both orthogonal (lateral and sagittal) planes. However, the optical signal from the movement
in sagittal (vertical) plane alone is sufficient to determine the respiration signal. The sensor has been previously calibrated to minimize the position error (relative placement of the fibres) as discussed in section 4.4 even though only peak to peak cycle time and relative amplitude of signal are considered for analyzing the respiration signal. Figure 6.1 illustrates the results of fine calibration for a 0.5° interval in both lateral and sagittal axis from the sensor characterization. The sensor demonstrates a high resolution to be able to detect movement changes as small as 0.5° in both axis. The fine calibration was repeated three times and the average output ratio for each 0.5° bending angle was plotted as discussed in the previous section (5.2). The results show a linear response for bending angle range between -4° to 4° for both lateral and sagittal bending. The sensitivity for lateral bending is 0.1454/degree while the sensitivity for the sagittal bending is 0.1267/degree. The purpose of this characterization was to determine the sensitivity of the sensor being used as a respiratory monitoring sensor. However, no further calibration is required for the respiratory measurement as only the requirement for the signal detection is the main concern for this application.
Figure 6.1: The output ratio value at 0.5° step bending on both lateral and sagittal axis from the sensor calibration.

The sensor was tested on human subjects to investigate the respiration signal. The sensor probe was initially placed at the upper costal (chest) using a double-sided medical tape as shown in Figure 4.18. Figure 6.2 shows the time resolved raw data output (representing intensity) of the three output fibres obtained during the detection of normal breathing activity for a one minute duration. The voltage amplitude for each output varied individually and its value depends on the attachment point on the surface of the body. The respiratory waveform signal corresponds to the movement of the chest/ribcage at inhalation and exhalation. From the results, the output signal from fibre I1 and fibre I2 are almost in phase. Conversely, the waveform from fibre I3 is out phase with fibre I1 and I2. The reason for this is that when the lungs expand and contract during respiration, the movement of the chest causing the sensor to bend anterior and posterior in the sagittal direction. From the sensor’s configuration (refer to inset of Figure 6.2), fibre I3 is in the opposite direction of fibre I1 and I2 that are aligned horizontally. The periodic increase
and decrease of each output voltage is due to the positioning of the input fibre sensor that
is shifted upwards and downwards during the sagittal movement. When the input fibre is
shifted upwards due to the chest movement, the light coupled to fibre I3 increases while
fibre I1 and I2 decrease and vice versa.

![Diagram showing fibre configuration and output voltages](image)

Figure 6.2: The output voltage from the three output fibres during the measurement of
normal breathing. The inset shows the fibres configuration of the sensor during
respiration assessment.

From the output voltage of three fibres, the output ratio in lateral and sagittal
movement was obtained directly using the LabVIEW programme incorporating equations
3.26 and 3.27. The use of the normalized output ratio to calculate the respiratory signal
enables signal compensation that reduces interference caused by light source fluctuation
or temperature changes. As discussed previously in section 3.3.2, the volume of the chest
changes during the respiratory cycle. The chest region expands in three dimensions which
are in the anteroposterior, transverse and perpendicular planes [122]. The movement
measured by the sensor in different dimensions depends on the location of the sensor
attachment. Comparing the signal using the sagittal ratio (calculated from three output
fibres I1, I2 and I3) and lateral ratio (calculated from two output fibres I1 and I2),
movement in the lateral plane is relatively small (amplitude difference of 0.015) and less
stable compared to the movement in the sagittal ratio (amplitude difference of 0.080). This is observable from the similar phase that exists between fibre I1 and I2 in the lateral ratio and large out phase between fibre I3 and I1, I2 in the sagittal ratio. Consequently, in this respiratory assessment, only the output ratio from sagittal bending as shown in Figure 6.3 is considered for determining the respiratory signal as the signal reflects clearer respiratory movements.

Figure 6.3 shows a clear amplitude change of the respiratory signal. Each cycle of the waveform in the graph is formed by the inhalation and exhalation movement and represents one breath. From Figure 6.3, eleven peaks were observed in one full minute of recording the breathing signal. This value represents the breathing rate, which is 11 breaths per minute in this case. The response of the sensor demonstrates the practicability of the sensor for being used as a viable respiratory monitoring sensor. In all of the respiratory assessments, the measured data was normalized and smoothed using an in-house developed Matlab programme for comparison with the reference instrument output. Information from the breathing signal such as breathing frequency and data quantifying was studied using the same programme. The method of signal analysis is fully discussed in the next section (6.3).
6.3 Signal Analysis

This section discusses the signal processing technique used to analyze the breathing signal obtained from the sensor. Methods of signal normalization and data quantifying using MATLAB (Mathworks, Inc.) are described in this section. The respiration signal was generated in a modified ‘eye diagram’ graph to study the signal quality of the sensor.

In this work, the signal was recorded as 1 sample per 100 ms. Each of the breathing signals obtained were normalized and smoothed using MATLAB software. Data was normalized to the range between -1 to 1 using the following function:

\[ x = \text{Moving Average (10)} - \text{Moving Average (60)} \]  

\[ Y = \frac{x}{x_{\text{max}}} \]  

(6.1)

(6.2)

Moving average (10) represents a 10 point moving average signal where time for 10 points = 1 s, Moving Average (60) represents a 60 point moving average signal. The time for 60 points = 6 s, \( x_{\text{max}} \) is the maximum value of \( x \) during the cycle.

The first moving average in the equation is to smooth the signal in order to reduce the noise generated in the photodetector and amplifier circuit by averaging 10 subsequent data points in the signal (moving average 10). This value can be adjusted depending on the level of the noise in a signal. The second moving average value (moving average 60) is the mean value of the signal. This value is subtracted from the smoothed signal to shift the overall signal to the mean point, this being near zero in value. The resulting signal is divided by the maximum value of \( x \) during the cycle to normalize all data in the range of -1 to 1.

Breathing information from the patient’s respiratory signal such as respiration frequency and relative value of breathing depth (different breathing patterns) can be
obtained from the respiratory signal. Peaks in the signal can be calculated to determine the breathing frequency. As a substitute for manual counting, the frequency can be automatically extracted from the signal using the same Matlab programme. The period of each respiratory cycle was acquired by converting the normalized breathing signal into a sign function (signum function) (refer to Appendix C) and differentiating the instantaneous values as follows:

\[
Diff \text{Sign} = \text{diff} \left( \text{sign}(Y) \right); \tag{6.3}
\]

\[
X_p = \text{find} \left( Diff \text{Sign} >0 \right); \tag{6.4}
\]

\[
Freq = \text{length} \left( X_p \right)/(\text{time}); \tag{6.5}
\]

Where Y is the normalized breathing signal, \(X_p\) is the positive value of the signal found from the function ‘\(\text{diff}(\text{sign}(Y))\)’ and \(Freq\) is the calculated breathing frequency. Each positive value of the signal found from the function ‘\(\text{DiffSign}\)’ represents one period of the breathing signal (inhalation and exhalation). The breathing frequency, \(Freq\) was thus obtained by dividing the total number of breath counts by the total measurement time. Number of breaths in a minute can be acquired by multiplying the frequency value (in Hz or cycles per second) by 60 seconds.

The signal quality was further examined by calculating the signal to noise ratio and comparing the signal at each attachment positions of the sensor. The ‘eye diagram’ technique that is widely used in optical telecommunications were modified to assess the signal to noise ratio. Each period segment (one respiration peak cycle) was obtained using equations 6.3 and 6.4 above to capture one period of the breathing signal. The eye diagram graph from the breathing signal was thus generated by overlaying sweeps of different period segments on the same graph. For each respiration cycle, the maximum and minimum values were acquired. The range of each respiration waveform is the minimum value subtract from the maximum value. The mean and standard deviation of the signal’s
peak over a one minute measurement interval were calculated from the eye diagram graph using the mean and standard deviation function (std) available in Matlab software. In this case, the mean and standard deviation value calculated represent the variation of swing in peak to peak amplitude for each period of the signal.

6.4 Position Dependency

Section 6.4.1 discussed the result of the sensor placed separately at four different locations (front and back) when the subject performed normal breathing. The signal is compared and investigated to study the versatility of the sensor to be used at different positions. The second part of the results compared the signal when three separate sensors were simultaneously placed at three different positions on the same subject. The different breathing pattern signals (i.e. normal breathing, deep breathing, breath holding) obtained at the different locations were compared.

6.4.1 Comparison of four different positions

Figure 6.4 presents part of the respiratory signal (30 seconds) when the sensor was attached on a single subject at four different locations as shown in Figure 4.19. All signals at the four measurement locations reflect clear respiratory movements from inhalation and exhalation, without any apparent noise artefact. This indicates that the sensor is sufficiently sensitive to detect the respiratory signal when placed at any of the four locations according to the user’s necessity or comfort preference. In this case, the same sensor was attached on the same volunteer but at different measuring times and hence the variation in the respiration signal (frequency) is observed.
Figure 6.4: Breathing signal when the sensor was attached at four different positions on the body.

To examine the signal quality, the signal to noise ratio was determined as described in section 6.3 and calculated for each of the signals at different locations. Figure 6.5 represents the eye diagram formed from the respiratory signal after overlaying sweeps of period segments.

Figure 6.5: Eye diagram for the normalized breathing signal measured on four different locations: (a) upper costal, (b) lower costal, (c) abdomen and (d) back body below shoulder blade.
The amplitude of the graph (normalized intensity) was plotted such that the lower signal’s peak (minimum value of each cycle) occurs at or near the zero value. The horizontal axis corresponds to time with each sampling point on the horizontal axis representing an increment of 0.1 s. Therefore the full extent of the horizontal axis in each case corresponds to a total duration between 6 s for (a) and (b) and 8 s for (c) and (d). These represent breathing cycles captured from different breathing cycles and hence the small differences in the durations within each of the Figure 6.5(a)–(d). The small amplitude signal seen in the eye diagram in Figure 6.5(a) and (b) is the partial breathing signal of one full breath captured from one minute normal breathing.

From the eye diagram, the breathing signal for the sensor located at the back body position (see Figure 6.5(d)) exhibits less noise compared to the other positions. This provides a greater stability in obtaining the breathing signal and frequency data. However, it also shows the smallest deviation in amplitude swing (in successive cycles) which indicates that less information on the relative breathing depth can be obtained from this position. The reason is that lung expansion during normal breathing is more restricted posteriorly due to strong rib attachments. Hence a smaller change of body geometry is to be expected when the device is placed at the back position. The respiration signal obtained at the upper costal, lower costal and abdomen positions were similar but with slightly higher amplitude deviations compared to the back position. The results are further confirmed by treating the eye diagram signal as a mean and standard deviation of the signal’s peak over one minute measurement interval which are shown in Figure 6.6.

The bar chart in Figure 6.6 represents the mean value of each breathing depth (peak to peak amplitude) that corresponds to the locations in Figure 6.5. The error bar in Figure 6.6 represents the variation of each swing in amplitude as the Standard Deviation (SD). The SD value was obtained in each case from Figure 6.5 and was calculated using
the mean and standard deviation function in Matlab. From Figure 6.6, the signal from
the sensor placed at the back body position has the lowest peak variation, and has the
smallest error bar in the chart of Figure 6.6. The lower output signal variation at this
position is due to the smaller chest movement to the back during respiration as discussed
above. The variations observed in the peaks in Figure 6.5 and Figure 6.6 could also be
due to genuine fluctuations on the peaks of the respiratory signal which are known to
occur when the sensor was mounted on the subject at different times [134]. Therefore, in
the next section, three of the same POF sensors were placed simultaneously at different
positions to measure and compare the respiratory signal.

![Graph of mean and standard deviation of respiratory signal](image)

Figure 6.6: Mean and standard deviation of the respiratory signal at the four different
sensor locations.

In summary, the sensor when placed at all four locations exhibits a strong peak
signal for respiratory monitoring purposes. This indicates that the POF sensor can be
placed at different locations according to the patient’s comfort needs and perceived
suitability. The sensor placed at the front positions yielded a consistently higher
amplitude signal but also exhibited a larger variation in peak amplitude. On the contrary,
the signal obtained at the back body has a relatively lower peak variability but a lower
signal amplitude. In the next section, three identical POF sensors are placed on the same
subject at different positions to simultaneously measure the respiration signal. In this case the respiration signals and especially the frequency measured at the different positions are compared.

6.4.2 Simultaneous Measurement Using Three Sensors Located at Three Different Positions

Figure 6.7 shows the breathing signal of three sensors recorded simultaneously when attached at different positions on the body. Three sensor probes were placed on the three different locations on the body as discussed in section 4.6.1. The attachment locations for each sensor were switched at each measurement following the layout listed in Table 4.1. The subject was assessed for three different breathing patterns: normal breathing, breath holding and diaphragmatic breathing (deep breath) activities.
Figure 6.7: The respiration assessment results evaluating normal breathing, breath holding and deep breathing on three different positions (a) sensor 1 on back, sensor 2 on chest and sensor 3 on belly, (b) sensor 1 on belly, sensor 2 on back and sensor 3 on chest, (c) sensor 1 on chest, sensor 2 on belly and sensor 3 on back.
The result of Figure 6.7 shows a consistent sinusoidal waveform over the first 70 seconds of recording, which represents normal inhalation and exhalation motion. All of the sensors at the three positions exhibit an approximately equal peak values for normal breathing which represents the same breathing rate measurement. The variation of amplitude peak in the waveform is due to the positioning of the sensor, where the range of body movement during breathing is highly dependent on the attachment location of the sensor.

At the time around 70 seconds the subject was instructed to hold their breath. At this time the respiratory signal at all positions exhibits a relatively flat signal. This is due to the body movement from inhalation and exhalation activity being temporarily paused when the subject performed breath holding. Consequently the amplitude of the respiratory signal waveform temporary decreases to a near flat level. The breathing signal returned to a strong waveform when the subject resumed normal breathing and ceased again at the second minute (around \( t = 130 \) s) as soon as the subject holds their breath again. The ability for the sensor to detect the cessation of respiration gives the potential for the sensor to be used in monitoring patients with sleeping disorders such as obstructive sleep apnea syndrome where breathing repeatedly stops and starts during sleep [148]. Irregular breathing patterns of users can also be potentially detected from the change of respiration signal amplitude using this sensor.

At the time around 150 seconds (after 2\textsuperscript{nd} breath holding), the subject was instructed to perform the diaphragmatic breathing (deep breathing) pattern. The result is a significant increase in the signal amplitude when the subject switched from normal breathing to diaphragmatic breathing. This represents the movement of the lungs when respiration activity increases due to maximum lung expansion. Optimum breathing is known to occur to provide the highest lung capacity for gas exchange during diaphragmatic breathing [140].
For a better comparison, the breathing signal of the sensor in layout 1 (from Figure 6.7(a)) was plotted in the eye diagram graph using same method as discussed in Section 6.3. The normalized amplitude of normal breathing and diaphragmatic breathing signal at the three recording locations are plotted in Figure 6.8. All three sensors output response are correlated well. The figure shows a clear variation of signal’s amplitude, i.e. maximum normalized amplitude peak of around ±0.5 a.u. for normal breathing in Figure 6.8(a) and ±1.0 a.u. for diaphragmatic breathing in Figure 6.8(b). This explains the larger amplitude variation compared to the normal breathing due to the larger lung expansion for diaphragmatic breathing. Comparing the eye diagram of the respiratory signal located at the chest (upper costal), belly (abdomen) and back (below shoulder blade), the signal at the front chest and back position exhibit a more consistent signal compared to the belly position. The main breathing muscle is the thoracic diaphragm that is attached to the lower ribcage, which expands and contracts causing air to flow in and out of the lungs. The movement at the belly (lower abdomen) during breathing mechanism is caused by the diaphragm muscle, indirectly making the belly move outwards and inwards. Hence, the movement of upper body around the chest (ribcage) and lower abdomen are varied during the respiration.
Figure 6.8: Eye diagram of (a) normal breathing and (b) diaphragmatic breathing from the sensor signal at position 1.

To confirm the results, the mean and standard deviation of the signal’s peak for one minute duration from Figure 6.8 is shown in Figure 6.9. During normal breathing in Figure 6.9(a), the signal of the sensor placed at the back of the body has the lowest peak variation (smallest error bar). This is due to the lung expanding more to the front during normal breathing as discussed in the previous section. For deep breathing, the sensor at all three locations exhibit a similar amount of peak variation. It can be observed that the normalized signal amplitude of normal breathing for all locations is much lower than in the case of the deep breathing signal. This indicates that the sensor is able to monitor and identify the change of breathing pattern (from normal breathing to deep breathing).
Figure 6.9: Mean and standard deviation of the (a) normal breathing and (b) deep breathing signal at the three different sensor locations.

The measured frequency of normal breathing at the first minute and deep breathing for one minute are summarized in Table 6.1.

Table 6.1: Respiratory frequency of normal breathing and deep breathing (diaphragmatic breathing) measured using three sensors at different positions.

<table>
<thead>
<tr>
<th>Layout</th>
<th>Normal Breathing Frequency (Hz)</th>
<th>Deep Breathing Frequency (Hz)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Back</td>
<td>Chest</td>
</tr>
<tr>
<td>Layout 1</td>
<td>0.17991</td>
<td>0.17979</td>
</tr>
<tr>
<td>Layout 2</td>
<td>0.20921</td>
<td>0.20772</td>
</tr>
<tr>
<td>Layout 3</td>
<td>0.18301</td>
<td>0.18131</td>
</tr>
</tbody>
</table>

For deep breathing, the average respiratory rate decreases compared to normal breathing. The increase in the waveform amplitude and reduction of respiratory frequency in diaphragmatic breathing is normal, which indicates the increase in respiratory volume [149]. All three sensors measured a similar breathing rate (less than one breath difference for one minute breathing) at the three different measurement locations during normal breathing. This confirmed the consistency of the sensor manufactured as well as the
excellent performance of the individual sensors. The slight difference in breathing frequency measured is primarily due to the positioning of the sensor. During breathing, the movement of the body especially the chest and abdomen are clinically recognized to be slightly different [150]. Comparing the signals from the three sensors at the three locations, the back body and chest positions measurements produced a similar respiration frequency compared to the belly positions especially for deep breathing activity. This suggests that the location at chest and the back are better positions for monitoring the breathing signal.

The results of this section (6.4) have allowed potential variability in the breathing patterns of real patients at separate locations on the body to be identified and quantified. It is possible that such variations are normal as discussed above. This hypothesis will again be tested by comparing the signal from the POF sensor with a commercial respiratory monitoring sensor on four different patient. In the next section of the experiment, the signal recorded from the POF sensor is compared with a commercial product, the Neulog belt respiratory monitoring sensor to verify the accuracy and versatility of the sensor.

6.5 **Comparison with a Commercial Respiratory Monitoring Device**

In this section, the respiration signal from the POF sensor was simultaneously compared with a commercial product, the Neulog respiratory monitoring belt as discussed previously in section 4.6.2. The Neulog respiratory sensor was set to record 50 samples per second and its output is displayed as arbitrary units. The signals from the POF sensor and the Neulog sensor were subject to normalization to their respective maximum value and placed on the same time resolved graph with the POF sensor for direct comparison. In the first field test, the normal breathing signal from four volunteers was recorded for
one minute. Using this data it was possible to assess the accuracy and practicability of the POF sensor when tested on different subjects. In the second test, the POF sensor was tested using two different breathing patterns and data was recorded simultaneously with the Neulog sensor in order to enable direct comparison. For all of the breathing assessments, no calibration was required for the POF sensor to be adapted to the user’s specific features.

6.5.1 Normal Breathing

Figure 6.10 depicts the respiration signal of four volunteers under regular breathing (normal breathing) for one minute. The data was captured simultaneously using the POF sensor and the Neulog respiratory monitoring belt. Figure 6.10 clearly shows the classical increasing and decreasing pattern of the respiration signal from the inhalation and exhalation periods. All four subjects were healthy and exhibited normal breathing frequencies in the range of 12 to 18 breaths per minute. This is normal as regular breathing rate for adult is considered to be in the range from 10 to 20 breaths per minute [134]. The POF sensor captures a clear respiratory signal and this correlated well with the commercial sensor, even when the sensor was placed on the upper costal (chest). This constitutes location with the greatest disparity in signals between the belt device (worn around the abdomen) and the POF sensor located high on the chest. The slight variation in the signal from the POF sensor compared to the commercial sensor can be explained by the positioning of the two sensors at different body locations. The movement of the upper torso at the chest part and abdomen is slightly different during respiration activity [150].
The quality of the breathing signal was further analyzed and compared by studying the eye diagram as presented in Figure 6.11. The mean and standard deviation of the eye diagram is presented in Figure 6.12, which represents the mean and variation in swing of signal’s amplitude.
Figure 6.11: Modified eye diagram of respiratory signal from (a) the conventional Neulog respiratory sensor and (b) POF sensor placed at the chest region on four different subjects.

Figure 6.12: Mean and standard deviation of the peak amplitude of respiration signal.
The mean value from the Neulog sensor was deliberately offset by 0.4 a.u. to avoid overlapping of signals and to allow easy comparison of the standard deviation between both sensors. Both sets of signals exhibit a level of consistency in the variation of the peak amplitude, though not the normalized amplitude. This is strongly indicative that the POF sensor and the commercial sensor measure a similar variation in the respiratory peak amplitude even when located at different positions on the body. This also indicates that swing in the amplitude peak is the genuine variation of the signal amplitude and therefore it is not noise. The higher amplitude peak observed in Figure 6.11 signifies that the subject breaths deeper (taking a larger breath) while a smaller amplitude peak signifies a smaller breath. The depth of breath causes a change of displacement of the chest and abdomen which in turn result in a smaller or larger amplitude waveform. The measured depth of breathing may also uniquely depend on an individual subject’s respiration pattern.

From Figure 6.11 and Figure 6.12, it can be observed that the Neulog sensor exhibits a larger variation of standard deviation value among the subjects during normal breathing. This might be due to the difference in the pressure change from the respiration activity that is dependent on the Neulog belt’s air bladder. When the user inhales, a respiration pressure is applied directly to the Neulog belt’s air bladder and detected by the internal pressure sensing unit. The pressure detected (corresponding to respiration depth) is highly dependent on the pressure in the flexible membrane. A manual hand-pump is used to initially fill the belt’s bladder until it forms a snug fit to the subject before use and the pressure is primarily according to the user’s perceived degree of comfort. The observed variation of the standard deviation value (or amplitude sensitivity) is at least partially dependent on the internal pressure which ultimately determines the flexure of the sensor’s membrane when external pressure from the abdomen is applied during breathing (i.e. dependent on the pressure differential).
The respiratory frequency was obtained from the breathing signal as discussed in section 6.3 above. The resulting data and percentage difference of the POF sensor was calculated in terms of a difference between the results of the POF sensor and Neulog sensor. Table 6.2 shows the respiratory frequency from all subjects and the difference between both sensors. Results from all subjects show excellent agreement of breathing frequency recorded between the Neulog sensor and POF sensor of this investigation with the largest difference being 1.1%.

Table 6.2 : Respiratory frequency monitored on four volunteers from the commercial Neulog respiratory sensor and the POF Sensor.

<table>
<thead>
<tr>
<th>Subject No.</th>
<th>Respiratory Frequency (Hz)</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
</tr>
</thead>
<tbody>
<tr>
<td>Neulog Respiratory Sensor, $S_N$</td>
<td></td>
<td>0.30020</td>
<td>0.20013</td>
<td>0.28838</td>
<td>0.20013</td>
</tr>
<tr>
<td>POF Sensor, $S_{POF}$</td>
<td></td>
<td>0.30151</td>
<td>0.20134</td>
<td>0.28523</td>
<td>0.20134</td>
</tr>
<tr>
<td>% Difference</td>
<td></td>
<td>0.4363%</td>
<td>0.6046%</td>
<td>1.0923%</td>
<td>0.6046%</td>
</tr>
</tbody>
</table>

$$\text{% Difference} = \left| \frac{S_N - S_{POF}}{S_N} \right| \times 100\%$$

6.5.2 Assessment of Different Breathing Patterns for Sitting and Lying Positions

Figure 6.13 and Figure 6.14 show the normal breathing and deep breathing signals when the subject is at sitting and lying posture. Both signals were subject to normalization to their respective maximum value in each case and placed on the same time resolved graph. The signal from the POF sensor at chest position correlated well with the Neulog belt sensor in both cases. As discussed in the previous section, the cyclic (oscillatory) waveform in the figure represents the breathing signal formed by inhalation and exhalation during the breathing motions. When the subject switched from normal breathing to chest breathing, the overall signal amplitude increases for both sensors. Both
the fibre sensor and Neulog respiration sensor exhibit a consistently larger amplitude signal during deep breathing. However, the POF sensor exhibits a relatively larger amplitude change during the deep breathing. This is probably because the sensor is attached at the front chest position of volunteers and chest rises more during this type of breathing, causing the superior movement of upper front body compared with normal breathing. The slight noise observed in Figure 6.13(b) at time approximately 60 second on the Neulog sensor is due to the accidental motion of the subject when trying to change from normal breathing to deep breathing. This is termed motion artefact and is not observed on the POF sensor case. The small variance between signal from the POF sensor and the Neulog sensor is due to the different positioning of two sensors. As previously discussed, the movement of chest and abdomen are recognized to be slightly different.

![Figure 6.13](image_url)

Figure 6.13 : Normal breathing and deep breathing signal for both subjects during the sitting posture.
The average respiratory frequency of both sensors in sitting and lying postures were calculated and are shown in Table 6.3. Results for all subjects show good agreement in the measurement of breathing frequency for both sitting and lying postures. The difference between both sensors are up to a maximum discrepancy of 4%. All differences in the respiratory rate for the two sensors are less than 1 breath per minute, which is indicative of a high reliability of the POF sensor being used to monitor respiration signal. The good correlations between both sensors in the lying posture opens the possibility of using the sensor for bed bound patients or patients being asked to lie down when entering scanning machines e.g. MRI. The variation in signal amplitude at deep breathing also
gives the potential for the sensor to distinguish between different types of breathing. Parameters measured from the POF sensor in monitoring the respiratory signal provide possibility to measure the psychophysiological state of the patient, for example, when deep breathing with low frequency indicates that the patient is in a relaxed state [151].

Table 6.3 : Average respiratory frequency of the Neulog sensor and POF sensor for both subjects in sitting and lying posture.

<table>
<thead>
<tr>
<th>Respiration Position</th>
<th>Respiratory Frequency (Hz)</th>
<th>Sitting</th>
<th>Lying</th>
</tr>
</thead>
<tbody>
<tr>
<td>Subject No.</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Neulog Respiratory Sensor, $S_N$</td>
<td></td>
<td>0.15036</td>
<td>0.18429</td>
</tr>
<tr>
<td>POF Sensor, $S_{POF}$</td>
<td></td>
<td>0.15655</td>
<td>0.18519</td>
</tr>
<tr>
<td>% Difference $\frac{S_N - S_{POF}}{S_N} \times 100%$</td>
<td></td>
<td>4.083%</td>
<td>0.488%</td>
</tr>
</tbody>
</table>

6.6 Conclusions

The feasibility and reliability of the 3-D printed POF sensor as a respiratory monitoring sensor has been investigated and demonstrated. The compact optical fibre sensor in this investigation comprises a single structure that can attached to the patient at one of four attachment points. Results have shown that the sensor produces a strong signal amplitude and is sensitive enough to accurately monitor the respiration signal at all four locations. This indicated that the POF sensor is not restricted to being located at only one sensing region as is the case with many commercially available respiratory monitoring sensors. The POF sensor is less restrictive and offers greater comfort due to its small size. The user can also have the options of placing the sensor on the back of the body if necessary. The simultaneous testing of three identical sensors at different locations has demonstrated that the sensor can reliably detect normal breathing, deep breathing and
holding of breath from the change of signal amplitude at all positions. The good agreement obtained in the results of the three sensors also indicates that the sensor has good reproducibility even when fabricated in the laboratory or OFSRC at UL by the author of this thesis.

The sensor’s signal was verified via direct comparison with a commercial device, the Neulog Respiratory monitoring sensor. Testing was performed on four different volunteers, from which the results have shown good agreement with the Neulog instrument and demonstrated great versatility of the POF sensor. A comparison of the respiration rate measured with the Neulog sensor and POF sensor on four patients has shown that these values agree to within 1.0923% in the worst case scenario. When both sensor’s signals were compared with subjects performing the normal breathing and deep breathing in a sitting and lying posture, the difference rose to 4%. Results have been obtained in conjunction with physiotherapy practitioners, but the flexibility of the design of the sensor will allow it to be used with the patients that required special medical needs e.g. bed bound patients or the elderly. It also creates the possibilities for this sensor to be used in conjunction with an MRI or other electrically hostile medical scanning systems.
7.1 Conclusions

The work presented in this thesis has realized a low cost, portable and easy implementation of a 3-D printed all plastic POF based sensor for monitoring physiological parameters in clinical environments. The design, development and evaluation of the sensor for both lumbar spine bending and respiratory monitoring was fully implemented and described. The sensor works based on intensity modulation and illustrated an operating range of ±12° bending angle in both sagittal and lateral planes. The bending response has been evaluated with theoretical calculations and shown an excellent agreement when the experimental data was compared with the theoretical results. The POF sensor exhibits a non-linear bending response in both directions and the performance of the sensor was obtained and summarized in Table 7.1.

<table>
<thead>
<tr>
<th>Sensor Properties</th>
<th>Measurement plane</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Lateral</td>
</tr>
<tr>
<td>Max. Sensitivity</td>
<td>0.1454/1° (-4° to 4°)</td>
</tr>
<tr>
<td>Min. Sensitivity</td>
<td>0.0131/1° (-12° to -8°)</td>
</tr>
<tr>
<td>Accuracy (%)</td>
<td>92.88%</td>
</tr>
<tr>
<td>Resolution</td>
<td>0.0360°</td>
</tr>
</tbody>
</table>

The POF sensor has a good accuracy and resolution for the clinical based spine bending applications. Besides, the high sensitivity of the sensor, particularly at the small bending range of -4° to 4° allows the sensor to be applied in monitoring respiration signal while maintaining the same advantages.
In the first investigation, the experimental results were captured from the sensor mounted on human subjects and the results have been compared with angular deformation values obtained from an ad-hoc imaging acquisition method. The results have shown the sensor to both repeatable and reproducible for spinal bending in both sagittal and lateral planes. The sensor was further evaluated in a test involving 18 subjects and has shown a good correlation results with respect to the commercial Biometrics goniometer instrument. The feasibility of the fabricated sensor was demonstrated throughout the multi-subject sensor test. The ability of the sensor to be applied in 2-D (sagittal and lateral) continuous measurement and at the same time compatible for use in EMF environment making the sensor unique for spine assessment.

Apart from the application in spine monitoring, the feasibility of the POF sensor as a respiratory monitoring sensor has also been demonstrated. The sensor measured a strong and accurate signal when placed at four different locations of the body. This provides an additional advantage for the POF sensor in that it can be used at different locations (including the patient’s back) based on necessity and/or patient preference. Moreover, the sensor has shown the capability in detecting different breathing patterns (i.e. normal breathing, deep breathing and breath holding) when three of the identical sensors were fabricated and tested on different locations simultaneously. The POF sensor was further simultaneously compared with the conventional Neulog belt respiratory monitoring sensor when performed on four subjects. The small difference in value (less than 1.1%) between both sensors represents the good performance of the sensor for respiratory monitoring. The POF sensor also demonstrated a good agreement with the breathing signal measured using the Neulog sensor when subjects perform normal breathing and deep breathing in sitting and supine positions. This allows the sensor to be used for bed bound patient and when perform different breathing patterns.
Specifications of the all-plastic POF sensor in both spine monitoring and respiratory monitoring is summarized in Table 7.2.

Table 7.2: POF sensor specifications for both spine bending monitoring and respiratory monitoring.

<table>
<thead>
<tr>
<th>Sensor properties</th>
<th>Specifications</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sensor Probe Dimensions (H × W × L) (mm)</td>
<td>10 × 25 × 70</td>
</tr>
<tr>
<td>Sensor gap between Pods (mm)</td>
<td>30</td>
</tr>
<tr>
<td>Operating Region (deg.)</td>
<td>-12° &lt; 0 &lt; 12°</td>
</tr>
<tr>
<td>Measuring Plane</td>
<td>Sagittal and Lateral</td>
</tr>
<tr>
<td>Self referencing (automatic compensation)</td>
<td>Yes</td>
</tr>
</tbody>
</table>

In conclusion, the flexible and small size of the portable sensor system allow more comfort options (to a wide variety of body sizes) for long-term monitoring purposes in home monitoring or clinical environment. In the case of spine monitoring, the sensor demonstrate a good performance in multi-subject test and can be applied in both sagittal and lateral bending. When used as a respiratory monitoring sensor, the sensor demonstrates a distinctive characteristic of being able to accurately measure the signals when attached at several attachment points (include the back). Moreover, the all-plastic composition of the sensor provides the additional advantage that it can be used in conjunction with MRI scanning machines as well as X-ray based diagnostic scanning instruments. The compact design and ease of attachment coupled with its completely non-invasive properties of the sensor make it a viable contender for commercialization and widespread use in clinical settings.
7.2 Summary of Contributions Made in This Investigation

The research achieved a number of practical outcomes and advances in the state of the art as listed below:

- An all plastic fibre sensor head was developed to apply on various sizes of body and ages through direct mounting on the skin surface using only standard medical tape as an attachment means and the sensor can uniquely be used in special clinical environments such as MRI or X-ray scanning systems.

- The entire system is low cost and highly portable, where the measurement results can be stored in on-board SD card before transfer to a computer later.

- The sensor is capable of continuously measuring the bending of the lower back spine in both the lateral and sagittal directions.

- The sensor is able to accurately measure the human breathing signal and can be used in one of four attachment points at the body, including the upper front (upper costal, lower costal, abdomen) and at the back of body (below shoulder blade).

- The sensor is able to distinguish between irregular and regular breathing patterns including breath holding and diaphragmatic breathing. No pre-calibration of the sensor needed for the monitoring of respiratory signal.

- Flexibility and small size of the POF sensor allow more comfortable options for long-term monitoring purposes, and opens the possibility to be used with patients with special medical needs e.g. the elderly or neonatal.

7.3 Future Work

The work described in the thesis has demonstrated the development and fabrication of sensor device which is simple in construction, low cost in production and
portable. As the sensor proved to be effective in measurement of lower spine bending and the respiratory signal, a few measures can be taken in future to enhance the sensor performance and broaden its application space.

- Develop a sensor that able to simultaneously measure multiple point of spine segment.
- Develop a sensor that able to measure the axial rotation of the spine instead of flexion and extension.
- Incorporate a wireless module in the sensor system for better freedom of movement when wearing the sensor. However, adding the wireless module into the sensor may ultimately result in less movement freedom for patient due to larger size and would prohibit its use in an MRI environment if the transmitter is attached on the body.
- Different sensor can be designed and fabricated for other health monitoring applications such as to measure the limb motion or gait analysis.
REFERENCES


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6. Kam, W., O'Keeffe, M., O'Sullivan, K., Mohammed, S.W., O'Keeffe, S., Lewis, E., and Viphavakit, C., “A validation study of polymer optical fibre sensor for monitoring the angle of movement at lumbar spine for low back pain”, The Spine Journal. (Submitted)
APPENDIX A: LABVIEW BLOCK DIAGRAM AND FRONT PANEL

Labview block diagram part 1:
Labview block diagram part 2:
Labview front panel:
APPENDIX B: PROGRAMMING CODE FOR ARDUINO PLATFORM

#include <SPFD5408_Adafruit_GFX.h> // Core graphics library
#include <SPFD5408_Adafruit_TFTLCD.h> // Hardware-specific library
#include <SPFD5408_TouchScreen.h> // Touch library
#include <SD.h>
#include <SPI.h>
#include<Wire.h>

#define SENSIBILITY 300 // Calibrates value
#define MINPRESSURE 10
#define MAXPRESSURE 1000
#define YP A3 // must be an analog pin, use "An" notation!
#define XM A2 // must be an analog pin, use "An" notation!
#define YM 9 // can be a digital pin
#define XP 8 // can be a digital pin
short TS_MINX = 873;
short TS_MINY = 166;
short TS_MAXX = 72;
short TS_MAXY = 997;
TouchScreen ts = TouchScreen(XP, YP, XM, YM, SENSIBILITY); // Init TouchScreen :

#define LCD_CS A3 // LCD Pin
#define LCD_CD A2
#define LCD_WR A1
#define LCD_RD A0
#define LCD_RESET A4 // Optional : otherwise connect to Arduino's reset pin

#define BLACK 0x0000 //human readable common 16-bit values.
#define BLUE 0x001F
#define RED 0xF800
#define GREEN 0x07E0
#define CYAN 0x07FF
#define YELLOW 0xFFE0
#define WHITE 0xFFFF
#define GREY 0xC618
#define DarkGreen 0x03E0
#define MAROON 0x7800
#define Purple 0x780F
#define GreenYellow 0xAFC5
#define Pink 0xA808
File myFile;
Adafruit_TFTLCD tft(LCD_CS, LCD_CD, LCD_WR, LCD_RD, LCD_RESET);  // Init LCD
TSPoint p;

#define BOXSIZE 30
#define PENRADIUS 1
int currentcolor;
const int F3 = A15;
const int F2 = A14;
const int F1 = A13;
int bend = 1;
float Lat;
float Sag;
int Lat_norm;
int Sag_norm;
String fileName="test";
String Data = "",
int counter = 0;
int Start = 0;
int Saving = 0;
int Raw =0;
int SDopen =0;
unsigned long Time, Time0;

void setup(void){
  Serial1.begin(9600);
tft.reset();
tft.begin(0x9341);   // SDFP5408 LCD namework TFT 2.4 with the controller SPFD5408 w library
tft.setRotation(3);   // changed for choice or rotation initial

tft.setCursor (85, 50); // Initial screen
tft.setTextSize (3);
tft.setTextColor(MAROON);
tft.println("UL - BU");
tft.setCursor (65, 85);
tft.println("Technology");
tft.setCursor (65, 150);
tft.println("Optical sensing");
tft.setCursor (80, 200);
tft.setTextSize(1);
tft.setTextColor(BLACK);
pinMode(10, OUTPUT);

if (!SD.begin(10, 11, 12, 13)) {
    tft.println("SD initialization failed!"); // return;
} else
    tft.println("SD initialization done.");
delay(500);

waitOneTouch(); // Wait touch

tft.fillRect(0, 0, 2*BOXSIZE, BOXSIZE, CYAN);
tft.drawRect(0, 0, 2*BOXSIZE, BOXSIZE, WHITE);
tft.fillRect(2*BOXSIZE, 0, 2*BOXSIZE, BOXSIZE, GREY);
tft.fillRect(4.5*BOXSIZE, 0, 2*BOXSIZE, BOXSIZE, BLUE);
tft.fillRect(6.8*BOXSIZE, 0, 2.2*BOXSIZE, BOXSIZE, DarkGreen);
tft.fillRect(9.2*BOXSIZE, 0, 1.5*BOXSIZE, BOXSIZE, Pink);
tft.setCursor(4, 10);
tft.setTextSize(2);
tft.setTextColor(BLACK);
tft.println("BEND");
tft.setCursor(4+2*BOXSIZE, 10);
tft.println("RESP");
tft.setTextColor(WHITE);
tft.setCursor(4+4.5*BOXSIZE, 10);
tft.println("Save");
tft.setCursor(4+6.8*BOXSIZE, 10);
tft.println("Start");
tft.setCursor(4+9.2*BOXSIZE, 10);
tft.println("Raw");
}

void loop()
{
    int f1 = analogRead(F1);
    int f2 = analogRead(F2);
    int f3 = analogRead(F3);
    Lat = (float(f1)-float(f2))/(float(f1)+float(f2));
    Sag = (float(f1)+float(f2)-2*float(f3))/(float(f1)+float(f2)+2*float(f3));
    Lat_norm = int(Lat*float(tft.height()-BOXSIZE)/2 + float(tft.height()+BOXSIZE)/2);
    Sag_norm = int(Sag*float(tft.height()-BOXSIZE)/2 + float(tft.height()+BOXSIZE)/2);
counter ++;
if (counter>tft.width()){
    counter =1;       //graph pixel
    tft.fillRect( 0, BOXSIZE, tft.width(), tft.height()-BOXSIZE, BLACK);
}
if (Raw<1){

    if (bend>0){
        currentcolor = YELLOW;
        DrawPoint(counter,Lat_norm);
    }
    currentcolor = GREEN;
    DrawPoint(counter,Sag_norm);
}
else if (Raw=1){
    currentcolor = RED;
    DrawPoint(counter,f1);
    currentcolor = GreenYellow;
    DrawPoint(counter,f2);
    currentcolor = BLUE;
    DrawPoint(counter,f3);
}

digitalWrite(13, HIGH);     // set digital pin 13 as output
TSPoint p = ts.getPoint();
digitalWrite(13, LOW);
pinMode(XM, OUTPUT);
pinMode(YP, OUTPUT);

if (p.z > MINPRESSURE && p.z < MAXPRESSURE) {
    p.x = map(p.x, TS_MINX, TS_MAXX, 0, tft.width());
    p.y = map(p.y, TS_MINY, TS_MAXY, 0, tft.height());;

    if (p.x < 25){
        if (p.y > 155){
            bend = 1;       //graph pixel
            tft.fillRect( 0, BOXSIZE, tft.width(), tft.height()-BOXSIZE, BLACK);
        } else if (p.y > 115){
            bend = 0;
            counter =1;     //graph pixel
        }
    }
}
```c
  tft.fillRect( 0, BOXSIZEx, tft.width(), tft.height()-BOXSIZEx, BLACK);
} else if (p.y > 64){
  keyBoard();
  Saving = 1;
} else if (p.y > 10){

if(Start<1){
  tft.fillRect(6.8*BOXSIZEx, 0, 2.2*BOXSIZEx, BOXSIZEx, RED);
  tft.setCursor (4+6.8*BOXSIZEx, 10);
  tft.println(" End");
  Time0 = millis();
  Start = 1;
} else{
  tft.fillRect(6.8*BOXSIZEx, 0, 2.2*BOXSIZEx, BOXSIZEx, DarkGreen);
  tft.setCursor (4+6.8*BOXSIZEx, 10);
  tft.println("Start");
  Start = 0;
  myFile.close();
}
} else if (p.y > -23){
  if(Raw<1){
    Raw = 1;
  } else if (Raw = 1){
    Raw = 0;
  }
}
}
delay(100);

if (Saving){
  String Name = fileName + ".csv";
  myFile = SD.open(Name.c_str(), FILE_WRITE);
  Saving =0;
}

if (myFile){
  tft.fillRect(4.5*BOXSIZEx, 0, 2*BOXSIZEx, BOXSIZEx, Purple);
  tft.setCursor (4+4.5*BOXSIZEx, 10);
  tft.println("Rec");
} else {
  tft.fillRect(4.5*BOXSIZEx, 0, 2*BOXSIZEx, BOXSIZEx, BLUE);
}```
tft.setCursor (4+4.5*BOXSIZE, 10);
tft.println("Save");
}

if (Start){
    Data = String(millis() - Time0) + "," + String(Lat) + "," + String(Sag);
    myFile.println(Data);
    Serial1.print(millis() - Time0);
    Serial1.print(",");
    Serial1.print(Lat);
    Serial1.print(",");
    Serial1.print(Sag);
}

if (bend < 1){
    if (millis() - Time0 > 30000)
        Start = 0;
    myFile.close();
    tft.fillRect(6.8*BOXSIZE, 0, 2.2*BOXSIZE, BOXSIZE, DarkGreen);
    tft.setCursor (4+6.8*BOXSIZE, 10);
    tft.println("Start");
}

tSPoint waitOneTouch() {
    TSPoint p;
    do {
        p= ts.getPoint();
        pinMode(XM, OUTPUT); //Pins configures again for TFT control
        pinMode(YP, OUTPUT);
    } while((p.z < MINPRESSURE )|| (p.z > MAXPRESSURE));
    return p;
}

void DrawPoint(int x, int y){
    tft.fillCircle(x, y, PENRADIUS, currentcolor);
}

void keyBoard(){
    fileName = "";
}
char Letters[39] = {'A','B','C','D','E','F','G','H','T','J','K','L','M','N','O','P','Q','R','S','T','U','V','W','X','Y','Z','0','1','2','3','4','5','6','7','8','9','!','@','#'};

int xo = 25;
int yo = 70;
int x,y;
float sx = 183/260;
float sy = (260-55)/120;
float ts_xo = 183 -5*sx;
float ts_yo = 53 + sy*20;
float dx = 20*sx;
float dy = 30*sx;
int Status = 1;
int kx,ky;
int Index=0;
tft.fillRect( 20, 50, 260, 120, BLACK);
tft.drawRect( 20, 50, 260, 120, WHITE);

tft.setTextSize (2);
for (int h=0; h<13; h++){
  for (int j=0; j<3; j++){
    x = xo + h*20;
    y = yo + j*30;
    tft.setCursor (x, y);
    tft.println(Letters[(13)*j + h]);
  }
}

while (Status){
  digitalWrite(13, HIGH);
  TSPoint p = ts.getPoint();
  digitalWrite(13, LOW);
  pinMode(XM, OUTPUT);
  pinMode(YP, OUTPUT);
  if (p.z > MINPRESSURE && p.z < MAXPRESSURE) {
    p.x = map(p.x, TS_MINX, TS_MAXX, 0, tft.width());
    p.y = map(p.y, TS_MINY, TS_MAXY, 0, tft.height());
    tft.fillRect( 50, 180, 100, 60, BLACK);
  }
  if (p.y < 183 & p.y>1 & p.x > 70 & p.x<177){
    kx = int((p.x- 70)/35);
    ky = int((180-p.y )/14);
  }
}
Index = kx*13 + ky;
if (Index < 37) {
    fileName = fileName+String(Letters[Index]);
    counter =0;
}
else if (Index < 38) {
    fileName.remove(0);
    counter =0;
}
tft.setCursor (50,180);
tft.println(fileName);
}
if (Index>37) {
    Status = 0;
    tft.fillRect( 0, BOXSIZE, tft.width(), tft.height()-BOXSIZE, BLACK);
}
delay(70);
}
The MATLAB function ‘sign’ implement the sign function or signum function that extracts the sign of a real number.

\[ Y = \text{sign}(x) \] returns an array \( Y \) the same size as \( x \), where each element of \( Y \) is:

- 1 if the corresponding element of \( x \) is greater than 0.
- 0 if the corresponding element of \( x \) equals 0.
- \(-1\) if the corresponding element of \( x \) is less than 0.
- \(x./\text{abs}(x)\) if \( x \) is complex.

The sign function is utilized to extracts the sign of the respiratory data due to similarity of the breathing signal where the corresponding element of the respiratory signal \( x \) that greater than 0 represents the inhalation period while the respiratory signal that is less than 0 represents the exhalation period. Thus, the breathing period can be calculated from the respiratory signal by identifying the returned \( Y \) of the sign function.
APPENDIX D: RESEARCH ETHICS APPLICATION FORM,
INFORMATION SHEET AND CONSENT FORM

Faculty of Science and Engineering Ethics Committee
Expedited Form for
research involving human participants

1: Applicants Details

<table>
<thead>
<tr>
<th>Field</th>
<th>Details</th>
</tr>
</thead>
<tbody>
<tr>
<td>Principal Investigator name (ie supervisor):</td>
<td>Elfed Lewis</td>
</tr>
<tr>
<td>Principal Investigator email</td>
<td><a href="mailto:Elfed.Lewis@ul.ie">Elfed.Lewis@ul.ie</a></td>
</tr>
<tr>
<td>Student name</td>
<td>Wern Kam</td>
</tr>
<tr>
<td>ID number</td>
<td>15272039</td>
</tr>
<tr>
<td>Email address</td>
<td><a href="mailto:Wern.Kam@ul.ie">Wern.Kam@ul.ie</a></td>
</tr>
<tr>
<td>Programme of study</td>
<td>PhD ECE</td>
</tr>
<tr>
<td>FYP, MSc or PhD Dissertation</td>
<td>For Journal work</td>
</tr>
<tr>
<td>Working title of study</td>
<td>Non-Invasive Measurement of lower back bending angle and breathing</td>
</tr>
<tr>
<td>Period for which approval is sought</td>
<td>Start Date: Date of Approval</td>
</tr>
<tr>
<td></td>
<td>End date: 30/6/18</td>
</tr>
</tbody>
</table>

2. Human Participants

Does the research proposal involve:

- Working with participants over 65 years of age? [Yes ☐ No ☒]
- Any person under the age of 18? [Yes ☐ No ☒]
- Adult patients? [Yes ☐ No ☒]
- Adults with psychological impairments? [Yes ☐ No ☒]
- Adults with learning difficulties? [Yes ☐ No ☒]
- Relatives of ill people (e.g. parents of sick children) [Yes ☐ No ☒]
- Adults under the protection/control/influence of others (e.g. in care/prison)? [Yes ☐ No ☒]
- People who may only have a basic knowledge of English? [Yes ☐ No ☒]
- Hospital or GP patients (or HSE members of staff) recruited in medical facility [Yes ☐ No ☒]

3. Subject Matter

Does the research proposal involve:

- Sensitive personal issues? (e.g. suicide, bereavement, gender identity, sexuality, fertility, abortion, gambling)? [Yes ☐ No ☒]
- Illegal activities, illicit drug taking, substance abuse or the self-reporting of criminal behaviour? [Yes ☐ No ☒]
- Any act that might diminish self-respect or cause shame, embarrassment or regret? [Yes ☐ No ☒]
- Research into politically and/or racially/ethnically and/or commercially sensitive areas? [Yes ☐ No ☒]

4. Procedures

Does the research proposal involve:

- Use of personal records without consent? [Yes ☐ No ☒]
- Deception of participants? [Yes ☐ No ☒]
- The offer of large inducements to participate? [Yes ☐ No ☒]
- Audio or visual recording without consent? [Yes ☐ No ☒]
- Invasive physical interventions or treatments? [Yes ☐ No ☒]
• Research that might put researchers or participants at risk? Yes ☐ No ☒
• Storage of results data for less than 7 years? Yes ☐ No ☒

If you have answered Yes to any of these questions in sections 2 to 4 above, you will need to fill in the S&E full application form and submit to the Faculty Ethics Committee for review. However, if the research is to be conducted during or after/associated with School Placement, and within the Department of Education subject syllabus outline, and provided the student has the permission of the class teacher and the school principal and that parent/guardians consent to participation, this expedited form can also be used. Please note that if the Faculty Ethics Committee deems it necessary you may be asked to fill in the full application form.

Please note that only 1 hard copy of the FREC form is required for the Faculty Ethics Committee. You can get more information and download the forms needed at this address: [http://scieng.ul.ie/research/research-ethics/](http://scieng.ul.ie/research/research-ethics/)

**NB:** If you answered Yes to the last bullet point in section 2 then you will need to apply to the local HSE ethics committee not the FREC.

If you have answered No to all of these questions, please answer the following questions in sections 5.

---

### 5 Research Project Information

**5a Give a description of the research.**

One main health problem associated with the poor condition of lumbar spine is lower back pain (LBP). There is a growing need to accurately measure the bending angle of the spine for patients undergoing physiotherapy for LBP in a low cost and effective manner within the Physiotherapy clinic setting. Hence, in the proposed investigation it is intended to utilise the existing optical fibre sensor system to accurately measure the angle bending of lower back movements.

On the other hand, the same sensor will be utilized for respiratory monitoring purpose. The sensor is used to detect the breathing rate from the movement of chest during respiration process. Respiratory signal obtained able to provide important indications of the patient’s health condition. This compact and small devices will allow to overcome the comfort problems and it is convenient to wear regardless variable size of patients to measure either back bending or breathing signal.

For the bending test, volunteer subjects is asked to stand straight and relax with sensor attached on the lower back of body. Subject is asked to conduct the bending task which is bending to the front, extension to the back, bending to the left and right positions. During the bending procedure, image of sensor is captured and signal is recorded to study the bending angle. For the breathing test, the same sensor is attached on the upper body position and subject is asked to breathe normally for 1 minutes. Signal from sensor is recorded for further analysis. During the testing of sensor, a second person will assist on putting on the sensor and collect data of measurements.

**5b Will the participants be recorded?** Yes ☒ No ☐

If Yes will the recordings be Video ☒ Audio ☐

Why is recording required?

---

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For Image processing purposes, where further data is required to extract from the video recorded. No image of faces will be recorded at any time. Only the video of sensor will be recorded to analyse how far the sensor will bend during the test.

5c Will a prototype be developed? Yes ☐ No ☒

If Yes what format will this prototype take?

5d How many participants will be involved?

Twenty

5e How do you plan to gain access to /contact/approach potential participants?

Colleague and friends

5f What are the criteria for including/excluding individuals from the study?

Heathy adults with no current back pain issue.

5g Have arrangements been made to accommodate individuals who do not wish to participate in the research? (NB This mainly relates to research taking place in a classroom setting)

Yes ☐ No ☐ N/A ☒

If Yes Please state what these arrangements are.

5h Can you identify any particular vulnerability of your participants other than those mentioned in section 2?

No

5i Where will the study take place?

In room E2051, optical research lab, main building, UL or Room Hs2-009a , Health science building, UL.

5j What arrangements have you made for anonymity and confidentiality?

Subjects 1, 2, 3, 4 and 5, name will not be used.

5k What are the safety issues (if any) arising from this study, and how will you deal with them?

No safety issues.
5f How do you propose to store the information once the project is completed? Will the file/computer be password protected?

Data will be stored on my supervisor’s password protected PC.

Where will the information be stored (room number):

D2032, main building, UL

5m Insurance Cover

Insurance cover is required for all research carried out by UL employees. Principal Investigators/Supervisors should carefully view the University’s ‘Guidelines on Insurance Cover for Research’ document and the University’s Insurance cover to ascertain if their proposed research is covered. These documents are available at www.ul.ie/insurance.

Where any query arises about whether or not proposed research is covered by insurance, the Principal Investigator/Supervisor must contact the University’s Insurance Administrator at cliona.donnellan@ul.ie to confirm that the required level of insurance cover is in place.

Please indicate by way of signature that the research project is covered by UL’s insurance policies:

PI/Supervisor signature: ________________________________

5n Please attach the relevant information documents and complete the following checklist to indicate which documents are included with application

<table>
<thead>
<tr>
<th>Document</th>
<th>Yes ☒</th>
<th>No ☐</th>
</tr>
</thead>
<tbody>
<tr>
<td>Participant Information Sheet</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Participant Informed Consent Form</td>
<td>Yes ☒</td>
<td>No ☐</td>
</tr>
<tr>
<td>Parent/Guardian Information Sheet</td>
<td>Yes ☒</td>
<td>No ☐</td>
</tr>
<tr>
<td>Parent/Guardian Informed Consent Form</td>
<td>Yes ☒</td>
<td>No ☐</td>
</tr>
<tr>
<td>School Principal Information Sheet</td>
<td>Yes ☒</td>
<td>No ☐</td>
</tr>
<tr>
<td>School Principal Informed Consent Form</td>
<td>Yes ☒</td>
<td>No ☐</td>
</tr>
<tr>
<td>Teacher Information Sheet</td>
<td>Yes ☒</td>
<td>No ☐</td>
</tr>
<tr>
<td>Teacher Consent Form</td>
<td>Yes ☒</td>
<td>No ☐</td>
</tr>
<tr>
<td>Child Protection Form (must be included if dealing with &lt;18 year olds)</td>
<td>Yes ☒</td>
<td>No ☐</td>
</tr>
<tr>
<td>Questionnaire &amp; Explanatory Cover Letter</td>
<td>Yes ☒</td>
<td>No ☐</td>
</tr>
<tr>
<td>Interview/Survey Questions</td>
<td>Yes ☒</td>
<td>No ☐</td>
</tr>
<tr>
<td>Recruitment letters/Advertisements/Emails, etc.</td>
<td>Yes ☒</td>
<td>No ☐</td>
</tr>
</tbody>
</table>

6. Declaration

The information in this form is accurate to the best of my knowledge and belief and I take full responsibility for it. I undertake to abide by the guidelines outlined in the UL Research Ethics Committee guidelines http://www.ul.ie/researchethics/

I undertake to inform S&EEC of any changes to the study from those detailed in this application.
<table>
<thead>
<tr>
<th>Student: Wern Kam</th>
<th>Name: Wern Kam</th>
<th>Date:</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Signature:</td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Principal Investigator*:</th>
<th>Name: Elfed Lewis</th>
<th>Date:</th>
</tr>
</thead>
<tbody>
<tr>
<td>Elfed Lewis</td>
<td>Signature:</td>
<td></td>
</tr>
</tbody>
</table>

* In the case where the principal investigator is not a permanent employee of the University, the relevant head of department must sign this declaration in their place.

You should return this form with signatures to the S&E Ethics Committee c/o Faculty Office, Faculty of Science & Engineering, University of Limerick. In addition, a single pdf file containing the completed form and additional information (e.g. participant information sheet) should be emailed to SciEngEthics@ul.ie This form must be submitted and approval granted before the study begins.
Dear,

I am Wern Kam and I am currently pursuing PhD in Electronics and Computer Engineering from the University of Limerick, Ireland. The topic of my research project is ‘Collection of lower back bending angle and breathing dataset’, under the supervision of Prof Elfed Lewis. The study is concerned with the measurement of lower back bending angles and breathing signal utilizing optical fibre sensor system.

The study will involve participants to stand and relax while conducting the following tasks: bending to the front, bending to the back, lateral left bending and lateral right bending.

- The sensor will be attached to the skin on the lower back and the whole procedure will take up to 45 minutes.
- An image of the sensor attached on the body will be recorded during the bending process for further analysis. No images of faces will be recorded at any time.
- For the breathing test, the sensor will be attached to the skin close to the lungs on the upper body and you will be asked to relax and breathe normally for 2 minutes. Respiratory signal will be recorded on a PC for further analysis. The data will be analysed and used as support for a journal paper submission.

To participate in the study, you must be a healthy adult between the ages of 18 and 65. You have the right to not participate or withdraw at any time. Your participation in the project will remain anonymous. There is no safety issues involved with the project.

If you have further questions regarding this research please feel free to get in touch with either myself or my supervisor using the email addresses listed below. If you have concerns about this study and wish to contact someone independent, you may contact: The Chair, Faculty of Science & Engineering Research Ethics Committee, University of Limerick, Limerick. Tel: 061 202802

If you agree to take part in the study, please sign the consent form overleaf.

Yours sincerely,

<table>
<thead>
<tr>
<th>Wern Kam</th>
<th>Elfed Lewis, ECE, +353-(0)61-202968</th>
</tr>
</thead>
<tbody>
<tr>
<td><a href="mailto:Wern.Kam@ul.ie">Wern.Kam@ul.ie</a></td>
<td><a href="mailto:elfed.lewis@ul.ie">elfed.lewis@ul.ie</a></td>
</tr>
</tbody>
</table>

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CONSENT FORM

Consent Section:

I declare that I am willing to take part in research for the project entitled “Non-Invasive Measurement of lower back bending angle and breathing”.

- I declare that I have been fully briefed on the nature of this study and my role in it and have been given the opportunity to ask questions before agreeing to participate.
- The nature of my participation has been explained to me and I have full knowledge of how the information collected will be used.
- I fully understand that there is no obligation on me to participate in this study.
- I fully understand that I am free to withdraw my participation at any time without having to explain or give a reason.
- I am also entitled to full confidentiality in terms of my participation and personal details.
- I declare that I am a healthy adult between the age of 18 and 65.
- I am aware that the image of sensor will be recorded but no images of faces will be recorded at any time.

______________________________________         __________________________
Signature of participant                                               Date
The responses to this survey will assist us in determining the validity and reliability of the sensor. The information will be maintained in strict confidence.

Name:
Age:
Occupation:
Sex:
Height:
Weight:
BMI:
Email Address:

1) Do you have any back pain history?

☐ YES    ☐ NO

*If Yes, please list the condition:

2) On a scale of 1-5 with 0 = ‘no discomfort’ and 5 = ‘extreme discomfort’, how would you describe the comfortability of the device?

☐ 1    ☐ 2    ☐ 3    ☐ 4    ☐ 5