

The University of
Nottingham

Optical Fibre Sensor for Measuring Endotracheal Tube Cuff Contact Pressure on Tracheal Wall

By
Yasmine Saoud

Thesis submitted to
The University of Nottingham
for the degree of
Doctor of Philosophy

June 2024

Abstract

An endotracheal tube (ETT) is a tube placed through the mouth and into the trachea by an anaesthetist. It provides airway management for unconscious patients by facilitating ventilation and oxygenation. The inflatable cuff located towards the bottom of the ETT is used to provide a seal to prevent liquid and bacteria in the mouth from entering the lungs; it is also used to prevent backflow of air from the ventilator escaping through the mouth. The main problems associated with ETTs are caused by the inflatable cuff. The ETT cuff is generally inflated to an intracuff pressure of 20-30 cmH₂O as per hospital guidelines. Over inflation of the cuff exerts high pressures (of over 30 cmH₂O) onto the tracheal wall which obstructs the tracheal capillaries, damaging the mucosa and muscle tissue, causing complications like tracheal stenosis, tracheomalacia, and even in some cases tracheoesophageal fistulas. Under inflation of the cuff can cause ventilator associated pneumonia (VAP). VAP not only increases hospital stay but it costs the NHS £10,000 – £20,000 per patient for treatment. It also increases patient morbidity, as 10-20% of patients that are mechanically ventilated go on to develop VAP and increases patient mortality- as between 3,000 and 6,000 people die from VAP every year.

Optical fibre sensors are well suited for use in biological environments as they are biocompatible, extremely small in size, can withstand harsh conditions, and are not affected by electromagnetic interference. Additionally, research shows that Fibre Bragg Grating (FBG) based pressure sensors can be modified to meet the needs of the application. Although initial research into using FBG pressure sensors show great potential for ETT cuff contact pressure sensing, these sensors were calibrated using weighing scales. Calibrating a sensor on a hard and flat surface for their application in a tube-like soft tissue structure (the trachea) may not be the best calibration approach. In addition to this, the sensors were calibrated using pressures of up to 600 cmH₂O, much higher than those traditionally used in the trachea (20-30 cmH₂O). Since higher pressures were used, it is possible that the sensor cannot sense pressures below a certain value and a better sensitivity may be required for lower pressures. In this thesis, a sensor is optimised for sensing pressures of up to 150 cmH₂O.

The aim of this thesis is to optimise a miniature FBG based pressure sensor from literature for better sensitivity to low pressures, to investigate the validity of this sensor

through repeatability tests, and to develop a new calibration technique to calibrate the sensors more accurately for measuring tracheal wall pressure.

The first part of this thesis investigated the shortcomings of an FBG based pressure sensor for tracheal wall contact pressure measurements from the literature. Different embedding materials for the FBG were tested for better sensitivity to lower pressures. This was achieved by sandwiching the FBG in between two material samples. Alongside this, a Finite Element Analysis (FEA) model was created of the measurement system and a comparison was made to determine if the model could be used to predict experimental changes. Dragon Skin and SYLGARD performed the best in these initial sandwich tests and were shortlisted for further testing.

The second part of this thesis evaluated the performance and repeatability of embedded sensors using the shortlisted materials. These sensors were created using a new fabrication process that ensured the production of good quality sensors. The FEA model was altered to the new sensor dimensions and utilised to determine methods to optimise the sensitivity of the sensor. Sensors were then optimised by increasing the thickness of the sensor as well as by applying epoxy anchors to either side of the embedding material. The coefficient of variation (CV) was calculated to determine sensor repeatability. Results showed the optimised Dragon Skin sensor had a CV of 0.16% indicating good repeatability and a sensitivity of $3.56 \pm 0.0058\text{pm/kPa}$ and the optimised SYLGARD sensor has a sensitivity of $3.75 \pm 0.03\text{pm/kPa}$ with a CV of 0.81%. The optimised sensors then went on for testing in a tracheal phantom.

The last part of this thesis described a new calibration approach (tracheal phantom) to test the optimised embedded sensors. This new method involved the use of a tube with a diameter like that of an adult trachea, a load cell to measure wall contact pressure, and an ETT cuff to simulate the applied pressure as it would be applied in an in vivo trachea. Results showed that the sensor needs further optimisation for use in a trachea and that the use of a double cuff affects sensor readings. Challenges like sensor movement during inflation-deflation cycles, the use of a double cuff, sensor orientation, and tube curvature were identified.

Despite these challenges, the findings of this thesis contribute to the growing body of knowledge on the use of FBG based pressure sensors for tracheal wall pressure measurement. This research underlines the necessity for further research and refinement

of calibration methods and sensor design to ensure the reliability and accuracy of optical sensors for tracheal applications.

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Acknowledgements

I would like to express my heartfelt gratitude to my supervisors, Dr. Ricardo Correia, Prof. Stephen Morgan, Prof. Barrie Hayes-Gill, and Prof. Serhiy Korposh, for their support and guidance throughout my PhD journey. Their invaluable insights, constructive feedback, and patience have been instrumental in shaping my research and my overall development as a researcher.

I am grateful to my family for their unconditional support and continuous cheerleading. Their encouragement and love have been a constant source of strength and motivation during my PhD journey.

I would like to thank my husband Iyad for his love, support, and understanding throughout this journey. His encouragement, belief in me, and patience gave me strength when I thought I had none. He is, and has always been, my hero. I am truly blessed to have such a supportive and loving partner by my side.

I would also like to express my gratitude to my colleagues at the L4 labs and tower building for their support and assistance throughout my PhD journey. I am grateful to have had the opportunity to work with such talented and dedicated individuals.

Lastly, I would like to thank EPSRC for their financial support, which has been crucial in enabling me to pursue my research. Without their support, this journey would not have been possible.

Abbreviations

ADC	Analogue to Digital Converter
AMF	Advanced Manufacturing Facility
ATP	Adenosine Triphosphate
CCM	carbon composite material
CV	Coefficient of Variation
DAQ	Data Acquisition Card
EMFC	Electromagnetic Force Compensation
ETT	Endotracheal Tube
FBG	Fibre Bragg Grating
FEA	Finite Element Analysis
FFT	Fast Fourier Transform
LED	Light Emitting Diode
LPF	Low Pass Filter
MMF	Multimode Fibre
OFS	Optical Fibre Sensor
PCC	Pearson's Correlation Coefficient
PDMS	Polydimethylsiloxane
PLA	Polylactic Acid
PU	Polyurethane
PVC	polyvinyl chloride
RI	Refractive Index
SLED	Superluminescent Light Emitting Diode
SMF	Single Mode Fibre
VAP	Ventilator Associated Pneumonia

Chapter 1 – Introduction

1.1 Introduction

An endotracheal tube (ETT) is a tube placed through the mouth and into the trachea by an anaesthetist. It provides airway management for unconscious patients by facilitating ventilation and oxygenation. Towards the distal end of the ETT is an inflatable cuff used to provide a seal to prevent liquid and bacteria from the mouth (known as aspirate) from entering the lungs; it is also used to prevent backflow of air from the ventilator escaping through the mouth.

The main problems associated with ETTs are caused by this inflatable cuff. Over inflation of the cuff exerts pressures above the recommended 20-30 cmH₂O onto the tracheal wall. This obstructs the tracheal capillaries, affecting circulation within the tracheal tissue and damaging the tracheal mucosa (Ansari *et al.*, 2014, Hamilton and Grap, 2012, Tsaousi *et al.*, 2018). Patients intubated for long periods of time are at risk of developing severe conditions such as tracheal stenosis (the narrowing of the trachea), tracheomalacia (collapsing of the airway), and in some extreme cases tracheoesophageal fistulas (a connection of the trachea to the oesophagus) (Hyzy, 2018).

Under-inflation of the cuff results in an inadequate seal and leaves gaps for aspirate to enter the lungs (Dave *et al.*, 2010, Delorenzo, Shepherd and Jennings, 2017). This causes ventilator associated pneumonia (VAP). VAP not only increases hospital stay but costs the NHS £10,000 – £20,000 per patient for treatment. It also increases patient morbidity, as 10-20% of patients that are mechanically ventilated will go on to develop VAP, and increases patient mortality- as between 3,000 and 6,000 people die from VAP every year in England (NHS England, 2016).

Advances in Fibre Optic research show that their biocompatibility, compact size, and resistance to electromagnetic interference, make them ideal for ETT cuff pressure sensing. Optical fibre sensors (OFS) are small in size, and do not generate electric charge, minimizing the risk of electrocution, which is a critical consideration for in vivo applications and patient safety. Additionally, research shows that Fibre Bragg Grating (FBG) contact pressure sensors of different designs, materials, and sizes can be

modified to meet the needs of the application (Nguyen *et al.* 2022, Hong *et al.*, 2019). Although some research into using FBG pressure sensors for ETT cuff contact pressure sensing does exist, current sensors were calibrated and tested with pressures higher (up to 600 cmH₂O) than those traditionally used in the trachea (20-30 cmH₂O) (Correia *et al.*, 2016, Liu *et al.*, 2018, Correia *et al.*, 2022). These sensors were also calibrated using weighing scales which may not be the best method for calibrating a sensor that will go on for practical use in the trachea (a curved, soft tissue).

The focus of this research is on the development and optimisation of a miniature fibre optic contact pressure sensor with enhanced sensitivity to lower pressures (up to 150 cmH₂O), aided by Finite Element Analysis (FEA) models. Additionally, this study examines the sensor's performance within a tracheal-like environment and an investigation into the validity of the sensor for such applications is carried out.

1.2 Aim

To optimise an FBG based pressure sensor from literature for better sensitivity to low pressures and to develop a new calibration technique to calibrate the sensors more accurately for measuring tracheal wall pressure.

1.3 Objectives

1. Recreate the fibre optic pressure sensor from literature and assess its performance at pressures below 50kPa.
2. Test suitable sensing materials to increase the sensitivity at pressures below 35kPa.
3. Implement a new fabrication method that allows for sensor fabrication repeatability.
4. Create Finite Element Analysis models based on experimental design to predict how different sensing parameters will affect experimental results.
5. Create an optimised sensor using new fabrication method.
6. Test sensor performance on weighing scales with pressures up to 15kPa (~150 cmH₂O).

7. Test sensor in a phantom trachea with ETT cuff inflation to evaluate sensor repeatability in a tube and to better understand sensor performance on curved surfaces.

1.4 Thesis Overview

Chapter 2- Literature Review: This chapter introduces the ETT and the ETT cuff, the consequences to the tracheal wall when the cuff is over-inflated, the consequences of an under-inflated cuff, and the cuff pressures normally used when intubating a patient. Once these have been addressed, a discussion on how cuff pressures change over time is presented. This highlights the need for a contact pressure sensor on the cuff to monitor contact pressures on the tracheal wall.

Different contact pressure sensors used today are then discussed, stating their pros and cons and why other sensing modalities should be considered. Optical fibre sensors are introduced describing their different sensing mechanisms as well as the use of FBGs.

A literature review looking at previous research using FBGs for pressure measurement is presented as well as, more specifically, embedding FBGs in different materials for contact pressure sensing. Finally, literature using these sensors to measure tracheal wall pressures are described.

Chapter 3- Embedding Material Selection Process and ABAQUS Model: This chapter discusses recreating a sensor from the literature for contact pressure measurement on the trachea and mentions its shortcomings.

The chapter then investigates testing different materials to determine an adequate embedding material for the FBG in order for low pressure detection and measurement. These initial tests were carried out by sandwiching the FBG in between two material samples. Alongside this, an FEA model was created of the measurement system and a comparison is made. The materials performing best in these initial tests were shortlisted for further testing.

Chapter 4- Embedded Sensor Testing and ABAQUS Embedded Sensor Model: This chapter demonstrates the next step of testing, embedding the FBG into a polymer patch made from the shortlisted materials to create the embedded sensor. It also introduces a new method of the fabricating sensors that ensures uniform sensing surfaces. An FEA

model was created replicating the dimensions of the new fabricated embedded sensor, the experimental results were compared to the model results and were used to determine methods to optimise the sensitivity of the sensor. The different optimised variations of the sensor are described in this section and went on for tube tests (tracheal phantom tests).

Chapter 5- Embedded Sensor Tube Testing: This chapter introduces the calibration tube setup and its features. It then goes on to test a total of three optimised sensors in this setup to determine how the sensors perform in a tracheal phantom and identify any challenges these sensors may face.

Chapter 6- Conclusions and Future Work: This chapter concludes the findings of this research and highlights ways to improve experimental analysis as well as possible new approaches to contact pressure sensing in a tracheal phantom.

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Chapter 2 – Literature Review

2.1 The Endotracheal Tube

2.1.1 Introduction

An endotracheal tube (ETT), as shown in Figure 2.1, is a tube placed through the mouth, past the glottis, and into the trachea by an anaesthetist. It provides airway management for unconscious patients by facilitating ventilation and oxygenation (Malamed *et al.*, 2015). The glottis serves as a barrier that partitions the mouth and lungs, preventing bacteria and saliva (known as aspirate) from entering the lungs (Esling, 2006). However, when an ETT is inserted into the trachea, the glottis is held open, and is unable to carry out its function of preventing airway aspiration. The inflatable cuff located towards the bottom of the ETT is used to provide a seal to prevent this aspirate from entering the lungs; it is also used to prevent backflow of air from the ventilator escaping through the mouth (Hamilton & Grap, 2012).

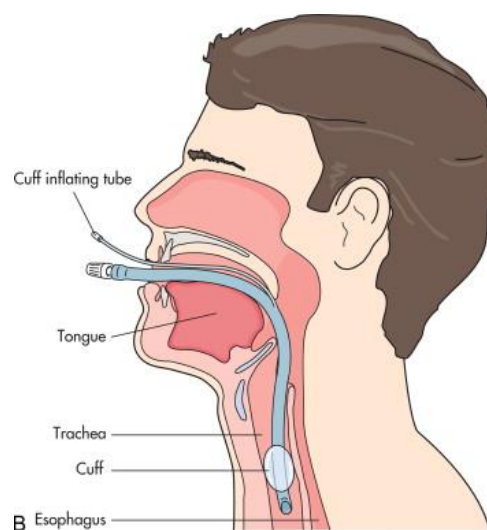


Figure 2.1: ETT inserted into trachea (Monahan, 2007)

2.1.2 Problems Associated with ETTs

The main problems associated with ETTs are caused by the inflatable cuff. Over inflation of the cuff exerts high pressures of more than 30 cmH₂O onto the tracheal wall. This obstructs the tracheal capillaries, damaging the tracheal mucosa and muscle tissue. Figure 2.2 illustrates what happens to the tracheal tissue when the ETT cuff is

over inflated. Patients who are intubated for just a few hours often complain of throat pain after extubation (Ansari *et al.*, 2014). Patients are at risk of developing more severe conditions such as tracheal stenosis (the narrowing of the trachea) and tracheomalacia (collapsing of the airway), if intubated for more than 7 days. Tracheoesophageal fistulas (a connection of the trachea to the oesophagus) can form if intubated at high cuff pressures for more than a month (Hyzy, 2018).

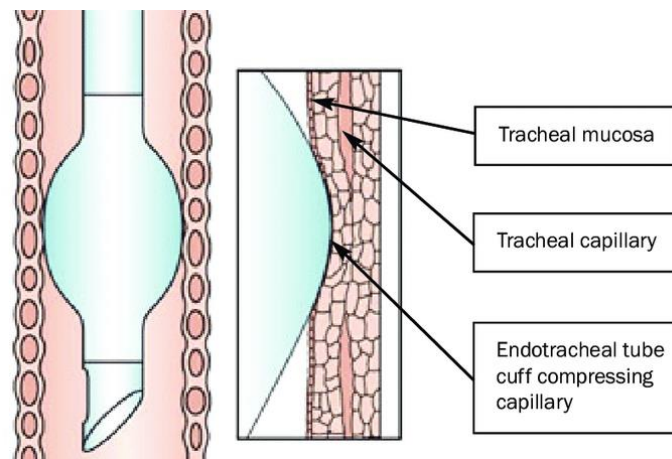


Figure 2.2: Obstruction of tracheal mucosa due to cuff over inflation

Once a patient is intubated, anaesthetists inflate the cuff using a syringe and the intracuff pressure is measured using a manometer. It is common practice that the intracuff pressure is maintained within 20-30 cmH₂O in order to provide an adequate seal to prevent aspiration while also maintaining good blood flow in the mucosa and tracheal muscle (Ansari *et al.*, 2014, Hamilton and Grap, 2012, Tsaousi *et al.*, 2018). Manometers are not always readily available and therefore lead to higher risks of complications occurring (Tsaousi *et al.*, 2018). Additionally, research has shown that intracuff pressure readings on the manometer do not accurately reflect the contact pressure exerted on the tracheal wall (Correia *et al.*, 2016). This discrepancy is likely due to the complex anatomy of the trachea, which features a non-cylindrical geometry and cartilaginous ridges. Furthermore, the variability in the ETT cuff's positioning within the trachea can result in the cuff inflating non-uniformly causing certain areas within the tracheal wall to experience heightened contact pressure compared to others. These variables can cause variations in pressure distribution along the tracheal wall, leading to differences between the pressure measured within the cuff and the actual pressure on the tracheal wall.

High ETT cuff pressures can be influenced by the cuff inflation technique, cuff shape, and the cuff material (Chapter 2.1.4). In addition to this, cuff pressures have also been shown to change during intubation (Chapter 2.1.5).

Another problem associated with ETTs is ventilator associated pneumonia (VAP). This is pneumonia caused by the ETT cuff not providing an adequate seal (when not inflated enough) allowing aspirate from the mouth to enter the lungs (Dave *et al.*, 2010, Delorenzo, Shepherd and Jennings, 2017). VAP not only increases hospital stay but it costs the NHS £10,000 – £20,000 per patient for treatment. It also increases patient morbidity, as 10-20% of patients that are mechanically ventilated will go on to develop VAP, and increases patient mortality- as between 3,000 and 6,000 people die from VAP every year (NHS England, 2016).

2.1.3 Properties of the Trachea

2.1.3.1 Anatomical Properties

The trachea is made up of 20 cartilaginous rings and is about 10cm long. The posterior part of each ring is made of muscle and connective tissue (McCullagh *et al.*, 2022). The normal transverse internal diameter of the trachea ranges between 1.5cm and 2.5cm in males and 1cm to 2.1cm in females (Breatnach *et al.*, 1984). Figure 2.3 shows a transverse cross-section of the trachea.

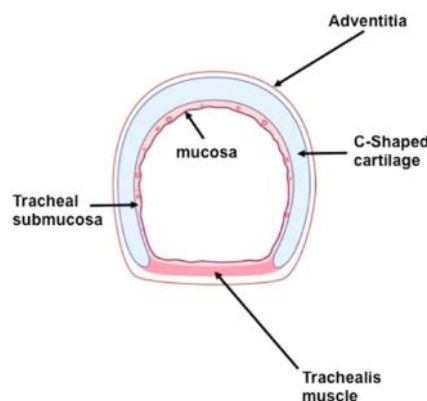


Figure 2.3: Transverse cross-section of trachea (Shroff *et al.*, 2016).

Due to the cartilage being c-shaped, the cross-section of the trachea isn't a perfect circle. This means that when the ETT cuff inflates, the cartilage restricts the expansion to the muscular back of the trachea, stretching it. It is when the tracheal muscle has been left to stretch for long periods of time that complications (like tracheal stenosis,

tracheomalacia, and tracheoesophageal fistulas) occur. It is important to be able to tell when the cuff comes into contact with this so that over-inflation does not occur.

2.1.3.2 Mechanical Properties

To understand why such effects occur on the trachea over time, it is important to look into the mechanical properties of the trachea.

As the trachea is soft tissue, it exhibits viscoelastic behaviour (Eskandari *et al.*, 2018). This means that the trachea experiences anisotropy and creep when a load is applied. Anisotropic behaviour means that the stress-strain curves of the trachea when a load is applied are different in different directions. Eskandari *et al.* examined the difference in tensile stress strain curves of a porcine trachea when the trachea is pulled both vertically and horizontally. Figure 2.4 shows their results.

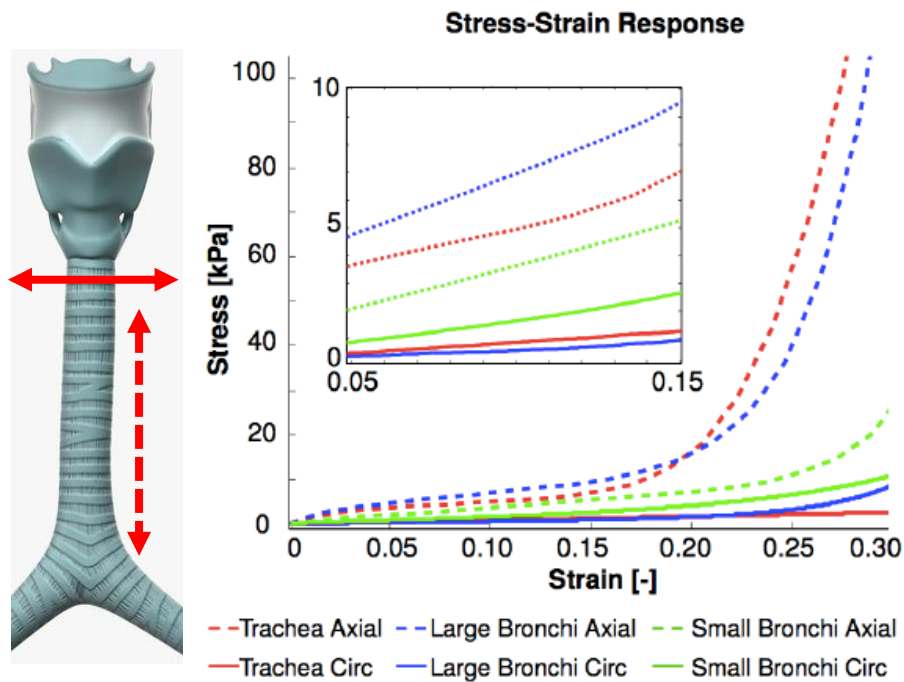


Figure 2.4: The red dotted line shows the stress-strain curves of the trachea when pulled vertically. The red solid line shows the stress-strain curves of trachea when pulled horizontally. Insert shows an enlarged portion of the graph between 0 and 10kPa of stress so that is easier to see the different strain responses (Eskandari *et al.*, 2018).

It is clear from Figure 2.4 that for tracheal elongation along the horizontal axis (the direction the ETT expands in the trachea) a small stress applied results in a very large strain (deformation). Whereas for the same small stress along the vertical axis, the strain exhibited is not as large.

In addition to this, as mentioned earlier in this section, soft tissues undergo creep. This means as a constant stress is applied to the trachea, the strain experienced increases

over time. This could mean that the longer the ETT is inflated, provided the load applied to the trachea is the same, the more strain or deformation the trachea exhibits (Bhattacharya & Mishra, 2015).

2.1.4 Factors That Influence Higher Cuff Pressures and Aspiration Volumes

High ETT cuff pressures can arise from the cuff inflation technique (Tsaousi *et al.*, 2018); the cuff shape can result in higher aspiration volumes (In, Shim and Chung, 2014), and the cuff material can bring about both higher aspiration volumes and cuff pressures (Fernandez *et al.*, 2012).

2.1.4.1 Cuff Inflation Technique

There are multiple methods to inflate the ETT cuff to achieve an adequate seal. Methods include: Palpation, Air Return, Minimum Volume, and Minimum Leak. The Palpation method involves feeling the throat when inflating the ETT cuff and stopping inflation once the ETT seal can be felt. The Air Return method involves inflating the cuff with a syringe of air and allowing the piston in the syringe to return excess air. The Minimum Volume method entails placing a stethoscope on the throat to listen while slowly inflating the cuff as air is ventilating into the lungs using a ventilator. Once there is no sound of air backflow (from the ventilated air to the lungs), then the cuff has reached an adequate seal. Lastly, the Minimum Leak method involves following the same guidelines as the Minimum Volume method except when no sound of the back flow of air can be heard the cuff is then deflated slightly until a small squeak can be heard. This ensures that the seal is not too tight causing high pressures on the trachea (Ansari *et al.*, 2014).

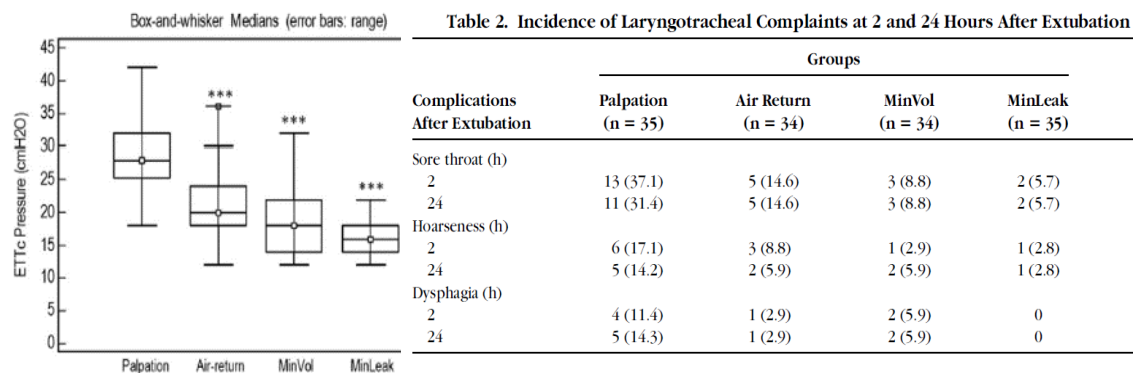


Figure 2.5: Comparison of the different cuff inflation techniques vs patients' complaints (Tsaousi *et al.*, 2018).

Research shows that cuff inflation techniques using a stethoscope to monitor inflation create a seal with lower pressures and therefore received fewer complaints from patients (Tsaousi *et al.*, 2018, Mali, Solanki and Deshpande, 2017, Borhazowal *et al.*, 2017). Figure 2.5 was taken from a study that compared these different inflation techniques. It can be seen that the Minimum Leak method resulted in the lowest ETT cuff pressures and also the least number of complaints from patients.

The current method of cuff inflation used in hospitals is the Minimum Volume method (NHS University Hospitals, 2022, NHS Guidelines, 2014).

2.1.4.2 Cuff Shape

The two most frequently used ETT cuff shapes are tapered and cylindrical. Research shows that tapered cuffs reduce aspiration compared to cylindrical cuffs (In, Shim and Chung, 2014, Dave *et al.*, 2010, Khashaba *et al.*, 2019, Chang *et al.*, 2017). This is due to the tapered shape having various diameters providing a good fit in the trachea while the cylindrical shaped cuff has a fixed diameter. If the trachea's diameter is smaller than the cuff's, small channels can form in the cuff from folding, causing aspiration into the lungs. Figure 2.6 illustrates this phenomenon.

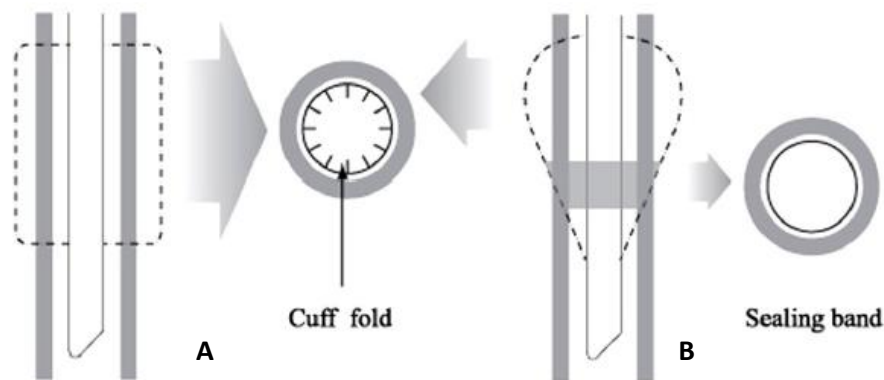


Figure 2.6: A: Cylindrical cuff, B: Tapered Cuff (In, Shim and Chung, 2014)

Chang *et al.*'s study showed that the tapered cuff also resulted in reduced throat pain (Chang *et al.*, 2017). The cuffs conventionally used in hospitals today are the cylindrical cuff (Seol *et al.*, 2022).

2.1.4.3 Cuff Material

Like the ETT cuff shape, there are two commonly used cuff materials; polyurethane (PU) and Polyvinyl chloride (PVC). Research shows that PU cuffs significantly reduce aspiration compared to PVC cuffs regardless of the cuff shape (Bassi *et al.*, 2015, Miller

et al., 2011, Chenelle *et al.*, 2017, Lau, Lam and Yan, 2014, Dullenkopf, Gerber and Weiss, 2003, Dave *et al.*, 2010, Blot, Rello and Koulenti, 2016, Fernandez, Levine and Restrepo, 2012). This is thought to be due to the PU membrane thickness being 7µm, whereas the PVC membrane thickness is more than 50µm (Miller *et al.*, 2011). This means that when the cuff material folds and channels form, the PU channels are much narrower than the PVC channels, reducing the amount of aspiration (Chenelle *et al.*, 2017). It is also said that with PU cuffs, adequate sealing can be achieved with pressures of 8-12 cmH₂O which is significantly lower than the 20-30 cmH₂O needed to seal PVC cuffs (Fernandez *et al.*, 2012).

It is worth mentioning that a tapered PU cuff tends to outperform all other ETT cuffs when it comes to reducing aspiration. Whereas a cylindrical PU cuff tends to perform better than a tapered PVC cuff (Dave *et al.*, 2010, Dullenkopf, Gerber and Weiss, 2003, Lau, Lam and Yan, 2014). This shows that although tapered cuffs are performing better than cylindrical cuffs in reducing aspiration, the material of the cuff seems to be of more significance than the shape of the cuff.

The cuff material conventionally used in hospitals is PVC (Suhas *et al.*, 2016). Although PU cuffs have been shown to provide a seal with lower applied pressure and reduce aspiration, the price of a PU cuffed ETT is about \$3-4. The price of the PVC cuffed ETT is \$1 (Fernandez *et al.*, 2012). Therefore, providing hospitals with PU cuffed ETT to replace the PVC cuffed ones would cost hospitals a considerable amount. However, it can be argued that by paying the extra cost for the PU cuffed ETTs it could save hospitals in the long run as the cases of VAP may be significantly reduced. This could save them much more than they save by buying PVC cuffs.

2.1.5 Factors Effecting ETT Cuff Pressure Changes Over Time

Intracuff pressure can be monitored while patients are intubated by using a manometer. Research shows that intracuff pressures tend to increase and/or decrease during intubation. There are many factors that can cause this to happen. These include: laparoscopic surgery and the use of nitrous oxide in anaesthesia gas, patient movement, and altitude.

2.1.5.1 Laparoscopic surgery and nitrous oxide in anaesthesia gas

Nitrous Oxide (N₂O) is known to increase the cuff pressure over time due to it diffusing into the cuff. Mogal *et al.* studied the cuff pressure increase from baseline over time for laparoscopic surgeries using either N₂O (Group N) or air (Group A) in the anaesthesia gas mixture (Mogal *et al.*, 2018). Figure 2.7 shows their results.

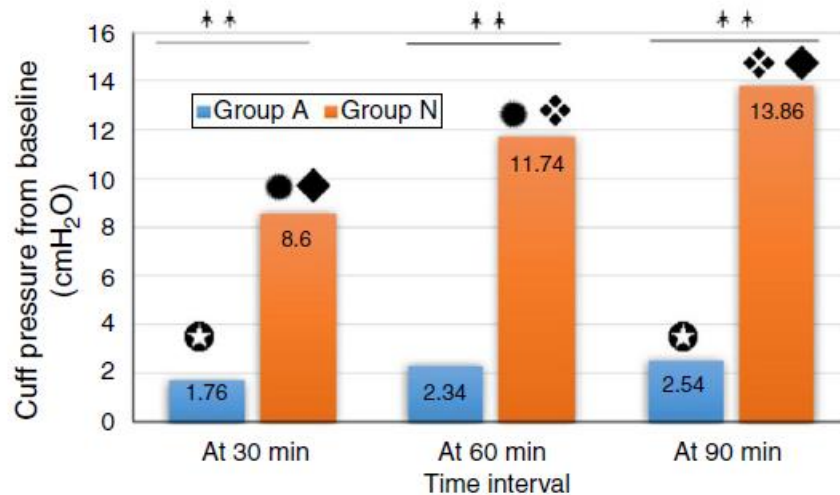


Figure 2.7: Bar chart showing the pressure increase for patients using N₂O (orange) and air (blue) during laparoscopic surgery. Similar symbols represent significant difference ($p < 0.05$) between two time points in a group (Mogal *et al.*, 2018).

From Figure 2.7 it is clear that N₂O causes a large pressure increase inside the cuff from baseline (baseline pressure is the pressure used to create a seal, between 20-30cmH₂O) with a pressure increase of 8.6 cmH₂O after 30 minutes of intubation and a pressure increase of 13.8 cmH₂O after 90 minutes of intubation. However, the patients receiving just air also exhibited an increase in intracuff pressure (Saracoglu *et al.*, 2018). This increase in pressure means that the cuff is now overinflated and could cause damage to the trachea if left for long periods of time. It is believed that this pressure increase is due to the surgical procedure itself. In laparoscopic surgery gases are used to expand the abdominal region. The gases used to do this may diffuse into the cuff causing the pressure increase (Mogal *et al.*, 2018).

2.1.5.2 Patient Movement

Moving the body while a patient is intubated can result in dislodgement of the ETT cuff and therefore a change in the cuff pressure (Saracoglu *et al.*, 2018, Minonishi *et al.*, 2013). Minonishi *et al.* studied the change in intracuff pressure as patients were turned in bed. Figure 2.8 shows their results.

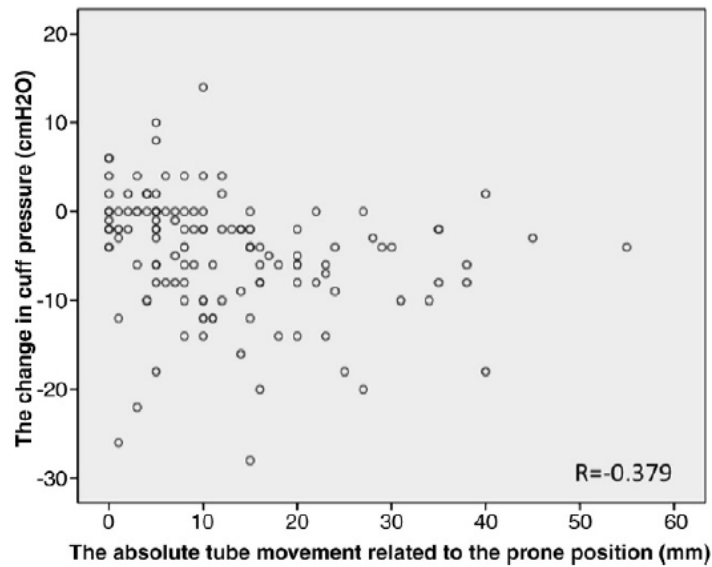


Figure 2.8: The change in intracuff pressure from baseline (0) and the tube movement as patients turned in bed (Minonishi *et al.*, 2013). the baseline, 0, indicates the pressure used to create a seal, any pressure above this is means the cuff is overinflated and can cause damage to the trachea, any pressure below this means the seal is not formed and underinflation occurs.

From Figure 2.8, it seems as though the change in intracuff pressure mostly decreases due to patient movement. This means that when patients move, they are at risk of the ETT cuff seal being broken and a passageway for aspirate to enter the lungs can form. For the patients that saw an increase in intracuff pressure when moved positions, they could be at risk of tracheal damage.

2.1.5.3 Altitude

In emergency cases unconscious patients are moved from place to place in helicopters. Many of these patients are relocated while intubated. It is known that as altitude increases atmospheric pressure decreases. This causes the pressure inside the cuff to be larger than the atmospheric pressure outside the cuff, causing the cuff to expand. The trachea limits the expansion of the cuff resulting in larger contact pressure. This will in turn result in larger intracuff pressures as the trachea is now applying a force onto the cuff. Henning *et al.* compared the pressures of air filled ETT cuffs at ground level and at 3000 ft altitude in 10 intubated patients. They found that the mean cuff pressure rise was 23 cmH₂O. They concluded that this pressure increase could lead to tracheal injury (Henning *et al.*, 2004).

2.1.6 Conclusion

The mechanical properties of the trachea, along with the many factors influencing intracuff pressures, demonstrate the complexities of airway management. With abundant evidence of intracuff pressures changing over time, there is a need for a feedback loop ETT system. The current method used to measure intracuff pressure, the manometer, is insufficient as different factors affect contact pressure. Inflating using the manometer will maintain a certain volume of air in the cuff, however if the patient moves, or the patient is moved from one place to another in a helicopter, the contact pressure this volume of air exerts on the trachea changes. More is needed to control the intracuff pressure and prevent the consequences of both high and low cuff pressures. A feedback ETT system can automatically alter the intracuff pressures to ensure contact pressures stay minimal while also achieving an adequate seal. For the feedback loop system to operate effectively, continuous monitoring of the contact pressure exerted on the tracheal wall is required. Additionally, the sensor must have the following target specification:

- be made of biocompatible materials
- response time: immediate (minimum 4 Hz)
- stabilisation time: immediate (< 1 second)
- be able to read pressures up to 10kPa or higher (to account for overinflation)
- exhibit minimal/no hysteresis
- show good repeatability (coefficient of variation $< 5\%$)
- be small in size to fit on ETT cuff (ideally smaller than 1cm x 1cm and a thickness no greater than 2.5mm).
- have a resolution in the range of 0.1kPa - 0.01kPa as 0.1kPa is a significant change in pressure that the sensor should detect and pressures below 0.01kPa are not significant. A sensor with a resolution below 0.01kPa are at risk of picking up minor disturbances that may not be related to the actual pressure being measured, like vibrations.

As the contact pressure has a direct influence on capillary blood flow and patient safety it is important to prioritise the measurement of contact pressure. This way, clinicians will be able to gain a more accurate understanding of the actual forces exerted on the tracheal wall, allowing for precise adjustments to optimise patient safety and outcomes.

This research discusses the creation of a miniature fibre optic pressure sensor to be placed on the ETT cuff to measure the contact pressure on the tracheal wall. This sensor will act as an integral part of an automated feedback loop system to control and alter the contact pressure experienced on the tracheal wall.

2.2 Contact Pressure Sensors Used Today

2.2.1 Introduction

There are numerous types of pressure sensors used today. As this research looks into contact pressure sensing specifically, the pressure sensors of interest convert a physical change caused by an applied force into an electrical signal. The resultant electrical signal is proportional to the applied pressure. Contact pressure sensing is done by utilising force sensors to calculate the force exerted on a specific area. These force and area values can then be used to calculate the contact pressure. These sensors can come in the form of piezoresistive pressure sensors, capacitive pressure sensors, or piezoelectric pressure sensors.

2.2.2 Piezoresistive Pressure Sensors

Piezoresistive pressure sensors are the most used of all pressure sensors. They utilise the change in resistance of a metal or semiconductor when stretched to calculate the pressure. This type of sensor uses a strain gauge made from this material that changes resistance when stretched. It is this change in resistance that results in a change in voltage in the output signal (Teesing, 2019).

The change in resistance in metals is mainly due to geometrical changes in the material when stretched. These can be changes in length and cross-section area (as determined by the equation $R = \rho \frac{L}{A}$). However, in semiconducting materials the piezoresistive effect dominates; meaning the resistivity increases with increased strain much more than it does to geometrical changes (Avnet, 2020).

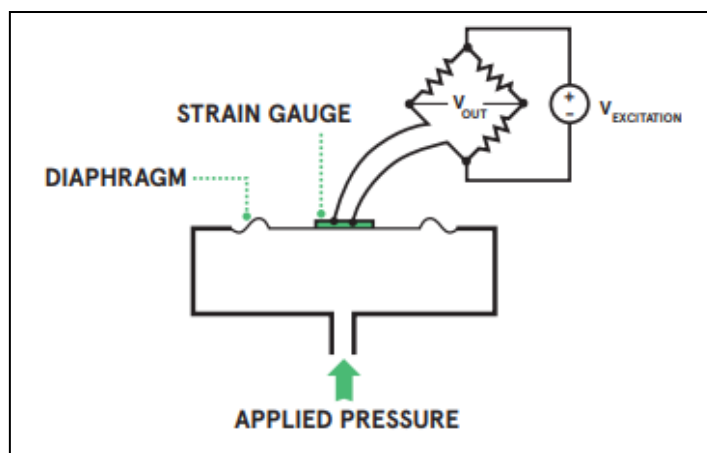


Figure 2.9: A piezoresistive pressure sensor sensing mechanism (Avnet, 2020).

As shown in Figure 2.9, the piezoresistive element of the sensor is bonded to the diaphragm. When pressure is applied, the diaphragm is strained and so is the piezoresistive material. The change in resistance of the material is then measured using a Wheatstone bridge circuit which converts these small changes in resistance to an output voltage. This means the output voltage is directly proportional to the pressure applied. Wheatstone bridge circuits are highly sensitive to small changes in resistance and provide highly accurate measurements, this makes them suitable for strain gauge applications (Avnet, 2020).

Advantages of this type of pressure sensor is that it is robust, less expensive compared to capacitive pressure sensors, and the performance and calibration is stable over time. Having metal film elements makes them simple to construct, durable, and they have a higher operating temperature of up to 200°C, whereas a sensor containing a semiconductor element is limited to operating temperatures below 100°C. However, they provide a much larger output signal making them great for low pressure applications. They can also be made much smaller than metal element pressure sensors meaning they can be implemented as micro-electro-mechanical systems (MEMS). This means that they can be integrated with electronics which can control for temperature dependence and non-linearity (Avnet, 2020, Teesing, 2019).

A disadvantage of piezoresistive pressure sensors is that they consume more power than some other types of pressure sensors. Therefore, they are not suitable for battery powered or portable systems (Avnet, 2020).

An example of a piezoresistive sensor used for medical applications include the in-vivo pressure sensor designed by *Mitrakos et al* for gastrointestinal pressure monitoring. Their sensor offers several advantages, including high sensitivity, compact dimensions (as small as 2.35 mm × 2.35 mm with a thickness of about 150 µm), rapid response and stabilization times, minimal hysteresis, and reliable performance under repetitive stress. It is biocompatible, with a pressure measurement range up to 25 kPa, making it suitable for integration into small-scale biomedical devices like ETT cuffs. However, these sensors also have disadvantages such as potential risks of electrical safety, and a complex and costly fabrication processes. Additionally, their sensitivity to temperature and humidity pose as challenges for tracheal contact pressure measurement (Mitrakos et al., 2024).

2.2.3 Capacitive Pressure Sensors

Capacitive pressure sensors measure pressure by detecting changes in capacitance between two parallel conducting plates, where one (or in some cases both) of these plates is a diaphragm. These conducting plates are separated by a small gap. When a pressure is applied the diaphragm plate moves and the gap between the two plates then becomes smaller. It is this that brings about the change in capacitance. Typically, only one of the plates is a pressure sensitive diaphragm and the other is fixed (Avnet, 2020, Teesing, 2019). An example of the inside of a capacitive pressure sensor is shown in Figure 2.10.

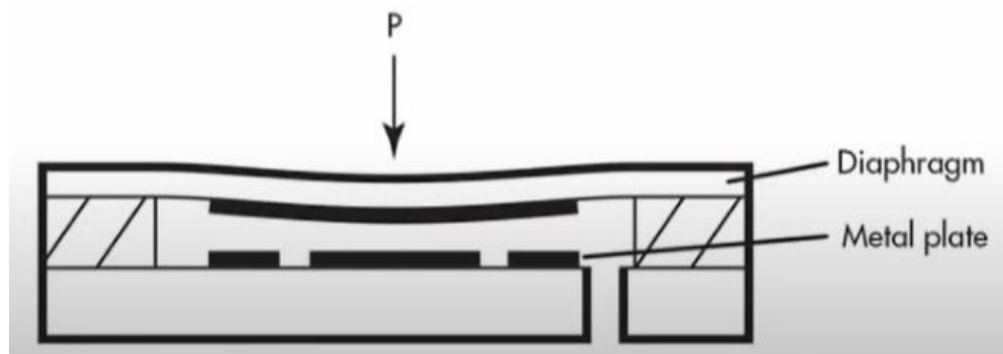


Figure 2.10: Inside a capacitive pressure sensor. Capacitive pressure sensors generally have a diameter range from 5mm to 50mm. The thickness of these sensors can be from 0.5mm.

Capacitive pressure sensors have advantages over other types of pressure sensors. They have very low power consumption and do not require a power source. They are also highly resistant to electromagnetic interference or radiofrequency interference. Capacitive pressure sensors can also be constructed directly on a silicon chip which allows for the construction of very small sensing elements. This means that they can also be a type of MEMS pressure sensor. Despite these positives, capacitive sensors exhibit non-linearity. This is due to the output being inversely proportional to the gap between the two conducting plates. They are also highly sensitive to vibrations (Avnet, 2020, Teesing, 2019).

An example of a capacitive sensor used for medical applications include the pressure sensor designed by *Thouti et al* for the measurement of wrist pulse waveform, demonstrating its capability for real-time pulse monitoring. Their sensor was biocompatible and offers high sensitivity, dimensions of 1cm x 1cm, immediate response and real-time stabilisation. However, these sensors also have disadvantages,

including potential risks of electrical safety, as well as complex and expensive fabrication processes (Thouti et al., 2020).

2.2.4 Piezoelectric Pressure Sensors

Piezoelectric pressure sensors utilise a phenomenon known as piezoelectricity. Piezoelectricity is the electric charge that accumulates across certain materials, like crystal quartz and some ceramics, when a pressure is applied to it. Once pressure is applied, the voltage across the piezoelectric material can be measured. This is essentially how this kind of pressure sensor works. The voltage output is therefore proportional to the pressure applied (Avnet, 2020, Texas Instruments & Karki, 2000, Carter *et al.*, 2010). Figure 2.11 shows the basic structure of a piezoelectric sensor.

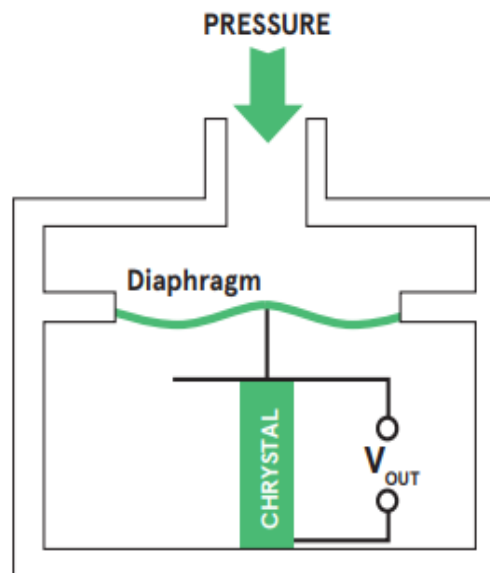


Figure 2.11: Inside a piezoelectric sensor (Avnet, 2020). The crystal is typically 10-16mm in diameter, with a thickness of 0.16mm (Zhu & Li, 2018).

Piezoelectric sensors, unlike piezoresistive and capacitive pressure sensors, do not require any external voltage or current source as they generate an output voltage from the applied pressure on the piezoelectric element. Detecting this signal requires a charge amplifier to convert the signal to a voltage (Avnet, 2020, Carter *et al.*, 2010).

Piezoelectric pressure sensors have many advantages. They are suitable for use in harsh environments, including high temperature environments. Depending on the piezoelectric element, some sensors can work in environments up to 1000°C. As the output signal is generated by their piezoelectric element, they have low power

consumption. Piezoelectric materials like quartz and tourmaline are inexpensive and therefore allow these sensors to be manufactured at low costs.

Despite these advantages piezoelectric pressure sensors also have a few disadvantages. Although these sensors can work in high temperatures, they are highly sensitive to temperature. The piezoelectric element would need to be carefully considered for such applications. As the signal output generated from the element is low, external electronic circuitry must be connected to them. For static pressure measurement, the output signal will gradually drop over time until it reaches zero, even when a contact pressure is present. This is due to a number of possible reasons like imperfect insulation and internal sensor resistance. This, of course, makes them unsuitable for static pressure sensing. These sensors however, are very useful for sensing dynamic pressure changes (Avnet, 2020).

An example of a piezoelectric sensor used for medical applications include the sensor designed by Jiang *et al* for body motion sensing. Their sensor was a thin film that was >2cm x 2cm. Their sensor, although large, offers high sensitivity, immediate response times and real-time stabilisation. However, this sensor's disadvantages mirror those of other sensors in this section, including potential risks of electrical safety, as well as complex and expensive fabrication processes (Jiang et al., 2024).

2.2.5 Conclusion

Piezoresistive, capacitive, and piezoelectric sensors all have unique benefits and have proven useful in medical applications. However, each has drawbacks for use in measuring tracheal pressure with ETT cuffs. Piezoresistive sensors are highly sensitive and responsive but consume more power and are complex to make. Capacitive sensors are low-power and sensitive but suffer from non-linearity and vibration issues. Piezoelectric sensors work well in high temperatures and for dynamic pressures but aren't suitable for static measurements and need external electronics.

These sensors shouldn't be completely ruled out for ETT cuff applications due to their advantages. However, their potential risks and complexities suggest that exploring other sensing technologies might be beneficial. Finding alternative solutions could offer safer and more effective ways to measure tracheal pressures without the risk of electrocution.

2.3 Optical Fibre Sensors

2.3.1 Introduction

Optical fibre sensors (OFS) are increasingly used in research for healthcare applications. This is because OFS offer many advantages: they are biocompatible, small in size, tolerate high temperatures, survive in harsh environments, and are not susceptible to electromagnetic interference. They are also readily multiplexed, allowing the measurement of simultaneous variables (Correia *et al.*, 2018).

2.3.2 Optical Fibre Structure

Optical fibres consist of 3 layers and sometimes an outer jacket. Figure 2.12 shows the structure of an optical fibre.

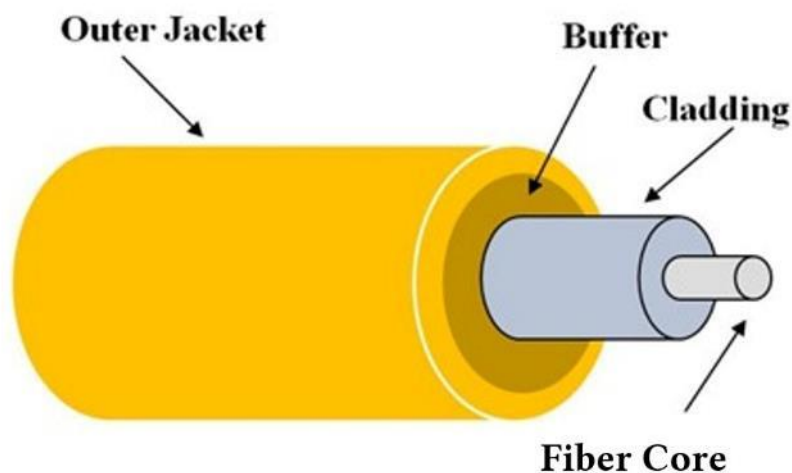


Figure 2.12: Diagram of an optical fibre.

The core (which is made of silica glass) and cladding of the fibre play roles in light guidance whereas the buffer and jacket of the fibre are for protection against damage. The cladding diameter is commonly 125 μm , whereas the core usually ranges from 9 to 50 μm depending on whether the fibre is single mode or multimode (Castrellon-Uribe, 2012).

Silica is predominantly used as the core material in optical fibres due to its excellent optical properties. The cladding is also typically made from silica; or a doped silica material to achieve the desired refractive index profile for light guidance. While silica is the most common material for optical fibres, other materials such as polymers and

certain glasses can also be used to form optical fibres depending on the specific application requirements (Zhang et al., 2022).

2.3.3 Transmission of Light Through an Optical Fibre

Light travels through the fibre core via total internal reflection. The cladding provides a lower refractive index (RI) at the core interface in order to facilitate total internal reflection within the core. This phenomenon will only occur if the angle in which the light that enters the fibre is within the acceptance angle range (Correia *et al.*, 2018).

2.3.3.1 Refractive Index

Refraction is the bending of light when passed from one medium to another. A material's RI indicates how much the path of light bends when being passed through it. The RI of a specific material is determined by the ratio of the velocity of light through a vacuum to the velocity of light through that specific material. Equation 2.1 shows this relationship.

$$n = \frac{c}{v} \quad (2.1)$$

where n is the RI, c is the velocity of light in vacuum, and v is the light's velocity in the material (Rivera-Ortega *et al.*, 2019, Wright *et al.*, 2017)

In an optical fibre, the RI of the core n_1 is greater than the RI of the cladding n_2 . This means that when light passes from the core to the cladding interface, light bends away from the normal line. The angle at which the light bends away from the normal line depends on the angle at which the light enters the second medium. This angle of light

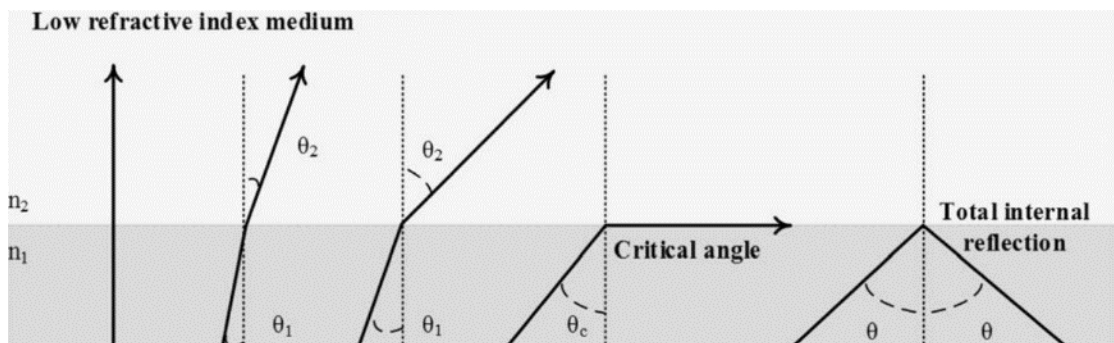


Figure 2.13: Refraction of light as it passes from a medium of high RI to a medium of lower RI: a) no refraction occurs when the light is passed perpendicular to the interface; b) slight refraction at low angle of incidence; c) larger refraction at larger angle of incidence; d) angle of incidence is equal to the critical angle causing light to travel along the interface; e) total internal reflection (Wright *et al.*, 2017).

entering the second medium is known as the angle of incidence. Figure 2.13 illustrates this bending phenomenon.

The angle of refraction of the light can be calculated using Snell's law (Equation 2.2).

$$n_1 \sin \theta_1 = n_2 \sin \theta_2 \quad (2.2)$$

where n_1 and n_2 is the RI of material 1 and material 2, and θ_1 and θ_2 is the angle of incidence and angle of refraction. By knowing the RI of both materials and the angle of incidence, the angle of refraction can be calculated.

2.3.3.2 Total Internal Reflection

A specific angle of incidence known as the critical angle (shown in Figure 2.13d) results in a 90° angle of refraction ($\theta_2 = 90^\circ$). The required angle of incidence (θ_1) to achieve this 90° angle of refraction can be calculated using Snell's law (equation 2.2). It is known that if the angle of incidence is larger than the critical angle, then total internal reflection will occur (shown in Figure 2.13e). This means that the light will not be refracted into the second material but virtually 100% of the beam will be reflected back into the first material. This is essentially how light is transmitted in the core of an optical fibre (Castrellon-Urbe, 2012).

In order for total internal reflection to occur in the core, the light ray must first enter the fibre at a specific angle. This angle is known as the acceptance angle.

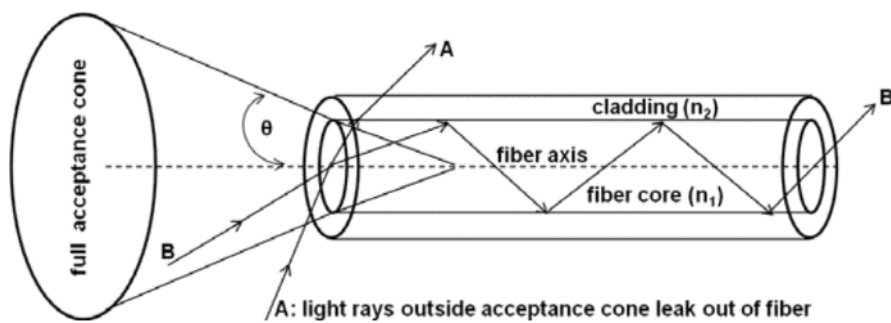


Figure 2.14: Rays of light entering the fibre with an angle inside that of the acceptance cone will undergo total internal reflection (such as ray B). All rays of light outside this acceptance cone will escape into the cladding (Such as ray A) (Kothapalli *et al.*, 2012)

Figure 2.14 shows how a ray of light entering the fibre at the wrong angle, such as light ray A, can be lost to the fibre cladding. This is because when the ray enters the glass core, it refracts. This refracted light then travels to the core-cladding interface and as its incident angle is smaller than the critical angle, it refracts into the cladding rather

than reflecting back into the core. Light ray B however, when entering the glass core, refracts and travels to the core-cladding interface where its incident angle is larger than the critical angle allowing the ray to reflect back into the core and propagate through the fibre (Kothapalli *et al.*, 2012).

2.3.4 Single Mode Fibres and Multimode Fibres

Two types of fibres are commonly used: single mode fibres (SMF) and multimode fibres (MMF). MMFs have a larger core diameter than SMFs (around 50 μm compared to a SMFs 9 μm) which allow for light to pass through them along different paths (or modes). This essentially is what gives them their name as multiple modes can propagate through a MMF, whereas only one mode propagates through a SMF (Curran & Shirk, 2015, Castrellon-Urbe, 2012) . Figure 2.15 shows the differences between MMFs and SMFs.

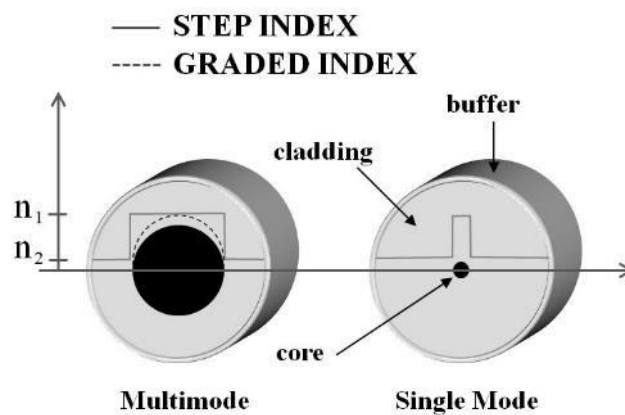


Figure 2.15: The differences between MMFs and SMFs. MMF have a larger core and can be either a step index or gradient index fibre. SMFs are only step index fibres (Fidanboyu & Efendioglu, 2009).

MMFs can further be divided into step index and gradient index fibres. SMFs are step index fibres. Step index fibres have a core with a uniform RI throughout, which experiences a sudden change in RI at the boundary between the core and cladding. In step index MMFs, the paths (or modes) along which the light rays travel differ depending on their incident angle. As a result, the different modes arrive at the end of the fibre at different times. Figure 2.16a shows a step index MMF and the propagation of light through it. On the other hand, in gradient index (sometimes called graded index) fibres the RI is highest in the centre of the core and gradually decreases outwardly from

the core to the core-cladding interface. This reduces the modal dispersion of that experienced in step index fibres, and allows for the different modes to arrive at the end of the fibre at the same time (International Telecommunications Union, 2009, Curran & Shirk, 2015). Figure 2.16b shows a gradient index MMF and the propagation of light through it.

2.3.5 Components of Optical Fibre Sensing Systems

All OFS are comprised of three basic components, an optical source, a sensing element, and a receiver.

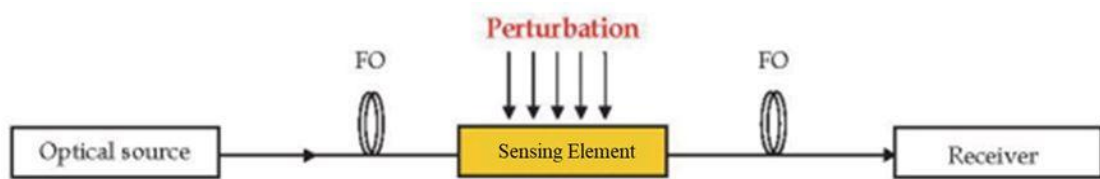


Figure 2.17: Components of a OFS system, adapted from Castrellon-Urbe, 2012.

Figure 2.17 shows the components of an OFS system. Here, Lasers, halogen lamps, and LEDs can be used as optical sources. Single-mode fibres, because of their narrow core diameter, require light sources with narrower spectral widths to minimise modal dispersion and ensure efficient coupling of light into the fibre. Because of this, lasers and Superluminescent LEDs (SLEDs) are used for single mode applications. LEDs however, have a larger spectral width than lasers and SLEDs and are used in multimode applications because their broad spectral width matches the multiple propagation modes of multimode fibres. The optical fibres themselves are used as the sensing element, by converting the perturbation to a change in optical signal. Lastly, a photodetector is used as the receiver. Photodetectors can detect the variation in the optical signal that is caused by the measurand or perturbation (Castrellon-Urbe, 2012, Curran & Shirk, 2015).

2.3.6 Classification of Optical Fibre Sensors

OFSs can be classified based on different parameters. These classifications help determine what type of optical sensor is needed for a specific application. OFSs can be classified by their sensing location and their operating principle.

2.3.6.1 OFS Classification Using Sensing Location

An OFS can be classified as extrinsic or intrinsic. In an extrinsic OFS the fibre acts only as a means of carrying light to and from an external source where the sensing takes place.

In an intrinsic OFS the optical fibre itself is the sensitive material while also the medium to transport the optical signal. One or more of the physical properties of the fibre undergo change which results in changes in some of the characteristics of the propagating light in the fibre. In this case, the sensing takes place in the fibre rather than an external optical device (Pendão & Silva, 2022). Figure 2.18 shows a diagram of these two types of OFSs.

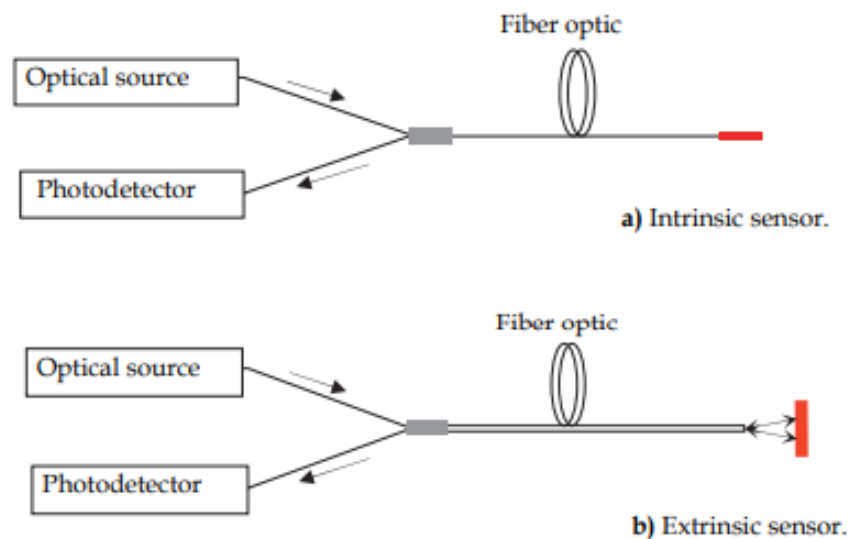


Figure 2.18: a) diagram of an intrinsic OFS b) diagram of an extrinsic OFS (Castrellon-Urbe, 2012).

Based on this classification, an intrinsic sensor is required (for the application of tracheal contact pressure measurement) in order to minimise components placed in the body.

2.3.6.2 OFS Classification Using Operating Principle

An OFS can be classified as either intensity based, phase based, wavelength based, or polarisation based. Detecting changes in these parameters allow for sensing of different external perturbations.

- a) Intensity based sensors

Intensity based OFSs focus on detecting a change in the light signal's magnitude. This generally means that these sensors rely on the signal undergoing some loss. This can be achieved using two different detection methods. The first utilises apparatus to convert what is being measured into a force that bends the fibre and causes attenuation of the signal. The other method is through absorption. In this approach, various thin films with specific absorption characteristics are deposited onto the fibre to create a coating. When light propagates through the fibre, it interacts with this coating, leading to changes in absorbance at certain wavelengths or spectral ranges. These changes in absorbance can then be correlated with the variations in the sensed parameter. Intensity based sensors typically require a higher amount of light to compensate for the attenuation caused by the sensing mechanism. Therefore, MMFs are commonly used (De Acha *et al.*, 2021, Fidanboyly & Efendioglu, 2009). MMFs allow for multiple modes of light to propagate through the fibre, including some that penetrate into the cladding and interact with the deposited coating. This interaction results in changes in absorbance, which can be accurately measured to determine the sensed parameter. Coated OFSs are commonly used to sense chemicals, gases and biological substances like enzymes and proteins (Fidanboyly & Efendioglu, 2009).

b) Phase based sensors

Phase modulated sensors generally employ two SMFs. One is used as the sensing fibre and the other as the reference fibre. The phase of the light in the sensing fibre is compared to the reference fibre using an interferometer. The light from a laser light source is split into both these fibres, when the sensing fibre is exposed to the perturbation, a phase shift occurs and can be detected by the interferometer by comparing it to the phase of light in the reference fibre (which was isolated from the perturbation). There are four commonly used interferometers: the Mach-Zehnder, Michelson, Fabry-Perot, and Sagnac (Fidanboyly & Efendioglu, 2009, Castrellon-Urbe, 2012). Phase modulated sensors are highly sensitive and more accurate than intensity based sensors; however, they are also more expensive (Pendão & Silva, 2022).

c) Wavelength based sensors

Wavelength-modulated sensors measure the wavelength change in the fibre. The fibre Bragg grating (FBG) sensor is the most widely used type of wavelength modulated sensors as it is capable of both single point and multi-point sensing (Pendão & Silva,

2022, Fidanboyly & Efendioglu, 2009). An FBG is created by making periodic changes in the RI of the core of a SMF. This causes the FBG to reflect a small portion of the light spectrum propagating through it. The reflected wavelength of light corresponds to the Bragg wavelength and the remainder of the spectrum transmits along the fibre. When the fibre is placed in a sensing environment, changes in the grating properties cause the reflected wavelength of light to change. This change can be detected by an optical interrogator. More details on FBGs are mentioned in Chapter 2.4.1.

d) Polarisation based sensors

Polarisation-based sensors detect changes in the polarisation state of light running through a fibre. There are different types of polarisation states; these include linear, elliptical, and circular. First, light from a source is polarised using a polariser and then propagated along a polarisation maintaining fibre. The RI of the fibre changes when placed under a stress or strain. These changes are in the same direction as the applied stress or strain which causes a phase difference in the polarisation states. This phase difference can be detected (Pendão & Silva, 2022, Fidanboyly & Efendioglu, 2009).

While intensity-based OFSs can be used as contact pressure sensors, they may not be ideal for all applications, especially those requiring rapid response times or very high precision. These sensors typically rely on the bending of the fibre to measure pressure, which can lead to fibre breakages (Perrotton et al., 2013).

When looking at phase-based OFSs, they can be excellent for contact pressure sensing due to their high sensitivity and accuracy, which enable the detection of minute pressure changes with great precision. However, of the four interferometers, the Michelson is the simplest. It consists of a light source, optical couplers, two mirrors (FBGs can be employed for this), and a photodetector that records the interference pattern. However, since one of the fibres serves as the reference, it must experience the same conditions and environment (such as temperature and strain) as the pressure sensor, but without being subjected to pressure, which is the variable being measured (Kirkendall & Dandridge, 2004). This could be difficult to achieve in a tracheal application as both the reference fibre and sensing fibre would need to be placed on the ETT cuff to ensure both fibres experience the same temperature and strain changes. Design alterations would need to be considered for how to prevent the reference fibre from experiencing tracheal contact pressure. Another option for a pressure sensor is the Fabry-Perot

interferometer. This can be achieved by aligning two optical fibre ends parallel to each other with a known separation distance. Changes in the fibre separation distance due to applied pressure can be measured by monitoring the interference pattern of light transmitted through the fibre ends. This however is more complex to construct than other possible sensing options (Xia et al., 2015). Phase-based OFSs are generally more complex and expensive compared to other types, due to the need for interferometric setups. Additionally, they are sensitive to environmental changes like temperature fluctuations and mechanical vibrations, necessitating compensation mechanisms to maintain accurate readings (Xavier et al., 2011).

Wavelength-based OFS, can be highly effective for contact pressure sensing. These sensors offer several advantages, including high sensitivity, accuracy, and the ability to perform both single-point and multi-point sensing (Albert et al., 2013). FBG sensors work by measuring changes in the wavelength of reflected light, which correspond to pressure-induced deformations in the fibre. This makes them particularly suitable for applications requiring precise and reliable measurements. They are also highly sensitive to minute pressures (Othonos *et al.*, 2006). Additionally, FBG sensors are small, making them easier to integrate into compact spaces like an ETT cuff. However, these sensors also require sophisticated interrogation equipment to analyse the reflected wavelengths accurately. They are also sensitive to temperature changes, which can affect measurement accuracy unless compensated for (Massaroni *et al.*, 2017).

Polarisation-based OFSs can be effective for contact pressure sensing due to their high sensitivity to changes in the polarisation state of light caused by external perturbations, such as pressure. When a polarisation-maintaining fibre is subjected to stress or strain, the resulting change in refractive index induces a phase difference between the polarisation states. This phase difference can be detected and correlated with the applied pressure, allowing for precise measurements (Bashan et al., 2021). However, polarisation-based OFS may face several challenges that could limit their suitability for ETT cuff placement and tracheal applications. These sensors often require complex and precise alignment and sophisticated interrogation systems to accurately detect polarisation changes. Additionally, they require pressure large enough to create a deformation in the fibre core causing a change in polarisation modes (Chehura et al., 2004). The pressures required to create such a deformation are much larger than those experienced in the trachea. This, coupled with their need for careful alignment can also make them less practical for tracheal contact pressure sensing.

2.3.7 Conclusion

In the application of tracheal pressure sensing, FBG sensors offer distinct advantages. They can minimise patient discomfort and risk of tissue damage due to their small size and flexibility. Furthermore, they require no electric charge to flow through the sensor. FBG sensors are easy to fabricate and exhibit high sensitivity, enabling precise measurement of subtle pressure changes within the trachea. Comparatively, other optical configurations like Fabry-Perot interferometers, while capable of measuring pressure changes through interference fringes, are more complex to construct and calibrate. They are also sensitive to mechanical stresses, which could affect their accuracy in physiological environments like the trachea. Unlike Fabry-Perot interferometers, FBGs are inherently simpler in structure, consisting of a single optical fibre with periodic RI changes, making them robust, compact, and less susceptible to environmental influences such as bending. The basic Michelson interferometer requires an additional reference fibre to undergo the same environment as the pressure sensor without experiencing the pressure perturbation making it challenging to integrate for this application.

Overall, OFSs are well suited for use in biological measurement as they are biocompatible, extremely small in size, can withstand harsh conditions, and are not affected by electromagnetic interference. When comparing the different OFSs that can be used for pressure sensing, wavelength-based OFSs seem the most simple and viable sensing approach for in-vivo tracheal contact pressure sensing.

2.4 Optical Fibre Pressure Sensors

2.4.1 Fibre Bragg Gratings

An FBG is a periodic inscription on the fibre core creating a periodic change in RI. This can be done by exposing it to an intensity pattern from a UV or femtosecond laser. The FBG causes a narrowband of wavelengths to be reflected back through the fibre while transmitting the other wavelengths (Othonos *et al.*, 2006, Correia *et al.*, 2018).

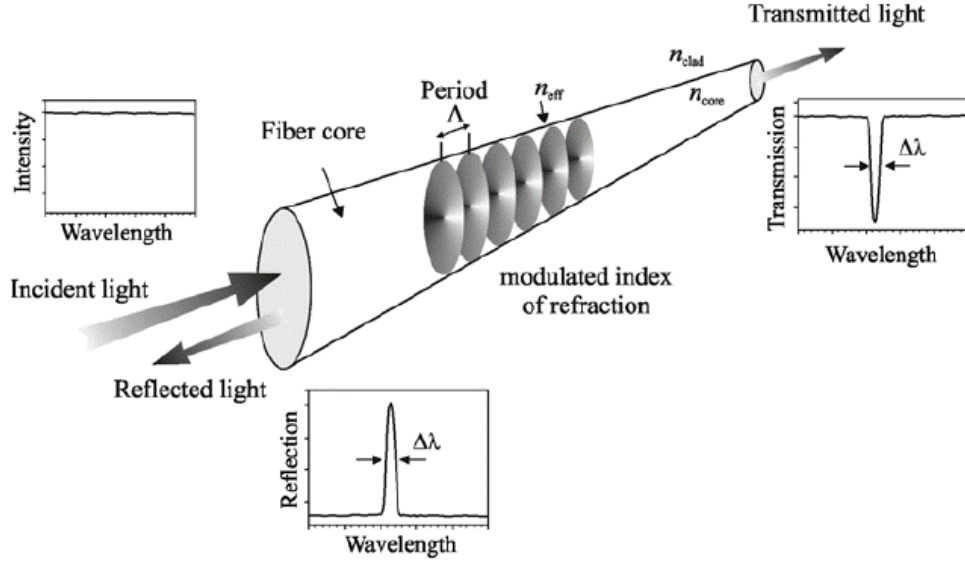


Figure 2.19: Diagram representing a Bragg grating inscribed into the core of an optical fibre and the transmitted and reflected wavelengths. (Othonos *et al.*, 2006).

Figure 2.19 illustrates the broadband of wavelengths that make up the incident light going into the fibre core. As the light travels through the core and passes into the FBG, a narrowband of wavelengths are reflected back while all other wavelengths are transmitted (Othonos *et al.*, 2006, Correia *et al.*, 2018). The reflected Bragg wavelength is determined by equation 2.3 (Massaroni *et al.*, 2017):

$$\lambda_{\text{Bragg}} = 2\eta_{\text{eff}} \Lambda \quad (2.3)$$

Where η_{eff} is the effective RI of the FBG and Λ is the period of the grating. Therefore, the Bragg wavelength is sensitive to factors that affect the RI and the period of the grating such as temperature and strain (Massaroni *et al.*, 2017). Strain on the fibre reduces the RI due to changes in the photoelastic effect; strain also causes the period of

the grating to be pulled further apart. An increase in temperature results in changes to the grating period and RI due to thermal expansion (Yeo et al., 2005).

The wavelength shift of FBGs is measured using an optical interrogator, which typically includes a laser light source and a photodetector. The photodetector captures the reflected spectrum from the FBG and analyses the Bragg wavelength shifts. A typical interrogator can measure wavelengths in the range of 1525 to 1575nm and have a resolution of 0.1pm (Ibrahim *et al.*, 2020).

Because FBGs reflect the Bragg wavelength, FBG sensors are highly suitable for multiplexing, as they allow for multiple gratings with different Bragg wavelengths to be written along a single fibre. This enables simultaneous monitoring of various parameters at multiple points along the fibre (Ma et al., 2016).

2.4.2 Optical Fibre Pressure Sensing Mechanism

The intrinsic sensitivity of FBGs to pressure is relatively low because direct pressure induces minimal changes in the refractive index and grating period. Therefore, it is necessary to transduce pressure into strain to amplify the measurable effect (Schenato et al., 2021).

The strain exerted on a fibre can be detected by an FBG. The period of the grating changes in response to the strain and a wavelength shift in the reflected light is generated. This phenomenon can be used to measure pressure (Alshaikhli, 2018). Pressure measurement can be achieved by attaching the FBG onto different diaphragms, springs or even by embedding it into a polymer patch. The diaphragm moves when subjected to a load, while the spring is either pulled or compressed. This causes strain on the FBG and a wavelength shift in reflected light can be detected. This wavelength shift is proportional to the load applied. If the load applied on the diaphragm or spring is known, the pressure can be calculated. Figure 2.20 shows how the wavelength shift increases as the pressure increases.

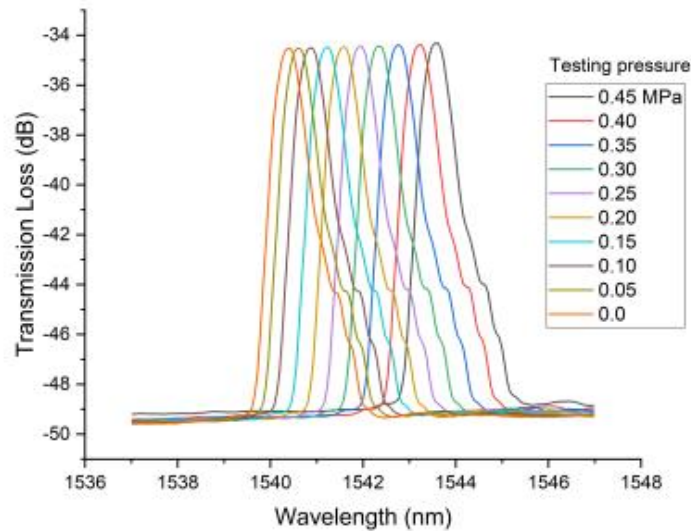


Figure 2.20: Clear wavelength shift (viewed from optical spectrum analyser) as compressive pressure increases.

Central wavelength 1540nm (Nguyen *et al.*, 2022).

Additionally, transducing pressure to axial strain helps avoid induced birefringence caused by loading an optical fibre directly. Birefringence is caused when the load causes the fibre core to deform out of its symmetrical shape; causing the Bragg peak to split into two distinct wavelengths corresponding to the orthogonal polarisation modes (Chehura *et al.*, 2004).

2.4.3 Research Using FBGs for Pressure Sensing

The publications mentioned in this section utilised one of two experimental variations: their sensors either contained a spring/bellows or a diaphragm. Figure 2.21 shows an example for each of these experimental setups. The experimental setup varies from publication to publication; however, the overall working principle of the sensor remains the same. For a setup involving a spring, as the spring compresses the FBG relaxes, and a wavelength shift will occur in one direction (wavelength decreasing). As the spring is stretched (tension) it will create strain in the fibre and the wavelength shift will move in the other direction (wavelength increasing). In the case of the diaphragm; when the diaphragm is relaxed, there isn't a shift in wavelength. However, as the diaphragm bends, so does the fibre, causing a wavelength shift that is proportional to the bend in the diaphragm.

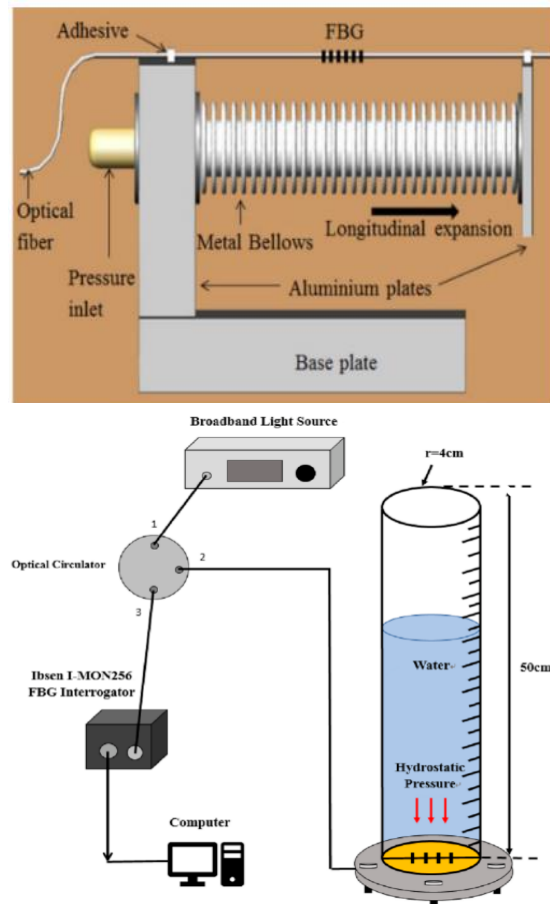


Figure 2.21: top- a spring/bellow sensor setup where the bellow length is 5cm and the diameter is 1.1cm (Pachava *et al.*, 2015), bottom- a diaphragm sensor setup where the diameter of the sensor is 16cm (Her & Weng, 2021).

Pachava *et al.* attached an FBG to metal bellows to create their sensor head. Their setup is shown in Figure 2.21 (top). The metal bellows structure had a length of 5cm with a diameter of 11mm. A pressure of up to 40 psi (276kPa) was applied in increments of 5 psi (34.5kPa). Their sensor showed a sensitivity of 90.6pm/psi, which is equivalent to 13.15pm/kPa (Pachava *et al.*, 2015). The pressures investigated in this experiment were much larger than those experienced in the trachea (2-3kPa to create a ETT cuff seal). Additionally, although the sensor is quite small, it is too large and bulky for placement on the ETT cuff. As the sensitivity of this sensor provides a resolution of 0.0076kPa (assuming a typical optical interrogator was utilised with a resolution of 0.1pm), this sensitivity may be too high and could pick up unwanted noise (as mentioned in Chapter 2.1.6).

Li *et al.* created a sensor by placing an FBG on a spring inside metal casing (As shown in Figure 2.22).

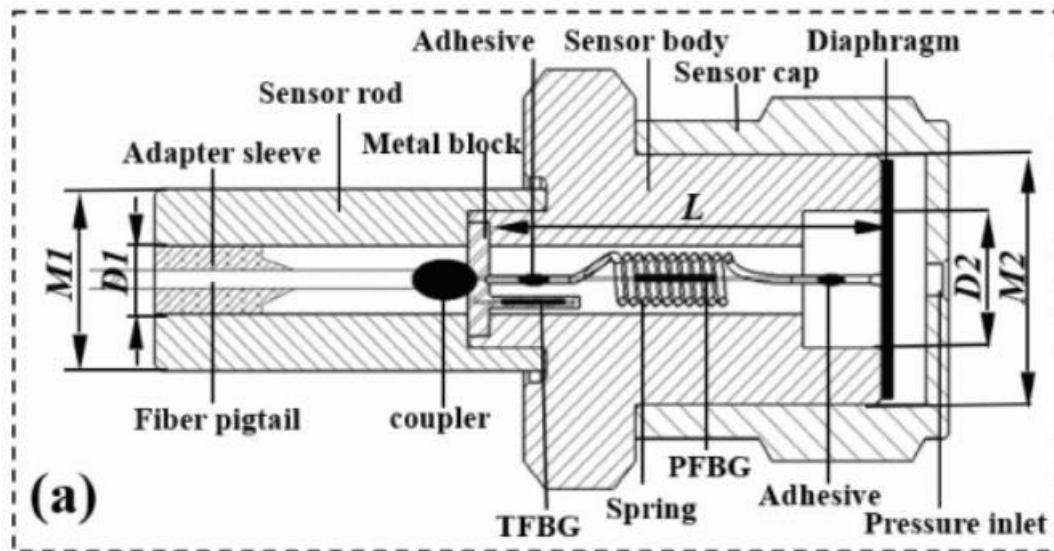


Figure 2.22: schematic diagram of pressure sensor using an FBG attached to a spring in metal casing (Li *et al.*, 2022).

They applied a pressure of up to 100MPa in 2MPa increments. Their sensor dimensions excluding the sensor rod was 4cm x 4cm with a diameter of around 4cm. Their sensor showed a sensitivity of 79.7pm/MPa, which is equivalent to 0.079pm/kPa (Li *et al.*, 2022). Once again, the pressures used in this experiment were much larger than those experienced in the trachea. Additionally, and their sensor size (4cm x 4cm sensor) and design make it is too large and bulky for placement on the ETT cuff. Additionally, as the sensitivity of this sensor provides a resolution of 1.26kPa, this sensitivity is too low. Swamy *et al.* created a sensor using an FBG, stainless steel disks and a copper diaphragm to create their sensor (as shown in Figure 2.23). Their sensor had an outer diameter of 60mm and an inner diameter of 30mm.

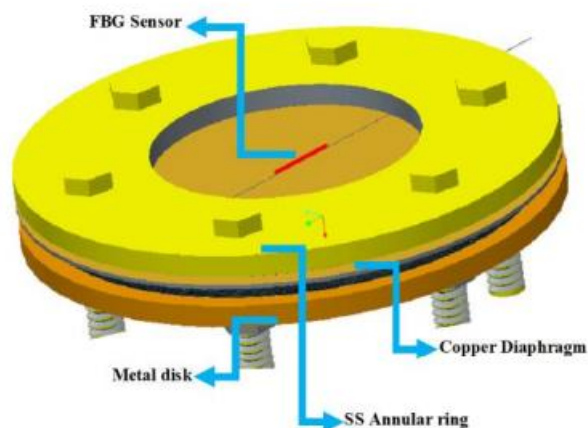


Figure 2.23: schematic diagram of pressure sensor using an FBG attached to a copper diaphragm (Swamy *et al.*, 2020).

Pressures of up to 0.04MPa were applied. Their sensor showed a sensitivity of 24.59 nm/MPa, which is equal to 24.59pm/kPa (Swamy *et al.*, 2020). Although the pressures their sensor was too large for use on the ETT cuff. The diaphragm part of their sensor (the inner diameter of 30mm) is a reasonable size and could potentially be adjusted for ETT cuff use. As the sensitivity of this sensor provides a resolution of 0.004kPa, this sensitivity is too high. However, by adjusting the diameter, a more suitable resolution could be achieved.

Huang *et al.* attached FBGs to a Bourdon tube which acted as a diaphragm. Pressure enters the tube at the bottom causing it to bend. The bending of the tube resulted in the stretching of the fibre and therefore the FBGs. The radius of their sensor was 33mm and a thickness of 5mm. Figure 2.24 shows a schematic diagram of their sensor configuration.

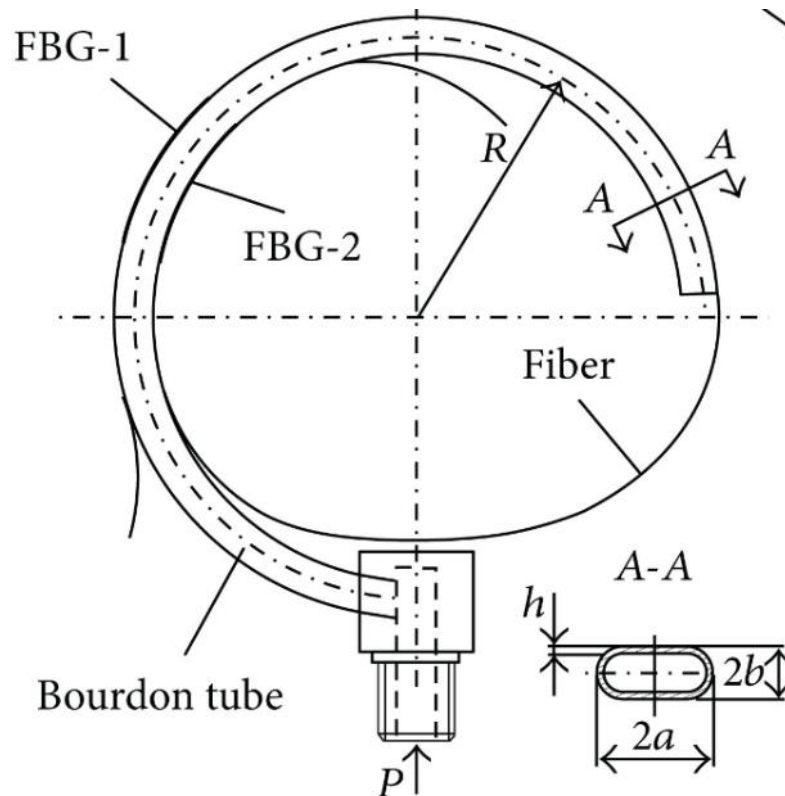


Figure 2.24: schematic diagram of pressure sensor using an FBG attached to a Bourdon tube (Huang *et al.*, 2013).

A pressure of up to 1MPa was applied in 0.1MPa increments. Their results show that their sensor has a sensitivity of 1.41pm/kPa (Huang *et al.*, 2013). The pressures used in the experiments were larger than those experienced in the trachea and the size of their sensor is bulky. Utilising a tube as a diaphragm requires both a larger length and a larger thickness that would not be feasible for ETT cuff use. Despite this, the sensitivity of

this sensor provides a resolution of 0.07kPa, which fits into the target specification range (described in Chapter 2.1.6).

Her & Weng created three pressure sensors by connecting an FBG to an epoxy diaphragm. The diameter of all sensors was 50mm and the thickness of the epoxy diaphragm varied with each of the sensors. The sensor was placed at the bottom of a cylindrical container and water was added. The water level began at 5cm and increased in steps of 5cm until it reached 30cm. These 5cm increments corresponds to a hydrostatic pressure increase of 0.49 kPa and a maximum pressure of 2.943 kPa was reached. Their sensor can be seen in Figure 2.21 (bottom). The sensitivities of their sensors with diaphragm thicknesses of 0.5mm, 0.7mm, and 1mm were 175.5 pm/kPa, 89.5 pm/kPa, and 43.7 pm/kPa, showing a decrease in sensitivity with increased diaphragm thickness (Her & Weng, 2021). This shows that the thickness of the diaphragm influences the sensitivity of the sensor. Additionally, the pressures used in these experiments were low pressures like those experienced in the trachea. The sensitivities achieved by these sensors are very high compared to other sensitivities seen so far, and therefore provide very high resolution (0.00057 pm/kPa, 0.001 pm/kPa, and 0.002 pm/kPa for the 0.5mm, 0.7mm and 1mm thick diaphragms respectively). These resolutions are too high and may detect unwanted noise. However, this experiment indicates that increasing the diaphragm's thickness can further reduce the resolution. With this modification, the sensor could potentially be adapted for use in ETT cuffs.

Yang *et al.* created a sensor by encapsulating an FBG in a dual bellows structure. The inner bellow diameter was 14mm and the outer bellow diameter was 28mm. A load of up to 10kPa was applied in 0.5kPa increments. Figure 2.25 shows a schematic diagram of their sensor configuration.

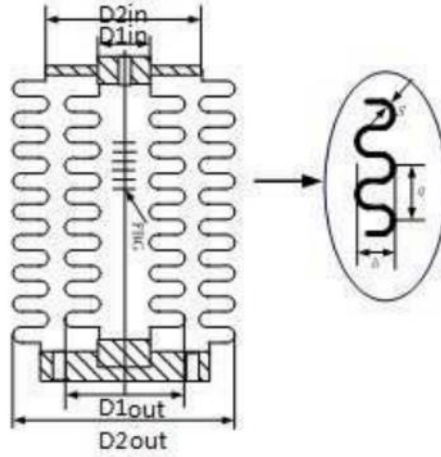


Figure 2.25: schematic diagram of pressure sensor using an FBG encased in dual bellows (Yang *et al.*, 2013).

Their sensor showed a sensitivity of 184.98nm/MPa which is equal to 184.98pm/kPa (Yang *et al.*, 2013). Although this sensor is too bulky to be placed on an ETT cuff, it showed a very high sensitivity. The resolution of this sensor is 0.00054pm/kPa which is too high a resolution.

Fajkus *et al.*'s sensor consisted of an FBG placed in 3D printed casing. The fibre was attached a 0.6mm thick movable membrane. The sensor had a diameter of 38mm and was 130mm long. The sensor configuration can be seen in Figure 2.26.

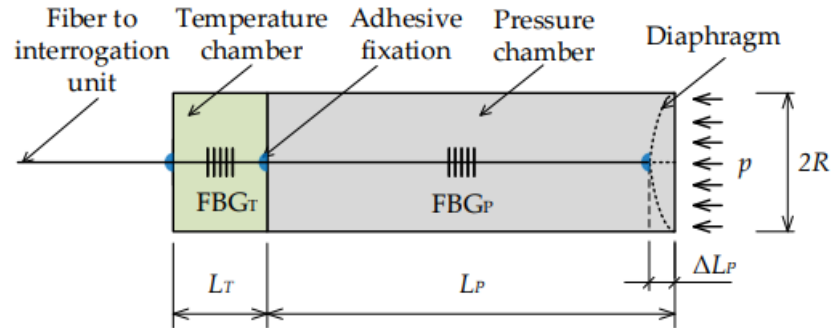


Figure 2.26: schematic diagram of pressure sensor using an FBG attached to a movable membrane. L_P : 100mm, L_T :30mm (Fajkus *et al.*, 2021).

A pressure of up to 8.5 mbar was applied. The sensitivity was calculated as 9.086 pm/mbar which is equivalent to 90.86pm/kPa (Fajkus *et al.*, 2021). This sensor configuration is unique as it utilises a diaphragm in the same way other papers in this section have utilised a spring. This sensor is bulky however, it does include a temperature FBG which the other sensors in this section have not utilised. A temperature FBG is needed for the ETT cuff application, and this publication shows that with the inclusion of another FBG the length of the sensor may increase. The resolution of this sensor is 0.001pm/kPa and is too high. This sensor could potentially

be made smaller with smaller casing however, by reducing the size of the casing, the diameter of the diaphragm will become smaller which may have a large effect on the sensitivity. Additionally, rigid casing should be avoided in tracheal applications, as it can pose a risk. Prolonged contact with rigid items could cause injury to the trachea.

Vorathin *et al.* created a sensor using an FBG attached to a rubber diaphragm. The diaphragm was held by metal rings and Allen bolts. The diameter of the diaphragm was 45mm and the thickness was 1.26mm. A pressure of 10kPa was applied in 1kPa increments. The sensitivity of the sensor was 100.7pm/kPa (Vorathin *et al.*, 2018). This paper also used lower pressures like those that are experienced in the trachea. Once again, the diaphragm approach has shown to create a highly sensitive sensor. Like the other diaphragm approaches, this sensor has too large a diameter for ETT cuff placement and the diameter would need to be altered for tracheal application. This could alter the performance of the sensor. Additionally, this sensors resolution is 0.00099pm/kPa, which is too high a resolution. However, altering the diaphragm could bring about a more suitable resolution to meet the target specification.

Wang *et al.* created their pressure sensor for measuring sub-bandage pressure in compression therapy. They attached two fibres each with an FBG to a metal disc in a double helix configuration, so that the FBGs runs through the centre hole of the disc. This allows the fibre to act as its own diaphragm. When a load is applied to the unit, the fibre bends into the hole in the centre of the disc creating strain on the fibre as it is attached on either end of the metal disc. Figure 2.27 shows their sensor configuration.

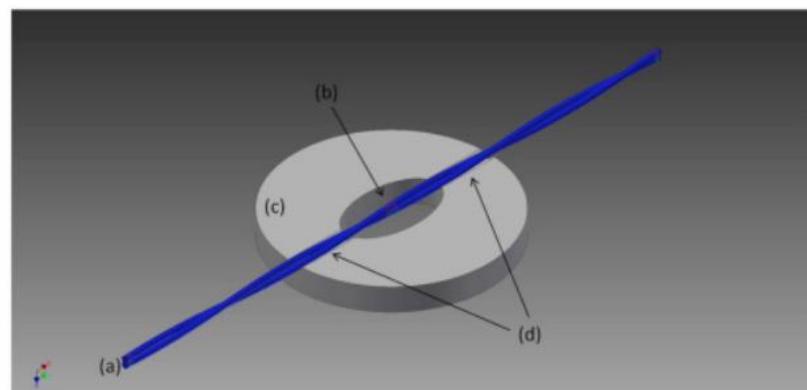


Figure 2.27: diagram of pressure sensor using two fibres (contain FBGs) twisted to form a double helix (a). The fibre is attached to the disc (c) and attachment points (d). The fibre acts as a diaphragm by bending into to hole in the disc (b) when a pressure is applied (Wang *et al.*, 2013).

Wang's sensor consisted of 4 of these units making it more than 30cm long and around 1.5mm thick (although each single unit had a diameter of approximately 7.5 mm and a thickness of 0.9 mm). A pressure of up to 40mmHg (5.3kPa) was applied to the sensor.

The sensitivity of the sensor was 4.8pm/mmHg which is equivalent to 36pm/kPa (Wang *et al.*, 2013). The dimensions of a single unit in this sensor is so small that it could possibly be integrated onto an ETT cuff as it is. However, it is crucial to consider the impact of rigid items, such as the metal disc, on the trachea over prolonged periods, as they could cause injury. Furthermore, despite the small diaphragm size, the sensor's sensitivity remains very high, with a resolution of 0.0027 pm/kPa, which may lead to the detection of unwanted noise.

Arkwright *et al.* created a sensor to measure muscular compressions in the oesophagus. He did this by using an FBG attached to a diaphragm which was connected to supports. Their sensor consisted of 32 of these units. The dimensions of the sensor are not mentioned however, the units are spaced at 10mm and the diameter of their sensor was 3mm. Figure 2.28 shows a diagram of their sensor.

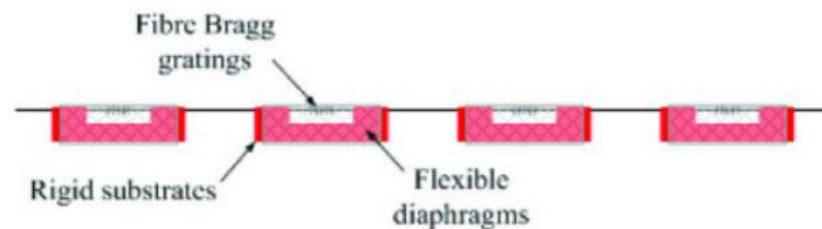


Figure 2.28: diagram of pressure sensor, each unit spaced at 10mm. (Arkwright *et al.*, 2009)).

A pressure of up to 200mmHg (26.6kPa) was applied. The sensitivity of the sensor was calculated to be approximately 1.2pm/mmHg which is equivalent to 9pm/kPa (Arkwright *et al.*, 2009). The resolution of this sensor is 0.011kPa which is just within range of the target specification. However, their sensor contains rigid substrates which could pose a risk for tracheal injury. In order to integrate this sensor onto the ETT cuff, other less rigid materials would need to be considered. It is important to consider by changing materials, the performance of the diaphragm sensor may change.

Table 2.1 shows a summary of the sensing mechanisms used by these researchers along with their dimensions and the sensitivities they achieved.

Table 2.1: Comparison of FBG pressure sensors' sensing mechanism and their corresponding sensitivities in different publications.

Publication	Pressure Sensing Mechanism	Wavelength Direction	Sensitivity
Pachava <i>et al.</i> (2015)	Spring (by using bellows)	Increasing (red shift)	13.14pm/kPa
Li <i>et al.</i> (2022)	Spring	Decreasing (blue shift)	0.079pm/kPa
Swamy <i>et al.</i> (2020)	Diaphragm	Increasing (red shift)	24.59pm/kPa
Huang <i>et al.</i> (2013)	Diaphragm	Increasing (red shift)	1.414pm/kPa
Her & Weng (2021)	Diaphragm of different thicknesses: 0.5mm, 0.7mm, and 1mm	Increasing (red shift)	175.5 pm/kPa, 89.5 pm/kPa, and 43.7 pm/kPa
Yang <i>et al.</i> (2013)	Spring (by using bellows)	Increasing (red shift)	184.98pm/kPa
Fajkus <i>et al.</i> (2021)	Diaphragm	Increasing (red shift)	90.86pm/kPa
Vorathin <i>et al.</i> (2018)	Diaphragm	Increasing (red shift)	100.7pm/kPa
Wang <i>et al.</i> (2013)	Fibre acting as diaphragm	Increasing (red shift)	36pm/kPa
Arkwright <i>et al.</i> (2009)	Diaphragm	Increasing (red shift)	9pm/kPa

Although many of the sensors mentioned (like those created by Pachava, Swamy, Her & Weng, Yang, Fajkus, and Vorathin) are shown to be very sensitive, they were created for fluid pressure measurement and are quite bulky (and do not meet the target specification for size/dimensions or sensing resolution). Many are also made with materials that would not be suitable for in-vivo applications. Wang's sensor was the only one with dimensions mentioned within the range of the target specification, at around 1.5mm thickness. Although Wang used multiple units in his sensor, a single unit could have been made to around 1cm long and 0.9mm. This is a very small sensor and would be suitable for ETT cuff placement. However, by attaching their FBG to a metal disc (like many of the diaphragm designed sensors mentioned in this section), this causes the sensor to be inflexible. For the ETT cuff contact pressure application, an

inflexible sensor could cause damage to the trachea as a rigid material would be placed against its curvature for long periods of time. Additionally, a rigid sensor could open pathways for aspirate to enter the lungs when the ETT cuff is inflated.

While some of the materials used in these papers were not biocompatible, and their dimensions may be too large, their concepts could potentially be adapted for this application with the correct materials and by altering the sizes. However, it is essential to consider whether miniaturisation would influence the performance of these sensors. The appropriate dimensions for a sensor in this application would be no more than 1cm x 1cm in length and width, and no more than 2.5mm in thickness (as mentioned in Chapter 2.1.6). Miniaturising the sensor to these dimensions could impact its sensitivity and accuracy, which are crucial for effective pressure measurement in the trachea. A sensor for this application should be made entirely from biocompatible materials, be as miniature as possible, and ideally, created from soft (non-rigid) materials.

2.4.4 FBGs Embedded in Polymer for Contact Pressure Sensing

When an FBG is embedded in a polymer patch, such as epoxy, and a force is applied on this polymer patch; the patch expands along the fibre resulting in a strain. Figure 2.29 demonstrates this. The wavelength that is reflected back will have a shift that is proportional to the strain applied to the fibre, and therefore pressure applied to the patch. From the patch size and dimensions, area can be calculated; and ultimately, the pressure applied can be determined (Correia *et al.*, 2018, Alshaikhli, 2018).

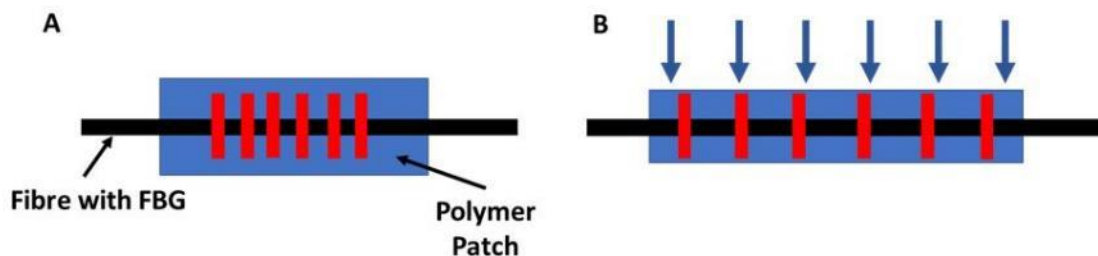


Figure 2.29: A: FBG in polymer patch without a load applied to it. B: load applied to patch causing it to expand along the fibre, creating strain in the fibre and the period between the gratings become larger.

Nguyen *et al.* created an FBG pressure sensor by embedding the FBG in a 3D printed oval shell made up of polylactic acid (PLA). Their sensor was 25mm thick, 36mm long, and 6mm wide as is shown in the first row in Table 2.2. They experimented with

pressures up to 0.45MPa in 0.01MPa increments and their sensor showed a sensitivity of 7.102nm/MPa, which is equivalent to 7pm/kPa (Nguyen *et al.*, 2022). Although this sensor is small, it is still too large and rigid to be placed on an ETT cuff and used in the trachea. Modifying its dimensions to better suit the trachea would alter its performance. Additionally, since the pressure increments tested were 0.01 MPa (10 kPa), it is difficult to determine the sensor's performance for pressures below 10 kPa. Assuming a typical optical interrogator with a 0.1 pm resolution, the sensor's resolution is estimated to be 0.014 kPa. While this resolution is suitable for a tracheal sensor, it assumes the sensor behaves consistently under loads below 10 kPa.

Hong *et al.* created a sensor by embedding FBGs into a 1cm thick, 10cm x 10cm 3D printed PLA block. The area around the FBG centre was hollow as shown in the second row in Table 2.2. A pressure of up to 2MPa was applied in 400kPa increments. They found their sensor to have sensitivity of 13.22kPa/pm, which is around 0.0756pm/kPa. (Hong *et al.*, 2019). Much like Nguyen, Hong's sensor is 3D printed with PLA. However, it is considerably larger. Their sensitivities are very different showing that by applying different shapes and thicknesses it is possible to achieve different sensitivities for the same material. Again, as the increments in pressure were 400kPa it is difficult to tell how this sensor performs for lower pressures. This sensor however, has a resolution of 1.3pm. This resolution is too low for a tracheal contact pressure sensor. Additionally, like Nguyen's sensor, Hong's sensor is too rigid to be placed in the trachea.

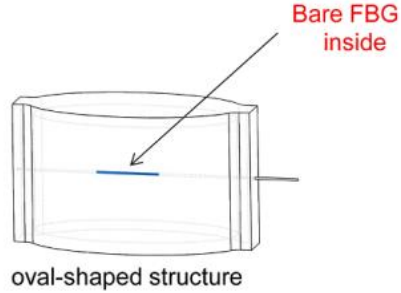
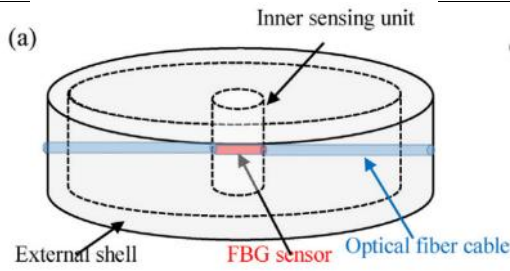
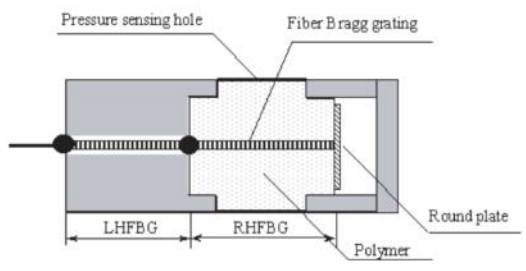
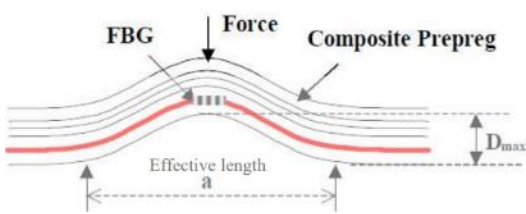
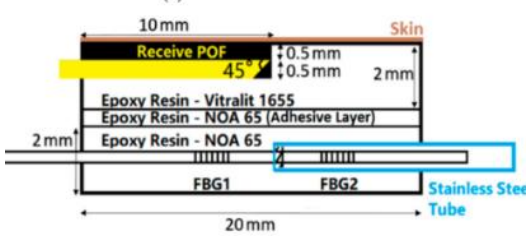
Zheng & Zhang created a sensor that consisted of an FBG embedded in a polymer filled metal cylinder. The polymer used was silicone rubber. The metal cylinder has two openings around the area where the polymer is located allowing for pressurisation (as shown in the third row in Table 2.2). The cylinder is 11mm in diameter and 20mm long. A pressure of up to 0.3MPa was applied. The sensitivity was recorded at 14.92nm/MPa, equivalent to 14.92pm/kPa (Zheng & Zhang, 2005). This sensor has demonstrated greater sensitivity compared to those made from PLA in the last two publications. Additionally, its resolution is 0.007 kPa, which is slightly high for a tracheal pressure sensor (target resolution of 0.01kPa – 0.1kPa). Moreover, similar to the previous publications, it is challenging to determine how this sensor responds to pressures below 0.4 MPa, as this is the smallest pressure applied, and the pressure increments used in this study are not specified.

Suresh *et al.* embedded an FBG in carbon composite material (CCM). Their sensor was around 2.2mm thick, 5mm wide and 40mm long (as shown in the fourth row on Table 2.2) and a pressure of up to 400kPa was applied in 50kPa increments. Their sensor was created to measure plantar pressure and the sensor was placed on the sole of a pair of shoes. Their sensor utilised the diaphragm approach as it was bent where the FBG was located. Their sensor showed a sensitivity of 1.3pm/kPa (Suresh *et al.*, 2015). Unlike the other sensors, this sensor was very long but also very thin. The thickness of this sensor would be suitable for a sensor on the ETT cuff, however the length would need to be altered. In terms of pressure, like the previous papers mentioned, very large pressures were used. The resolution of this sensor is 0.077pm; which is a suitable resolution for a tracheal pressure sensor. However, by altering the length of the sensor this resolution and sensitivity may also change. Additionally, as the smallest pressure applied was 50kPa, it is difficult to assess how the sensor performs at lower pressures like those experienced in the trachea.

Liu *et al.* created a pressure sensor embedded in epoxy resin. The sensor was 20mm long, 10mm wide, and 2mm thick. On top of this was another epoxy resin layer of the same dimensions containing a pulse oximeter embedded in it. This made the sensor 4mm thick in total. Their sensor can be seen in the fifth row of Table 2.2. A pressure of up to 25kPa was applied to the patch on a weighing scale in increments and then unloaded in the same increments. This was done 3 times. From this, a pressure (kPa) vs wavelength shift (pm) curve was created and an empirical formula for pressure calculation was created. The sensitivity of the sensor is not mentioned, however, from looking at the calibration curve displayed it can be estimated at around 6.6pm/kPa (Liu *et al.*, 2018). Although the pressure increments are not specified, the provided graph suggests an increment of around 2.5kPa. This indicates that Liu investigated much lower pressures than the other studies, focusing on pressures relevant to tracheal applications. However, a smaller pressure increment would have been preferable, as it would provide more data points within the pressure range of interest. Additionally, the sensor's resolution is approximately 0.015kPa, which is suitable for tracheal applications. This suggests that epoxy may be a suitable material for low-pressure sensing.

Table 2.2 shows a summary of patch materials used by different researchers and the sensitivities they achieved.

Table 2.2: Comparison of pressure sensor sensitivity and patch material used in different publications.

Publication	Patch Material	Sensitivity	Sensor
Nguyen <i>et al.</i> (2022)	PLA	7pm/kPa	 Bare FBG inside oval-shaped structure
Hong <i>et al.</i> (2019)	PLA	0.0756 pm/kPa	 (a) Inner sensing unit External shell FBG sensor Optical fiber cable
Zheng & Zhang (2005)	Silicone Rubber	14.92 pm/kPa	 Pressure sensing hole Fiber Bragg grating LHFGB RHFBG Polymer Round plate
Suresh <i>et al.</i> (2015)	CCM	1.3 pm/kPa	 FBG Force Composite Prepreg Effective length a D_{max}
Liu <i>et al.</i> (2018)	Epoxy resin	Approx. 6.8pm/kPa	 10mm Receive POF 45° 0.5mm 0.5mm 2mm Skin Epoxy Resin - Vitralit 1655 Epoxy Resin - NOA 65 (Adhesive Layer) Epoxy Resin - NOA 65 2mm FBG1 FBG2 20mm Stainless Steel Tube

It is worth noting that although these papers were measuring pressure, they all (with the exception of Liu) tested with pressures well above 100kPa. These pressures are much larger than those experienced in the trachea (20 cmH₂O to 30 cm H₂O which is

equivalent to 2-3 kPa). It is clear that different materials have different properties and therefore produce different pressure sensitivities. Additionally, different geometries and sizes produce different sensitivities as shown by Nguyen and Hong who both used PLA to embed their fibres but observed very different sensitivities (Nguyen *et al.*, 2022, Hong *et al.*, 2019). It is challenging to choose a material based on these publications as the pressure ranges are much higher than those of interest in this project making it difficult to determine how they would perform at lower pressures (up to 5kPa). Additionally, the sensor sizes were large; much too large to be placed on an ETT cuff. However, as these patch sensors can be quite simple to fabricate, this is a parameter that can be altered to better suit the tracheal application.

Overall, the epoxy/silicone sensors produced a higher sensitivity than the PLA and CCM sensors. In addition to this, Liu's epoxy resin sensor was used to measure lower pressures and therefore provides some validity that this can be achieved with an epoxy resin sensor.

2.4.5 FBGs Used for Pressure Sensing in ETT

Correia *et al.* created a pressure sensor by embedding an FBG in an epoxy patch. A pressure of 59.47kPa was applied in increments of 25g. The sensor was loaded and then unloaded in 2 cycles and the sensitivity was measured. Correia found that calibrating the sensor was quite difficult as the sensor may not have had time to relax before the second cycle. This resulted in the second cycle not reaching baseline when completed. The sensitivity of the sensor was calculated to be 10pm/kPa. The sensor was then placed on an ETT cuff and into a tube. Figure 2.30 shows the sensor configuration.

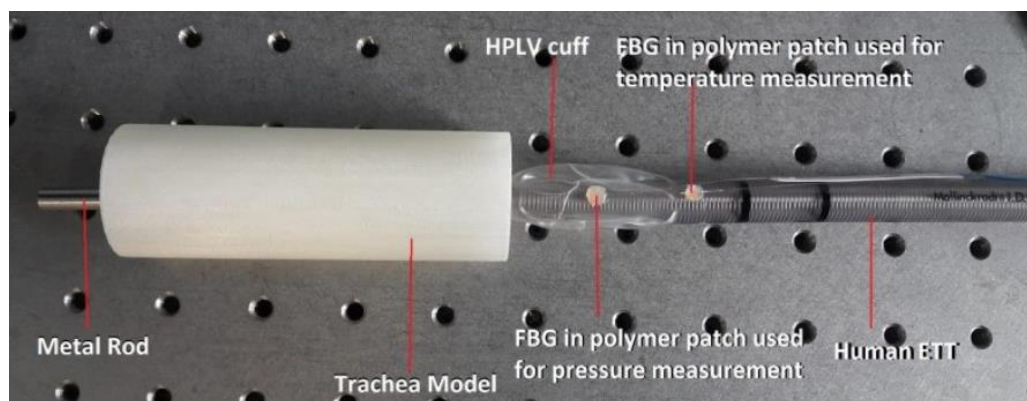


Figure 2.30: Photograph of the ETT containing FBG sensors and 3D printed model of the human trachea (Correia *et al.*, 2016).

The cuff was then inflated and deflated and readings from the manometer were compared to those from the FBG sensor. The sensitivity of the sensor had increased to 20pm/kPa inside the tube (Correia *et al.*, 2016). When Correia calibrated his sensor, it had a sensitivity of 10 pm/kPa, providing a sensor resolution of 0.01kPa, which is suitable for a tracheal sensor. However, when the sensitivity increased to 20 pm/kPa in a tube, the resolution increased to 0.005kPa. This high resolution could cause the sensor to pick up unwanted noise. Additionally, although Correia's sensor includes a temperature reference FBG for temperature compensation, it lacks an FBG for strain compensation, leaving it unclear whether these sensitivities are influenced by unwanted strain. Furthermore, no repeatability tests were conducted. Correia's sensor is very small and made of epoxy, which is not a rigid material, thereby reducing the risk of tracheal injury. Overall, this sensor is a promising starting point for tracheal pressure measurement.

Correia *et al.* further developed this sensor by creating a 1mm thick, 4mm x 4mm sensor using the same method as above, where the FBG is embedded in the same epoxy resin patch. The pressure sensitivity of this sensor is not mentioned, however, they calibrated the sensor using the same method as Liu *et al.* who calibrated the sensor using weighing scales (Liu *et al.*, 2018). Correia *et al.* then performed experiments in an animal study where 6 pigs were intubated and pressure readings were taken. To protect the sensor, it was placed between a double cuff. Figure 2.31 shows the sensor diagram and placement on the ETT cuff.

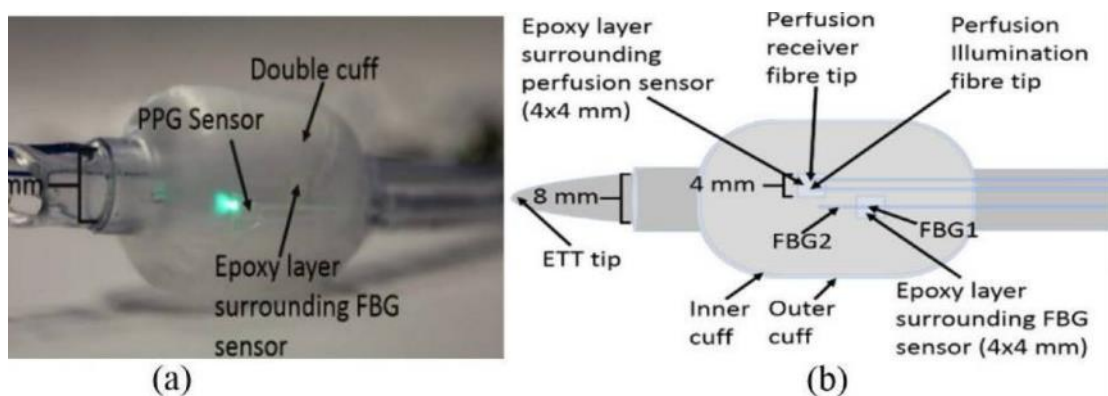


Figure 2.31: ETT illustrating the double cuff housing the sensors (a) Photograph. (b) Schematic (fibres in blue placed between inner and outer cuff). FBG1 and FBG2 represent the pressure and temperature reference FBGs respectively (Correia *et al.*, 2022).

The ETT cuff was placed into the trachea of the pigs and inflated (between 20-80 cmH₂O) using a manometer. Measurements were taken over several minutes and then

the cuff was deflated, rotated 90°, and then inflated again. This was done 4 times to cover the 4 sides of the trachea (Correia *et al.*, 2022). Results showed that the FBG contact pressure sensor correlated with the intracuff pressure from the manometer and is dependent on the orientation of the trachea. Interestingly, it was discovered that the anterior of the trachea (the cartilage side directly opposite the oesophagus) provided low correlation between the intracuff pressure and the contact pressure. Furthermore, the posterior of the trachea (side with no cartilage and closest to the oesophagus; described in Chapter 2.1.3.1) produced lower contact pressures than the other sides. This is probably due to the absence of the cartilage on this side of the trachea (Correia *et al.*, 2022). Like Correia's previous sensor (in the publication described before this), this sensor did not contain a strain reference FBG and therefore it is difficult to tell if their measurements contained unwanted strain measurements. Additionally, the repeatability and reliability of the sensor is not mentioned. Furthermore, Correia's sensor was calibrated according to Liu's calibration method which involved weighing scales. It is important to consider that calibrating a sensor on weighing scales (hard and flat surface) for use in a trachea (soft tissue with curvature) may not be the best method for sensor calibration and could cause sensing inaccuracies. Although Correia's sensor is a promising starting point for tracheal contact pressure measurement, improvements are warranted to test for sensor repeatability. It is also important to calibrate the sensor in a more suitable environment for tracheal contact pressure measurement and to test if strain has an effect on sensor readings as the inflation of the ETT cuff may cause straining of the fibre.

Overall, this animal study shows how intricate the trachea is and how current pressure measurement techniques need replacing. Further experimental testing of the sensor needs to be done to more accurately calibrate with even lower pressures (less than 25kPa) to determine and improve the sensitivity. It is also important to appropriately calibrate the sensor for tracheal measurements. This may be achieved with a tube-like structure to better and more accurately determine contact pressures on the tracheal wall. A method for better fabrication is also required to ensure sensors can be recreated repeatably.

2.4.6 Conclusion

Many FBG-based pressure sensors of various designs, materials, and sizes have been created and tested, showing potential for diverse applications. Research indicates that these sensors can be modified to meet specific needs. Various configurations of FBG sensors, such as those using diaphragms, springs, or bellows, have been explored for pressure sensing, each with varying degrees of sensitivity and resolution. While some designs demonstrate high sensitivity, many are too bulky or made from materials unsuitable for in-vivo applications. Modifying these sensors through miniaturisation or selecting more appropriate materials could enhance their suitability for tracheal applications. However, these alterations may impact their sensing sensitivity and resolution.

In the case of ETT cuff pressure sensors, initial research shows great promise for using FBG pressure sensors for tracheal contact pressure sensing. However, these sensors were calibrated using weighing scales, which raises the question of whether this is the best method for sensors intended for tracheal use. Calibrating on a hard, flat surface may not accurately reflect the sensor's performance in a soft, tubular structure like the trachea. It is essential to calibrate these sensors in a setup that more closely resembles the trachea.

Additionally, the sensors were tested at pressures higher (20-80 cmH₂O) than those typically used in the trachea (20-30 cmH₂O). This raises concerns that the sensors may not perform well at lower pressures. Furthermore, although these sensors incorporated temperature reference FBGs, they did not include a strain reference FBG, nor were repeatability or reliability tests shown in the publications. A deeper study into the performance of these sensors is needed, along with further investigation into how they can be optimised for their intended application.

2.5 Conclusion

In conclusion, the mechanical properties of the trachea, the factors influencing high pressures, and the fact that intracuff pressures change over time, highlight the need for an ETT cuff contact pressure sensor that can effectively control intracuff pressures and ensure patient safety.

To address these challenges, the study proposes the development of a miniature fibre optic pressure sensor to be integrated into the ETT cuff. This sensor will play a crucial role in an automated feedback loop system, allowing for precise monitoring and adjustment of contact pressures on the tracheal wall.

While various pressure sensors exist for contact pressure measurement, FBGs stand out due to their biocompatibility, small size, multi-point sensing capabilities, and high sensitivity without the risk of electrocution. They are also resilient to harsh environments, and immune to electromagnetic interference.

However, it's essential to consider the calibration and sensitivity of such sensors for tracheal use. The calibration method used in previous research, involving hard and flat surfaces, may not accurately represent the tracheal environment. Calibrating sensors in a setup resembling the trachea's conditions is paramount for reliable measurements.

Furthermore, the sensors were tested at pressures higher (20-80 cmH₂O) than those typically used in the trachea (20-30 cmH₂O) and the lack of strain reference FBGs, repeatability, and reliability tests raise concerns about sensor performance at lower pressures. A more comprehensive study is warranted to evaluate and optimise these sensors for tracheal applications.

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Chapter 3 – Embedding Material Selection Process and ABAQUS Model

3.1 Introduction

This chapter discusses recreating a sensor from the literature for contact pressure measurement on the trachea and describes its shortcomings. It then investigates testing different materials to determine an adequate embedding material for the FBG in order for low pressure detection and measurement. These initial tests were carried out by sandwiching the FBG between two material samples. Alongside this, an FEA model was created from the experimental parameters and a comparison was made. The materials performing best in these initial tests were shortlisted for further testing.

3.2 Epoxy as a Sensing Material

3.2.1 Introduction

An epoxy sensor was created according to previous research (Correia *et al.*, 2016) to provide a baseline for comparing improvements in sensor performance using different sensing materials and configurations. While the literature review covered various materials such as PLA, silicone, and CCM, epoxy was chosen as the starting point because the existing literature specifically demonstrates its use on the ETT cuff for contact pressure measurement in the trachea. By recreating the epoxy sensor, it is possible to identify areas for improvement and optimisation.

3.2.2 Methods

3.2.2.1 Fabrication Process

To fabricate the pressure sensor created by Correia *et al* (Correia *et al.*, 2016), three 3mm of the fibre coating was removed with spacing 30mm and 15mm and FBGs with reflection (~70%) wavelengths of 1541nm, 1550nm, and 1555nm were inscribed on the photosensitive fibre (ps1250, Fibrecore) using a UV laser (frequency quadrupled Nd:YAG laser emitting at 266nm) using phase mask technique. The FBGs were not recoated. Figure 3.1 shows the FBG spacing.

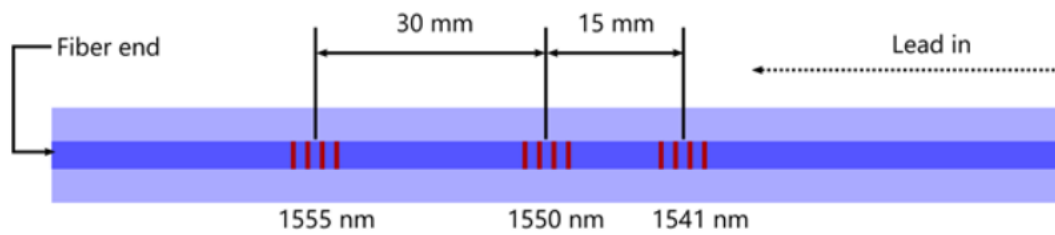


Figure 3.1: The FBG specification showing Bragg wavelength and spacing.

A 4mm x 4mm x 1mm patch was made around the middle FBG using epoxy (Vitalit 1655, Panacol-Elosol GmbH). The FBG on the distal end of the fibre was used for temperature compensation (as it was towards the fibre end, it was left free and not anchored- allowing for temperature measurement without the presence of strain), and the FBG on the proximal side was used for strain control. It is worth noting that the epoxy used in Correia *et al*'s paper was NOA65 (NOA65, Norland Product Incorporated) (Correia *et al.*, 2016). By looking at the technical datasheets of both these epoxies NOA65 is harder than Vitalit (NOA65: 50 Shore D, Vitalit 1655: 70-80 Shore

A) and Young's Modulus (NOA65: 137MPa, Vitralit 1655: 42-46MPa). As hardness and Young's Modulus are indicators of how deformable a material is, this shows that Vitralit is more deformable than NOA65 and therefore should yield better results in comparison to NOA65. The mechanical properties from both datasheets are displayed in Appendix A.

To create the sensor, a 4mm x 4mm x 1mm PTFE mould was cleaned using isopropyl alcohol (2-Propanol $\geq 99.5\%$, Fisher Chemical). The fibre was placed so that the middle FBG (to measure pressure) sat inside the square mould and stuck in place with tape at either end of the fibre outside the mould. About 3 drops of epoxy was added. A UV torch (SDO-UV 365, SiDiOU Group) centred at 365nm wavelength illuminated the top surface for 30 minutes to assist the curing of the epoxy. The patch was then left in the mould overnight to ensure a complete cure. This cure time was determined based on the direction on the Vitralit datasheet. The apparatus used to create the sensor can be seen in Figure 3.2. As the PTFE mould is an open mould, the patch cures with a slight bulge where the mould is open. Once cured, the patch was then polished using a steel file to create a flat top surface and then removed from the mould. This ensured all sensors were of the same height and had parallel sensing surfaces. Figure 3.3 shows the completed and removed 4mm x 4mm x 1mm pressure sensing patch.

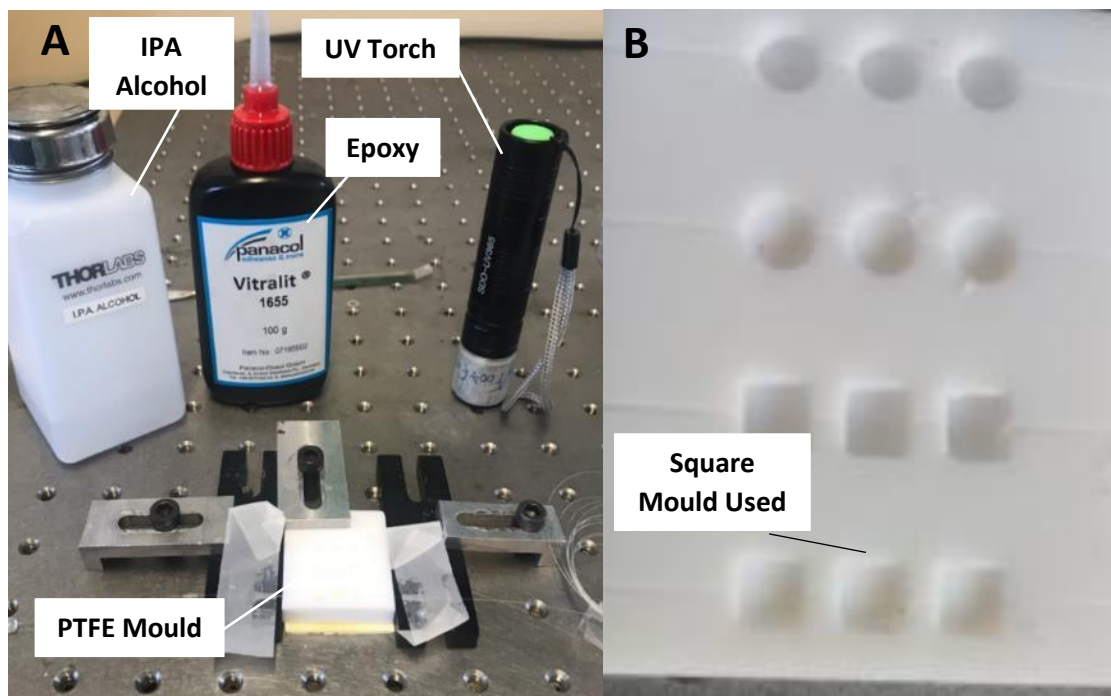


Figure 3.2: A: Equipment used to make epoxy patch on fibre. IPA Alcohol, Vitralit 1655 (epoxy), UV torch, and 4mm x 4mm x 1mm PTFE mould secured to laser table. B: Close up of the PTFE mould.



Figure 3.3: A 4mm x 4mm x 1mm cured Vitralit 1655 pressure sensing patch.

3.2.2.2 Calibration Process

To calibrate the pressure sensor a stepper motor loading device was used. Figure 3.4 shows the experimental setup.

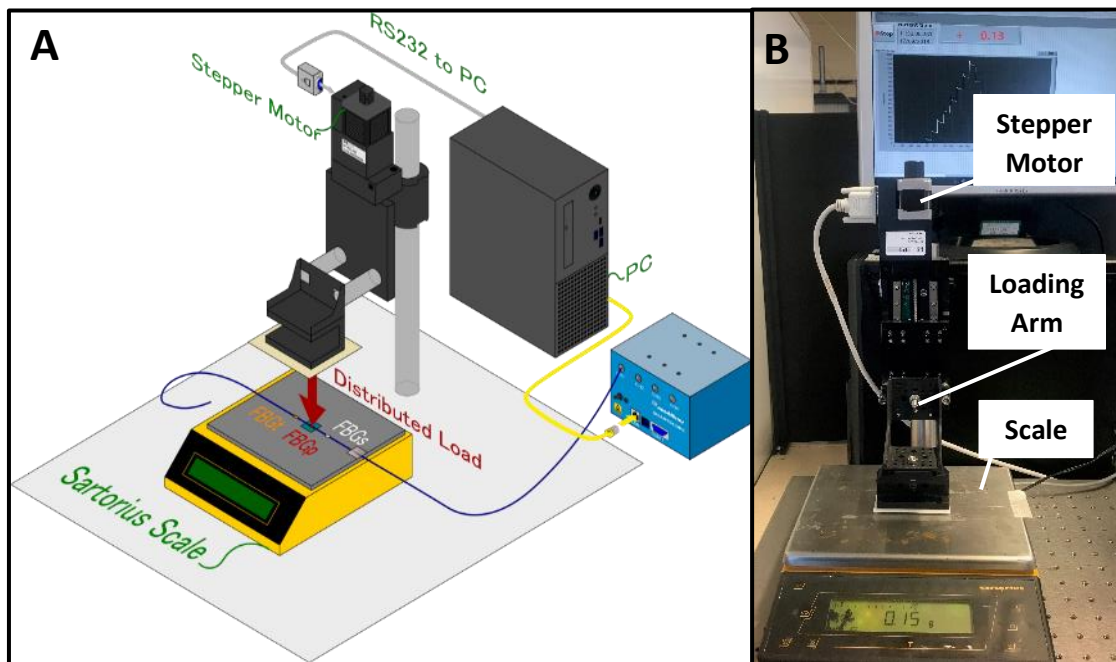


Figure 3.4: A: Diagram of experimental setup for pressure sensor calibration. B: Calibrating pressure sensor (implemented setup from 'A').

The pressure sensor is placed on the surface of highly sensitive scales (48kg x 0.01g, LC 4800P, Sartorius Laboratory). The optical fibre pressure sensor is connected to an optical interrogator with a wavelength range of 40nm (1528 to 1568nm) and a resolution of 0.1pm (SmartScope, Smart Fibres), which is connected to the PC. The stepper motor loading device is placed directly on top of the patch sensor at a distance so that it isn't touching the patch. This loading device was moved down incrementally onto the patch until the desired weight was seen on the scales. The motor loading device was then removed from the patch using the same incremental value but in the reverse direction (upwards) so that a calibration curve could be drawn for both loading and unloading patterns. Load increments and decrements were not performed until the wavelength from the sensor had stabilised.

To calculate the pressure from the weight applied, equation 3.1 was used.

$$P = F/A \quad (3.1)$$

Where P is pressure, F is force calculated by equation 3.2, and A is the area of the patch calculated to be $1.6 \times 10^{-5} \text{m}^2$ (4mm x 4mm).

$$F = mg \quad (3.2)$$

Where m is the mass shown on the scale in kilograms, and g is the gravitational constant 9.81m/s^2 .

Both the scales and the optical interrogator have data logging features to see how values change over time for data analysis. The sampling rate for the interrogator was 0.08 s^{-1} , whereas the scales was 0.1 s^{-1} .

3.2.2.3 Sensitivity Calculation

Due to material elasticity and friction of epoxy on the scale surface, the loading and unloading patterns may show some hysteresis. This is because when the load is applied by the loading arm, it forces the epoxy to expand and surface adhesion may occur, hindering the materials ability to return to its original shape when the loading arm is removed. To calculate the sensitivity a piecewise linear plot was created for the results showing non-linearity. The piecewise linear plots include lines of best fit for both the loading and unloading patterns. To calculate the sensitivity of the sensor, the average of the loading and unloading sensitivities was calculated and the hysteresis error was calculated.

To calculate the hysteresis error, the wavelength shift values for the loading response need to be subtracted from the wavelength shift values from the unloading response (assuming the pressure values for both loading and unloading cycles are the same) the absolute maximum difference between the loading and unloading cycles is considered the hysteresis error (*e-Handbook of Statistical Methods*, Nist-Sematech). This is shown in equation 3.3 where lambda is the wavelength in nm.

$$Hysteresis = \text{Max} | \Delta\lambda_{(loading)} - \Delta\lambda_{(unloading)} | \quad (3.3)$$

As this experimental setup utilised a loading arm, the loading and unloading pressures are not identical and therefore this direct approach will not work. Due to this, to find the hysteresis error, linear regression of loading and unloading cycles was utilised.

3.2.3 Results and Discussion

The first optical fibre pressure sensor fabricated did not have the temperature and strain reference FBGs. It contained one FBG embedded in an epoxy patch (Vitalit 1655, Panacol-Elosol GmbH). To calibrate this sensor, the setup described in Section 3.1.2.2 was used. The loading device increment was selected as 0.1mm until a load of 1.4kg was shown on the scales. Figure 3.5 shows the results of the calibration. Figure 3.5a shows both the wavelength (in nm) over time and the weight applied on the patch (and scale) over time. Figure 3.5b shows the calibration curve of the sensor. This was done by taking the average of the last 50 readings (5 seconds) of the stabilised wavelength and the scale at each increment.

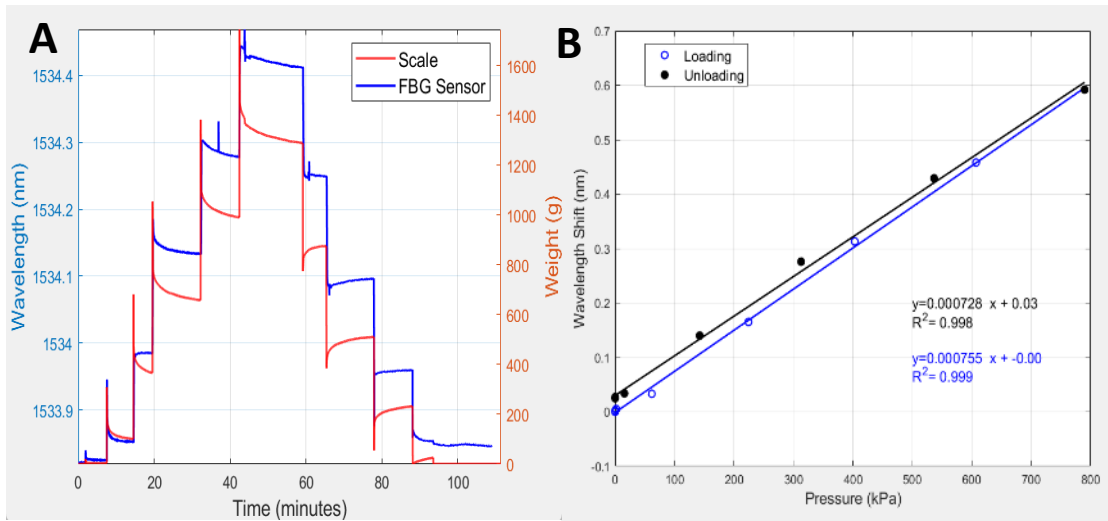


Figure 3.5: Vitalit 1655 presented data. A: wavelength vs time (from interrogator) and weight vs time (from scale), B: calibration curve.

When looking at the scale results (in red) in Figure 3.5a, it can be seen that the scale takes time to stabilise and, in some cases, doesn't stabilise even when the FBG has. This is thought to be due to the way the scales measures weight using electromagnetic force compensation (EMFC). This is something to be monitored and could be a cause for inaccurate sensitivity measurements. The FBG sensor (in blue) responds to the force applied, however, the sensor's wavelength does not go back to its original value at the end of the experiment. This could be due to a temperature change in the room or an unwanted strain on the fibre. As this sensor does not contain reference FBGs it is not

possible to tell what causes this. From Figure 3.5b, the sensor shows a linear performance ($R^2 = 0.998$) with a calculated sensitivity of approximately 0.74pm/kPa with a hysteresis error of 21pm. However, it is important to point out that as the patch area is $1.6 \times 10^{-5} \text{m}^2$, small weights produce high pressure. The 1.4kg applied created a pressure of around 800kPa. As mentioned in Chapter 2.1, the pressure values of interest are far below 100kPa. It can be seen in Figure 3.5b that changes in wavelength below 100kPa can't be determined. For this reason, the experiment was repeated using smaller increments on the loading device until 500g was recorded on the scale. Figure 3.6 shows the results.

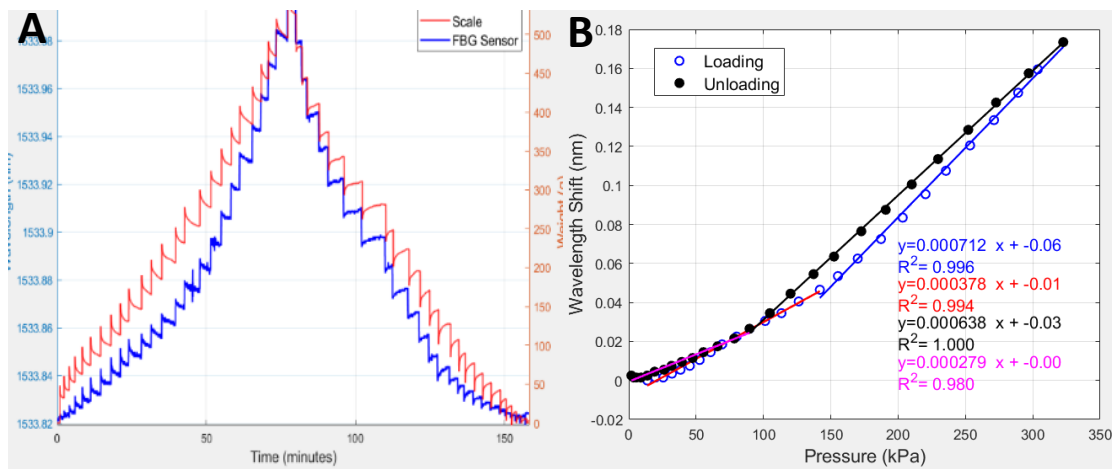


Figure 3.6: Vitalit 1655 presented data. A: wavelength vs time (from interrogator) and weight vs time (from scale), B: Piecewise linear plot showing calibration curve (pressure vs wavelength shift).

Figure 3.6a shows that even with lower weights, the wavelength increases as the weight increases. However, no wavelength shift was experienced on the sensor until 22g (13.73kPa) was placed on the patch. Figure 3.6b shows a non-linear response where the sensitivity decreased from 0.74pm (from Figure 3.5b) to 0.675pm/kPa with a hysteresis error of 25pm. However, it is clear from this graph that for pressures below 100kPa the sensitivity is greatly reduced to 0.33pm/kPa with a hysteresis error of 4.7pm. This means that this sensor is less sensitive to pressures in the range of interest.

This experiment was then repeated with the same sensor to see if the calculated sensitivity was reliable. This time using larger loading increments until 500g was shown on the scales. Figure 3.7 shows the results.

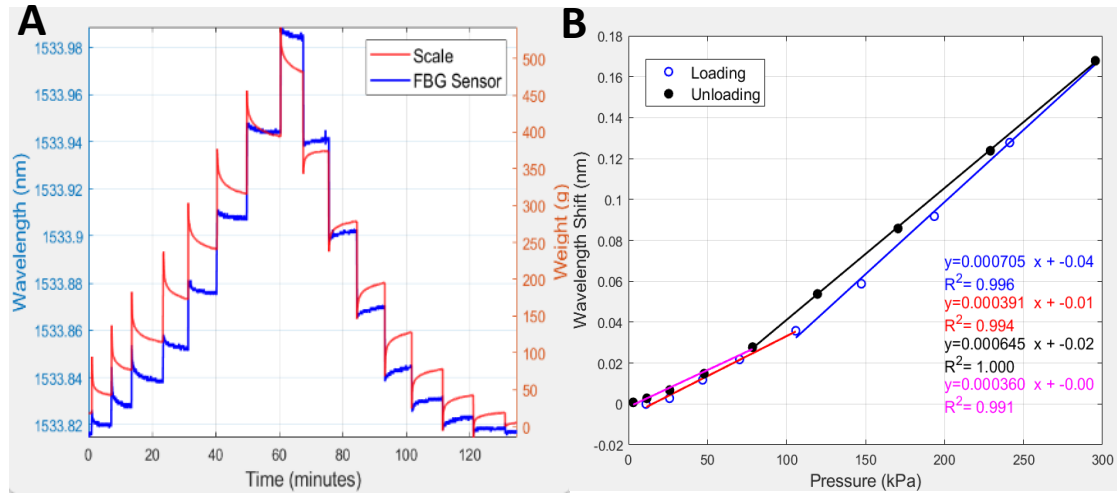


Figure 3.7: Vitalit 1655 presented data. A: wavelength vs time (from interrogator) and weight vs time (from scale), B: calibration curve.

Figure 3.7a shows the same problem occurs with the scale values; they do not stabilise when the FBG has. In terms of FBG response, the wavelength returns to baseline and although there was no wavelength shift observed in the previous experiment until 22g was placed, this experiment noticed a shift once 18g (11.17 kPa) was placed on the sensor. This could mean that the epoxy doesn't deform enough to cause axial strain on the fibre with these small loads. Figure 3.7b also shows a non-linear response and the calculated sensitivity of the sensor is 0.67pm/kPa with a hysteresis error of 1.5pm. which is very close to the 0.675pm/kPa from the previous experiment. Again, the sensitivity for pressures below 100kPa is greatly reduced; this time with a sensitivity of 0.375pm/kPa and a hysteresis error of 9.9pm.

Clearly, this sensor is less sensitive to lower pressures. This can be due to the nature of the epoxy material not expanding as much to lower pressures. Another important point is that the patch material doesn't induce a wavelength shift until about 18g have been placed onto it. This 18g corresponds to a pressure of 11.17kPa and is within the pressure range of interest. This could mean that there is very little deformation in the epoxy when loads less than 18g are applied and this cannot be registered by the FBG. The sensor's functionality relies heavily on a materials Poisson's ratio as the expansion of the material perpendicular to the applied force is what induces axial strain onto the fibre. The Poisson's ratio of epoxy, ranging from 0.3 to 0.35 (Tech tip 19, Epoxy Technology), suggests that it does not expand sufficiently along the fibre under lower loads and requires larger loads to do so effectively.

As this sensor did not contain the reference FBGs a new sensor was fabricated with 3 FBGs as described in Chapter 3.2.2.1. One as a temperature reference, one embedded in a patch for pressure sensing, and one as a strain reference. After the new sensor was fabricated, it was then calibrated (using the same setup as in Figure 3.4). Figure 3.8 shows the raw data from the 3 FBGs.

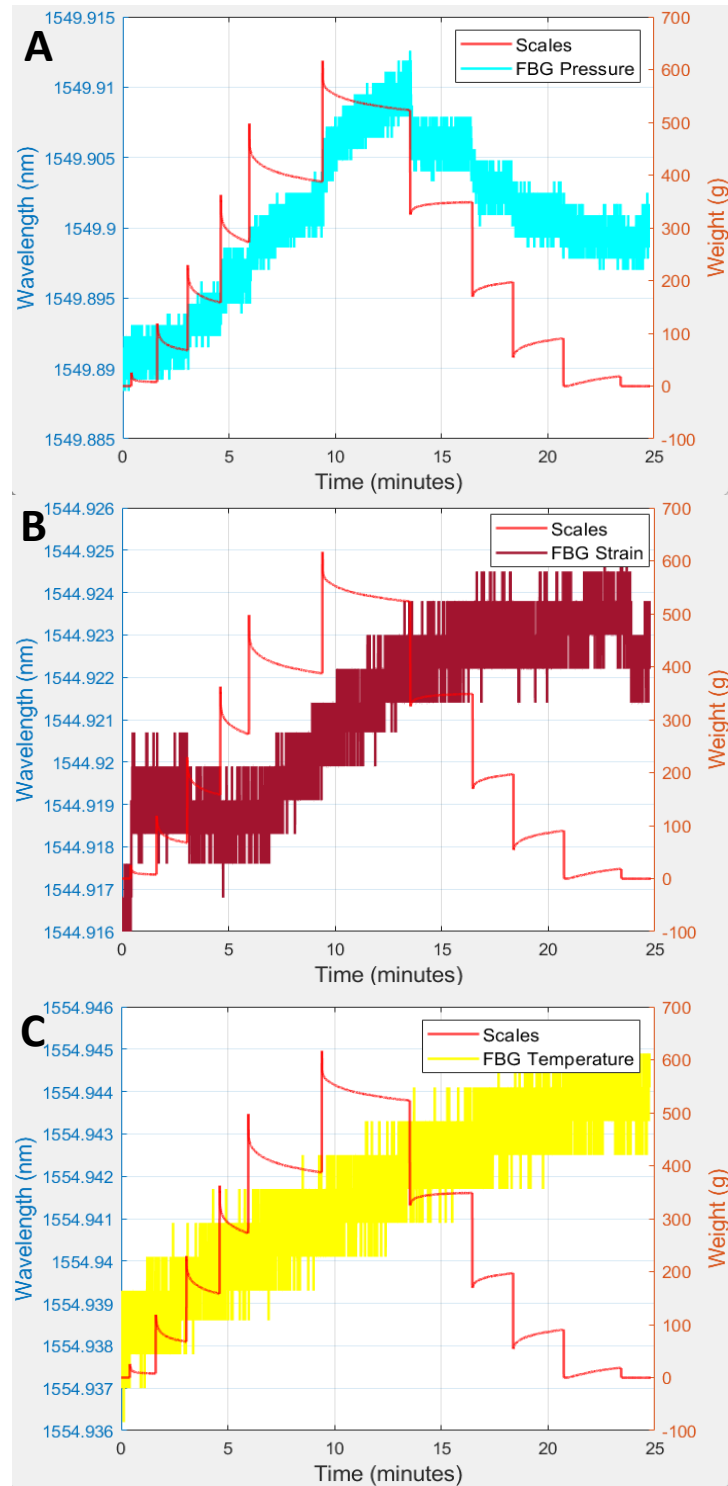


Figure 3.8: Raw response from A: pressure FBG (embedded in Vitralit 1655), B: strain FBG, and C: temperature FBG.

The wavelength shift from baseline was calculated for each of the strain and temperature values. The temperature FBG values (Figure 3.8c) were then subtracted from the strain FBG values (Figure 3.8b) to get the true strain response. The new strain values and the temperature values were then subtracted from the pressure wavelength values. The resultant wavelength values are presented in Figure 3.9. It is worth noting that to accurately compensate for temperature during experiments with temperature fluctuations, it's important to consider the coefficient of thermal expansion of the epoxy. Since these experiments were conducted in the same environment with minimal temperature variations the thermal expansion of the epoxy was not factored into the correction process. However, in scenarios where temperature fluctuations are expected, particularly if the sensors are placed in the human body, the thermal behaviour of the epoxy and its impact on the FBG sensors' readings need to be considered.

It can be seen from Figure 3.8b and 3.8c that the wavelength changes in the strain and temperature FBGs are similar. This is due to the sensor not experiencing much strain in the scale experiment and mainly being affected by temperature. This is due to the nature of the scale experiments where the fibre is suspended as the thickness of the patch holds it in mid-air. The loading arm only comes into contact with the patch and does not touch on the fibre outside the patch.

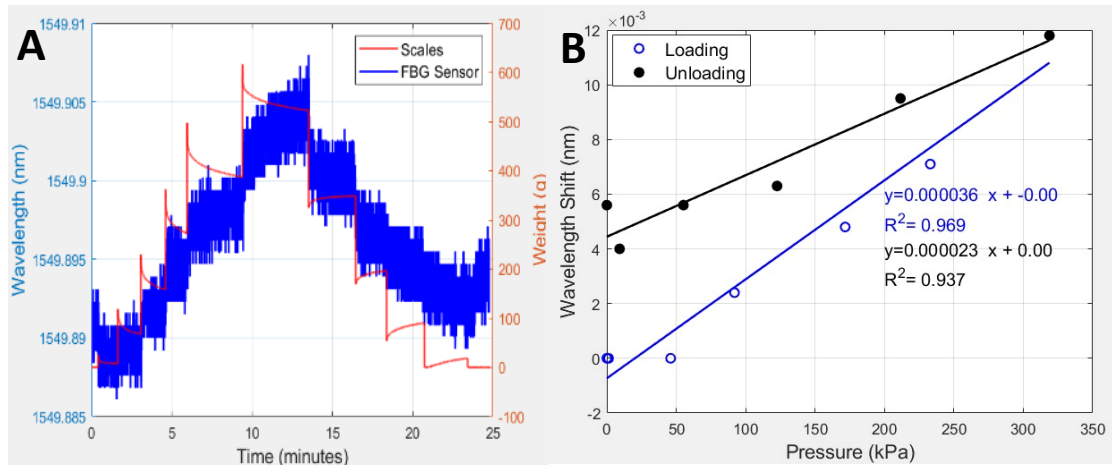


Figure 3.9: A: Vitalit 1655 results with temperature and strain compensation. Wavelength vs time (from interrogator) and weight vs time (from scale). B: calibration curve.

Figures 3.8 and 3.9 show that there is a problem. Although the fabrication process for this sensor was the same as the previous sensor, it seems that the PTFE mould cannot reproduce sensors effectively. Figure 3.9b shows that even though the load applied was more than 300kPa, the FBG wavelength shifted by 0.012nm creating a sensor with a

sensitivity of around 0.0295pm/kPa. This is also seen in Figure 3.8, as none of the FBGs experienced a shift like that seen on the previous sensor. This makes the sensors difficult to compare to each other. This could be due to two possible reasons:

- 1) The polishing of the sensor surface possibly produced an uneven surface for the patch so that the loading device was pressing down on one part of the patch and not the whole patch surface simultaneously.
- 2) The fibre was not in the centre of the mould during fabrication. As the PTFE moulds are open, the fibre is placed in through the slit in the mould and although it is held in place with tape, this does not prevent the fibre from moving up or down within the slit. This can cause sensors to be created with fibres not placed evenly throughout the cured sensing material. Figure 3.10 helps describe this observation.

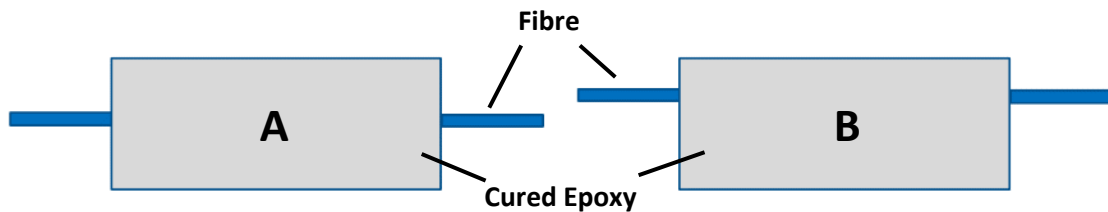


Figure 3.10: Side view of the 4mm x 4mm x 1mm sensors created by the PTFE mould. A: An example of an ideal sensor, where the fibre sits through the centre of the cured patch. B: Some sensors created using the PTFE mould can cure like this, where the fibre is not in the centre of the cured patch.

When the fibre is placed in the epoxy patch like in Figure 3.10a, and a load is applied to the patch; it expands, straining the fibre. The fibre will experience maximum strain as the epoxy will bulge out in the centre of the patch where the fibre is located. When the fibre is placed in the epoxy patch like in Figure 3.10b, the fibre will experience less strain for the same load because the fibre is not located where the patch bulges out the most. This will result in a smaller wavelength shift even though the load applied is the same.

When comparing these results to the target specification in Chapter 2.1.6, these experimental results show the sensor has low resolution, is not sensitive to pressures below 15kPa, and the sensor fabrication is not reproducible. To improve the sensors eligibility for tracheal contact pressure sensing, and to ensure optimal fabrication of sensors, a new fabrication process needs to be considered.

3.2.4 Conclusion

Two contact pressure sensors were developed using epoxy as an embedding material, their sensitivity was tested using a stepper motor loading device and highly sensitive scales. The first sensor contained only one FBG. This meant that its wavelength shift didn't just respond to pressure, but also strain and temperature. Results from this sensor showed that the pressure sensing patch material was not sensitive enough in the desirable range (up to 10kPa). To double check results another sensor was created, this time with 3 FBGs, one as a temperature reference, one for pressure (embedded in an epoxy patch), and one as a strain reference. However, results from this sensor showed that the fabrication method does not recreate uniform sensing patches every time. This sensor also showed that the pressure sensing material was not sensitive enough to read contact pressures of that experienced in the trachea.

Through these initial tests, it is clear that this sensor requires optimising for its application. A new sensing material is needed to improve sensing sensitivity and a new reliable fabrication technique is also required to improve sensor performance and create uniform sensors every time.

In addition to these findings, the scale response is not ideal for calibration use. The weight value showing on the scales after each load increment and decrement does not stabilise and can contribute to inaccurate sensitivity calculations. The scale setup does however provide a valuable insight to how the sensing material reacts to loading and unloading and a general idea of the material's sensitivity. When calibrating for tracheal use, another calibration method will need to be employed, ideally one that more closely replicates that of a trachea.

3.3 Patch Material Selection Process

3.3.1 Introduction

As a new sensing material is required to sense the lower contact pressures of those experienced in the trachea; this sensing material needs to be more deformable to lower pressures (below 10kPa) in order for it to induce more axial strain along the fibre. This requires the sensing materials to have a lower Young's Modulus than Vitralit (42-46MPa) and a higher Poisson's ratio than epoxy (0.3 - 0.35) (Tech tip 19, Epoxy Technology). This would require focusing on silicones, and PDMS materials rather than epoxies. Additionally, a UV curable sensing material is desirable as they generally have simpler curing procedures and have shorter curing times.

In this chapter, a total of seven samples went through initial tests to determine their performance as a sensing material. These tests were setup in such a way that the FBGs were not wasted. The pressure FBG was sandwiched between two pre-made samples and then tested using the loading arm and scale setup described in Section 3.2.2.2. This was done so that the best materials from these tests could go on to the next stage of testing- where the FBG is fully embedded and tested further as a complete contact pressure sensor. Two of the material samples tested (referred to in this Section as Samples 1 and 2) were obtained from the Advanced Manufacturing Facility (AMF) at The University of Nottingham. An additional three samples (referred to in this Section as Samples 2a, 2b, and 2c) were variations of one of the originals sent over from AMF. These were all UV curable samples. These three samples consisted of the same PDMS solution as Sample 2 but with a different number of crosslinks making their elasticity slightly different. The final two samples were bought, one was a PDMS (SYLGARD), and one was silicone rubber (Dragon Skin).

3.3.2 Methods

3.3.2.1 Materials and Methods

Table 3.1 shows all the PDMS materials used throughout this Chapter.

Table 3.1: Summary of materials used in this chapter, their manufacturers, Poisson's Ratio, and Young's Modulus.

Sample	Material Name	Manufacturer	Poisson's Ratio	Young's Modulus
Sample 1	SE1700	Dow Corning	0.48 (Weisgraber <i>et al.</i> , 2018)	0.91 MPa (Lin <i>et al.</i> , 2016)
Sample 2	UV Curable PDMS (as described by Foerster <i>et al.</i> , 2021)	The University of Nottingham	Unknown	Unknown
Sample 2a	UV Curable PDMS (as described by Foerster <i>et al.</i> , 2021)	The University of Nottingham	Unknown	Unknown
Sample 2b	UV Curable PDMS (as described by Foerster <i>et al.</i> , 2021)	The University of Nottingham	Unknown	Unknown
Sample 2c	UV Curable PDMS (as described by Foerster <i>et al.</i> , 2021)	The University of Nottingham	Unknown	Unknown
SYLGARD	SYLGARD 184	Dow Corning	0.495 (Müller <i>et al.</i> , 2019)	2 MPa (Dow Corning)
Dragon Skin	Dragon Skin	Smooth-On	0.4999 (Sparks <i>et al.</i> , 2015)	1.11 MPa (Sandoval <i>et al.</i> , 2019)

As five of the seven samples were provided by the AMF, the only samples that were created in this Chapter were the SYLGARD and Dragon Skin samples. All materials selected had a lower Young's Modulus to Vitralit (42-46MPa) and a higher Poisson's ratio than epoxy (0.3 - 0.35) (Tech tip 19, Epoxy Technology). The Young's Modulus of the PDMS samples provided by AMF were unknown, however, they were still tested as they were UV curable and would have desirable manufacturing (curing) conditions.

The procedure for creating the SYLGARD samples involved mixing the two-part solution at the recommended 10:1 ratio, this was carried out using a syringe at a 20ml:2ml ratio of the two part solution. The resulting solution was poured into a petri dish, and placed in the oven (Model FD 23, Binder) at 55°C for 2 hours to cure. It can also be cured at room temperature overnight.

The procedure for creating the Dragon Skin samples involved mixing the two-part solution as the recommended 1:1 ratio, pouring the solution into a petri dish, and leaving it to cure overnight at room temperature.

3.3.2.2 Patch Material Sandwich Tests

All seven samples were tested using the same method. Two rectangular pieces, approximately 10mm x 8mm x 1mm, were cut from each sample. These were tested using the loading arm and scale setup described in Chapter 3.2.2.2. For the last two samples, the rectangles were cut to 7mm x 5mm x 1mm due to insufficient material to create 10mm x 8mm x 1mm samples.

The first rectangular piece was placed in the centre of the scale, beneath the loading arm. The fibre, containing three FBGs (one for temperature, one for pressure, and one for strain), was placed on top of the rectangular patch. The second FBG (with a Bragg wavelength of 1550 nm, as shown in Figure 3.1) was positioned in the centre of the patch, while the other two FBGs were not in contact with the patch. The second rectangular patch was then placed on top of the fibre, sandwiching the second FBG between the two rectangular patches. This experimental setup is illustrated in Figure 3.4.

The loading arm was gradually moved down onto the patch in small increments until approximately 150g was registered on the scale. The load was then removed from the patch using the same incremental steps in the reverse direction (upwards). This process allowed for the creation of a calibration curve for both loading and unloading patterns. Load increments and decrements were only performed once the wavelength from the FBG had stabilised.

Using equation 3.1 pressure was calculated, where A is the area of the patch calculated to be $8 \times 10^{-5} \text{m}^2$ (for 10mm x 8mm samples), and $3.5 \times 10^{-5} \text{m}^2$ (for 7mm x 5mm samples).

3.3.2.3 Batch and Repeatability Tests

A large SYLGARD sample was created in a petri dish. The same 10:1 ratio (20ml:2ml of the two part solution) was used. This sample was cut into multiple rectangles of 7mm x 5mm x 1mm and tested using the same method as above (Chapter 3.2.2.2) to see if the sample manufacture was repeatable. This was to test if the sensors can be batch produced and have the same sensitivity throughout each sensing patch.

Another experiment was carried out with the SYLGARD sample to test sensor repeatability under cyclic loading. This was to assure that the wavelength baseline returned to its original baseline after each loading and unloading cycle.

3.3.2.4 Statistical Analysis

To assess the accuracy of the sensor, linear regression and test-retest repeatability for repeated/cyclic tests was evaluated using the coefficient of variation (CV):

$$CV = \left[\frac{SD(\text{sensor sensitivities of repeated tests})}{\text{mean}(\text{sensor sensitivities of all repeated tests})} \right] \times 100 \quad (3.4)$$

The CV values were interpreted as poor if $CV > 10\%$, moderate if $10\% \geq CV \geq 5\%$, and good if $CV < 5\%$ (Scott *et al.*, 2015, Santos *et al.*, 2022).

3.3.2.5 ABAQUS Patch Model Simulations

A finite element analysis (FEA) model was created using ABAQUS (version 2020, Simulia) to determine results for one of the materials that was selected for further testing. This was due to labs being in lockdown due to COVID-19. This allowed for the simulation of lab conditions without having to physically test the materials. This could then allow for the investigation of different material properties and parameters without having to manufacture them all.

a) Model design and material properties

The mechanical properties of SYLGARD 184 were utilised in this model. This is because ABAQUS does not allow the simulation of materials with a Poisson's Ratio more than 0.495 and Dragon Skin has a Poisson's Ratio of 0.4999 (Sparks *et al.*, 2015). As the Poisson's Ratio for SYLGARD 184 is 0.495 (Müller *et al.*, 2019), the material properties for SYLGARD 184 were used in the model. Once the model is confirmed to work well, material properties can be changed to simulate the response of other materials. The dimensions used for the patch were those used experimentally in the sandwich batch tests (described in Section 3.3.2.3). As two rectangles were used in an experiment, the dimensions were 7mm x 5mm x 2mm. Below the patch was a steel surface which would act as the scale surface. Running through the patch was a fibre of 1cm length, and on top of the patch was a steel plate made of the same material as the steel surface. This would mimic the way the load is applied to the patch making sure it was distributed uniformly. As pressure in ABAQUS relies on the area of the surface the load is applied to, the plate was made to be same size as the patch (7mm x 5mm).

As the fibre core, fibre cladding, patch, steel plate and steel surface were created as different parts, each had different material properties applied to ensure the simulation would react as expected to the load being applied. Table 3.2 shows the list of parameters used in the model and their values. The coefficient of friction throughout the model was set to 0.64, which is the coefficient of friction between a steel and rubber surface (*Coefficient of Friction*, Engineers Edge).

Table 3.2: Material properties and part dimensions of ABAQUS patch model

Model Part	Young's Modulus (MPa)	Poisson's Ratio	Dimensions
Fibre Core	72000 (Feng <i>et al.</i> , 2010)	0.17 (Wierzbza & Kosmowski, 2003)	9µm x 10mm
Fibre Cladding	72000 (Feng <i>et al.</i> , 2010)	0.17 (Wierzbza & Kosmowski, 2003)	125µm x 10mm
SYLGARD Patch	2 (Dow Corning)	0.495 (Müller <i>et al.</i> , 2019)	7mm x 5mm x 2mm
Steel Surface	200000 (<i>Young's Modulus of Steel</i> , AmesWeb)	0.3 (<i>Poisson's ratio</i> , Engineering Toolbox)	10mm x 10mm x 1mm
Steel Plate	200000 (<i>Young's Modulus of Steel</i> , AmesWeb)	0.3 (<i>Poisson's ratio</i> , Engineering Toolbox)	7mm x 5mm x 1mm

A boundary condition was added to the bottom of the steel surface to ensure that when the load is applied the surface doesn't move. A constraint was added to the core of the fibre to bind it to the fibre cladding and another constraint was added to the fibre to bind it to the patch. The uniform load applied on the plate of the model was a pressure, the pressures were from the experimental pressures calculated in the three batch sandwich experiments. Figure 3.11 shows the created patch model with the boundary conditions and load.

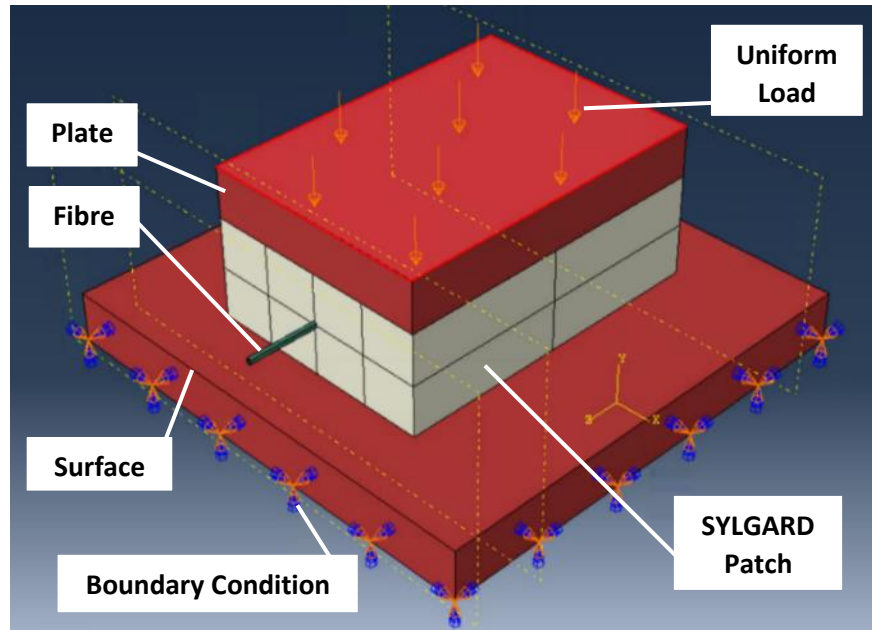


Figure 3.11: ABAQUS model showing load and boundary conditions. The red parts correspond to the material properties of Steel (plate: 7mm x 5mm x 1mm, surface: 10mm x 10mm x 1mm). The white part corresponds to the material properties of SYLGARD 184 (7mm x 5mm x 2mm), and the dark green parts correspond to the material properties of the fibre (fibre: 125µm diameter x 10mm in length).

b) Refining model mesh

The meshing of the model is of key importance as this will determine how accurate the simulation results are. The finer the mesh the longer the computational time; however, the more accurate the results will be. In ideal simulations the mesh is refined until the result converges. For example, if a mesh of 0.5mm for one part computes a simulated wavelength shift of 1nm for a specific load; then when changing the mesh to 0.2mm, the output should still result in 1nm. By applying a mesh of 0.5mm rather than 0.2mm the computational time is reduced, while the output is just as accurate, thus making the model more efficient.

Table 3.3 shows the mesh refining carried out on this model. The load applied stayed constant through all simulations, and the meshing of both the surface and plate (parts shown in red in Figure 3.11), and the patch (part shown in white in Figure 3.11) were altered. The meshing of the fibre was set at 0.01mm.

Table 3.3: Mesh refinement of ABAQUS patch model.

Mesh Sizing of Parts (mm)		Simulation Results (nm)
Plate and Surface mesh:	0.2	0.0576
Patch mesh:	0.1	
Plate and Surface mesh:	0.25	0.0576
Patch mesh:	0.1	
Plate and Surface mesh:	0.3	0.0576
Patch mesh:	0.1	
Plate and Surface mesh:	0.3	0.0583
Patch mesh:	0.15	
Plate and Surface mesh:	0.5	0.0576
Patch mesh:	0.1	
Plate and Surface mesh:	0.5	0.0568
Patch mesh:	0.08	

When refining the mesh it is clear that the meshing of the plates and surface does not influence the wavelength shift. Therefore, the mesh used for the surface and plate was 0.5mm. The meshing of the SYLGARD patch however clearly affects the wavelength shift. With a size of 0.1, the computational time is 3 hours on a machine with 2.44GHz AMD EPYC 7763 CPU processor and 160GB RAM (via virtual desktop). Using a mesh size of 0.08mm requires 8 hours of computational time and yields a simulation result that deviates by 1.5% from the 0.1mm mesh. However, refining the patch mesh further would require a computational time of over 12 hours and is unlikely to produce a significant difference in the simulated result. Thus, a mesh of 0.08 was employed. Figure 3.12 shows the meshing of the model.

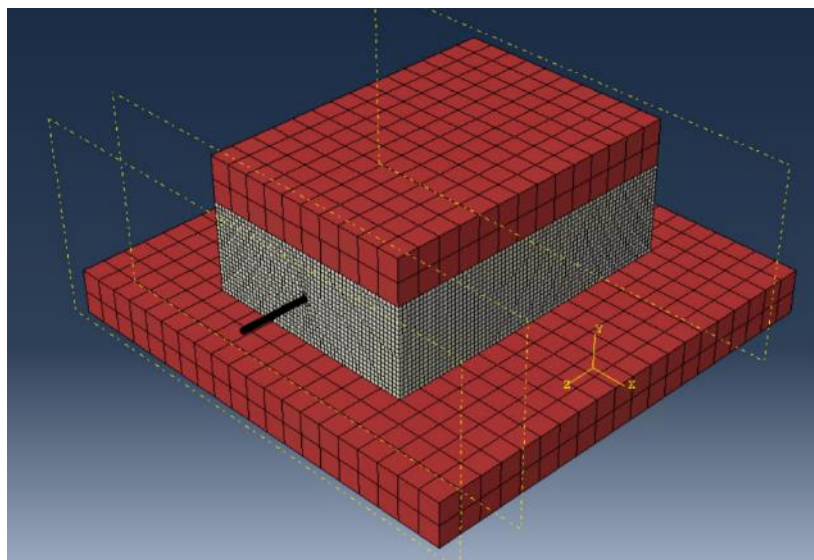


Figure 3.12: Meshed patch model; each red grid box for the mesh of the surface and plate is 0.5mm by 0.5mm. Each white grid box for the mesh of the patch is 0.08mm by 0.08mm. Each dark green grid box for the mesh of the fibre is 0.01mm by 0.01mm.

c) Strain outputs

Once the simulation was run, the resulting model was then analysed by creating a path from the nodes inside the fibre core as this is where the FBG is located. The longitudinal strain values (from the z dimension), of each node along the path, were exported in a text file and imported into a MATLAB script to calculate the wavelength shift experienced when placed under the load. This MATLAB script was created by a previous Nottingham University PhD student to determine the wavelength shift from the average of 300 strain values (resulting in the 3mm length of the FBG) in the centre of the fibre. The wavelength shift was calculated using equation 3.5:

$$\Delta\lambda = (1-p_e) \times \varepsilon \times \lambda_{Bragg} \quad (3.5)$$

Where $\Delta\lambda$ is the wavelength shift in nm, p_e is the photoelastic coefficient based on the material properties of the fibre, ε is the mean strain in the z direction experienced by the FBG (calculated from the 3mm length simulated model results), and λ_{Bragg} is the Bragg wavelength in nm.

Figure 3.13 shows the path (in red) created when selecting the nodes in the fibre core.

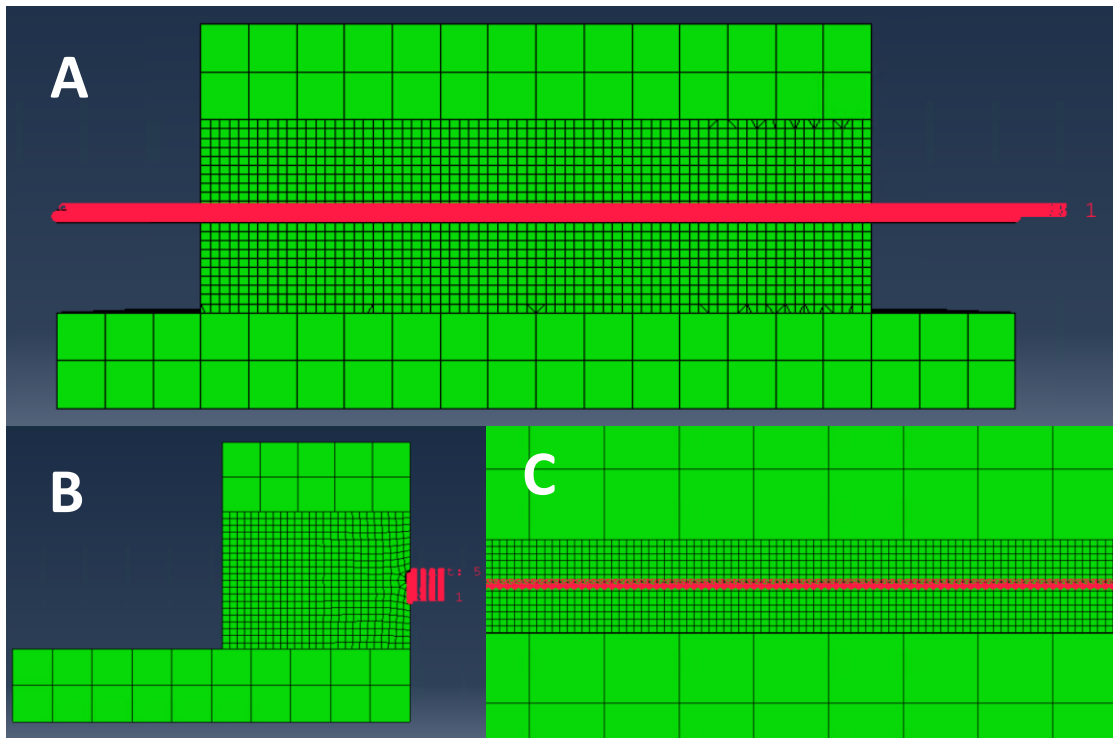


Figure 3.13: ABAQUS path (in red) created using nodes from centre of fibre core. a- the whole model split in half lengthways to access the nodes in the core. b- a better visual of the model split in half to access core nodes. c- zoomed in visual where the larger grid boxes are of the patch (0.08mm x 0.08mm) and the smaller grid boxes are the fibre core.

It is important to consider that transverse loading of a fibre can induce birefringence, leading to changes in the refractive index that may affect wavelength shift measurements. Although the effect might be minor, it is not necessarily insignificant. The simulations do not consider the changes in refractive index caused by transverse loading, which might create discrepancies when comparing simulated and experimental wavelength shifts. However, the effect of birefringence is likely minimal in this context because the applied loads are not large enough to cause significant physical deformation of the fibre.

3.3.3 Results and Discussion

3.3.3.1 Patch Material Sandwich Tests

The first two samples obtained from the AMF were samples 1 and 2 as shown in Figure 3.14.

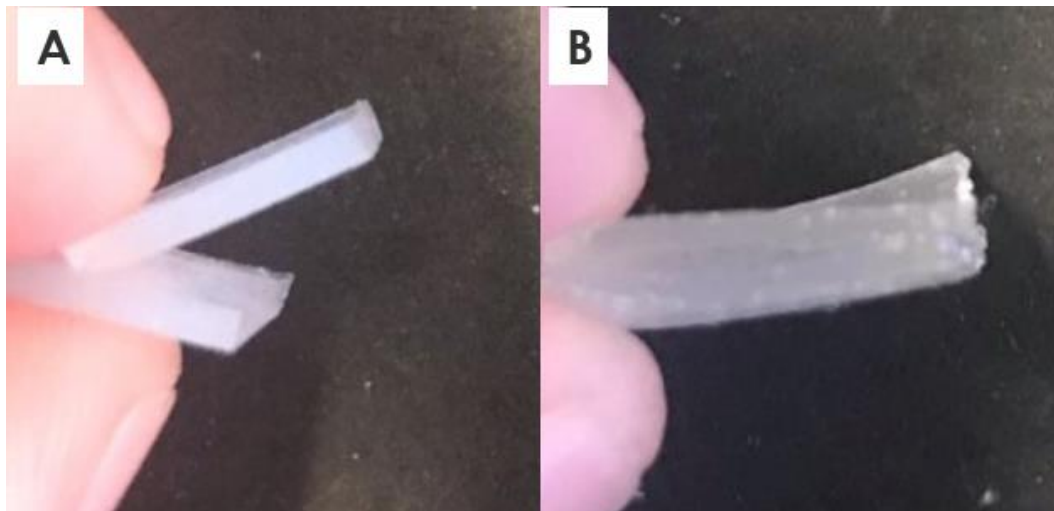


Figure 3.14: Samples from AMF that were tested. A: SE1700 silicone elastomer, B: PDMS elastomer. Each sample is 7mm x 5mm x 1mm.

Both these samples were tested using the setup described in Section 3.2.2.2. These samples were cut into two rectangles of 10mm x 8mm. It is worth noting that these samples were measured using a ruler and it is possible that the rectangles were not identical. This may affect the sensitivity readings, however, this should not affect what is being investigated in these experiments. As these are preliminary tests, the main purpose of these experiments is to assess how the material reacts to loading and unloading as well as to indicate its performance at lower pressures (below 15kPa). Figure 3.15 shows the results of Sample 1 (SE1700).

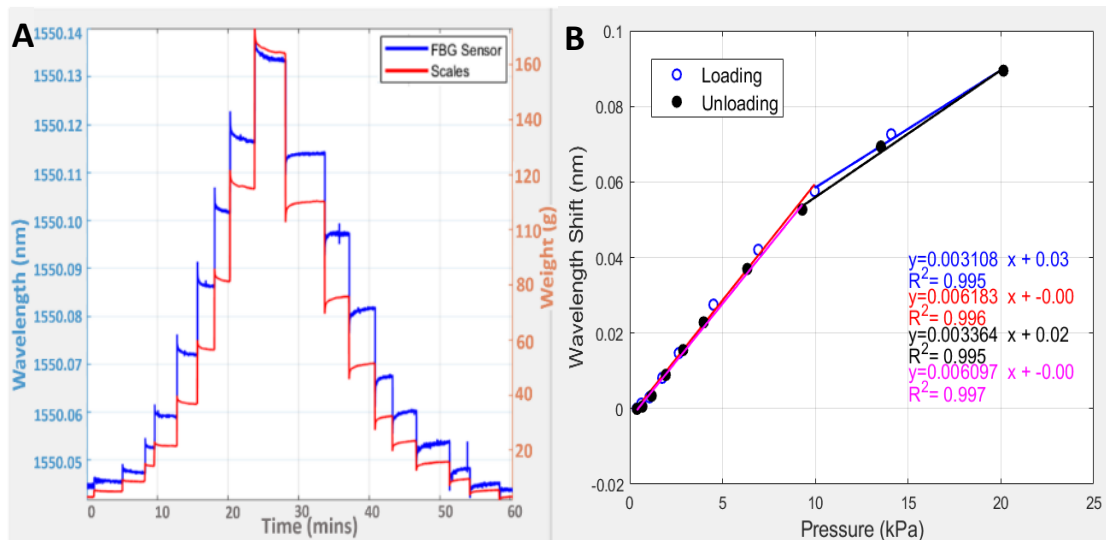


Figure 3.15: Testing results of Sample 1 (SE1700). A: wavelength vs time (from interrogator) and weight vs time (from scale), B: piecewise linear plot showing calibration curve (pressure vs wavelength shift).

Figure 3.15a shows that the wavelength returns to baseline and even records a wavelength shift after 3g is applied. Figure 3.15b shows that Sample 1 (SE1700) has a non-linear response. The sensitivity of the sensor below 10kPa is 6.1pm/kPa with a hysteresis error of 0.85pm; and above 10kPa it is 3.2pm/kPa with a hysteresis error of 7.4pm. For pressures below 10kPa, hysteresis is considerably reduced.

Sample 2 (UV Curable PDMS) was then tested, and Figure 3.16 shows the results.

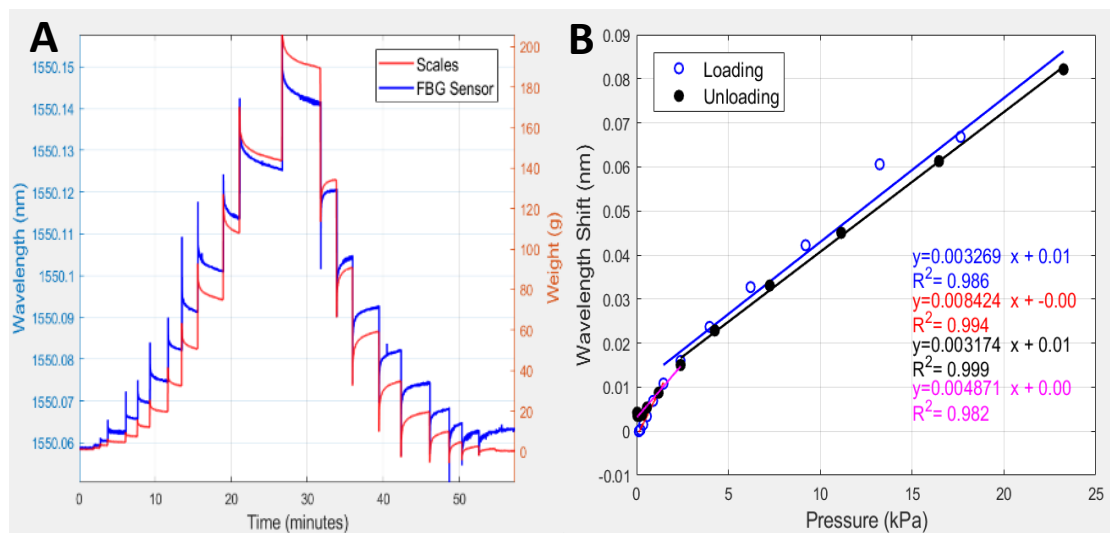


Figure 3.16: Testing results of Sample 2 (UV Curable PDMS). A: wavelength vs time (from interrogator) and weight vs time (from scale), B: piecewise linear plot showing calibration curve (pressure vs wavelength shift).

As shown in Figure 3.16a, the wavelength doesn't return to baseline (a difference of 6.1pm). This is thought to be due to the sample sticking to the loading arm and lifting

up as the loading arm was lifting up off the scales. However, the wavelength shifted once 1g was applied. Despite this, the sensitivity derived from Figure 3.16b for pressures below 2.5kPa is 6.6pm/kPa with a hysteresis error of 4.3pm, and for pressures above that is 3.2pm/kPa with a hysteresis error of 9.8pm. Although this is similar to the previous sample's sensitivity, this sample has a smaller region of low-pressure sensitivity (only for pressures below 2.5kPa). In addition to this, the material was sticky causing surface tension on the scales and sometimes lifting off the scales with the loading arm. When feeling the sample, it was easily deformable (it would bend quite easily and stay bent). This can be seen in Figure 3.14b; there are no straight edges to the material, it has a natural bend.

From these two samples, Sample 1 (SE1700) is clearly more desirable for its smooth loading and unloading transition, its region of no hysteresis in the beginning of the sensitivity graph, and for its non-sticky nature.

AMF then sent over 3 more samples (Samples 2a, 2b and 2c), all the same PDMS material as Sample 2 (UV Curable PDMS), but with each sample containing more crosslinks to reduce its elasticity in hopes to improve its loading and unloading response. The results of these tests are shown in Appendix B.

Overall, these 3 samples did not perform well due to their sticky nature. Increasing the number of crosslinks in the material did not seem to prevent this. This would cause issues as the patches would adhere to the surface making it hard for the patch material to return to its original shape after unloading.

Two more samples were bought to test as a patch material. Samples were made from SYLGARD and Dragon Skin. The results for the SYLGARD sample are shown in Figure 3.17.

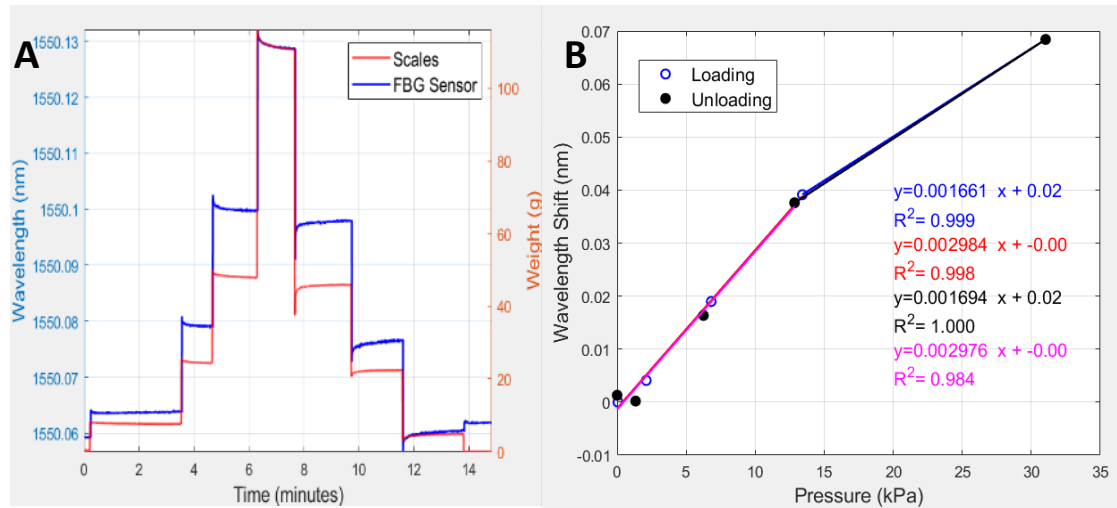


Figure 3.17: Testing results of SYLGARD. A: wavelength vs time (from interrogator) and weight vs time (from scale), B: piecewise linear plot showing calibration curve (pressure vs wavelength shift).

Figure 3.17a shows a smooth transitional response as the load increases and decreases as opposed to the loading and unloading spikes observed in Figure 3.16a. The wavelength and scales can be seen to stabilise within a few seconds possibly allowing for a more accurate sensitivity calculation. The ending wavelength is slightly different from the starting wavelength (by around 1.29pm). This sample, like others, produced a measurable wavelength shift after 1g was applied. Figure 3.17b shows that the loading and unloading curves are almost identical. The sensitivity was calculated as 2.98pm/kPa for pressures below 12kPa with a hysteresis error of 0.11pm and 1.67pm/kPa with a hysteresis error of 1pm for pressures above that.

The results for the Dragon Skin sample are shown in Figure 3.18.

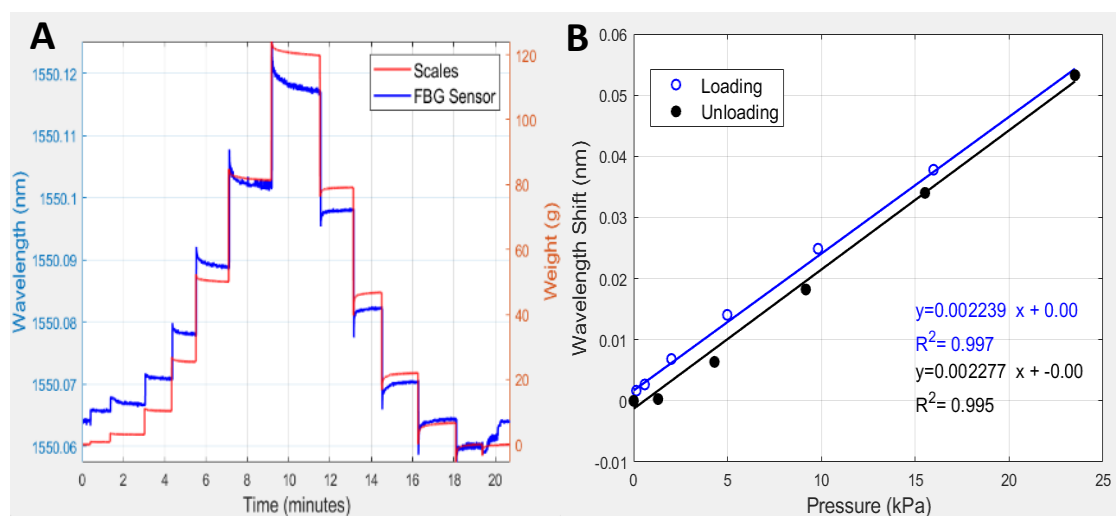


Figure 3.18: Testing results of Dragon Skin. A: wavelength vs time (from interrogator) and weight vs time (from scale), B: calibration curve.

Figure 3.18a also shows a smooth transitional response as the load increases and decreases, like in Figure 3.17a. The ending wavelength (1550.065nm) is similar to the starting wavelength (1550.064nm) despite the wavelength going below what was expected during the experiment. This sample also produced a measurable response when 1g was applied. Figure 3.18b shows that the loading and unloading curves are linear ($R^2=0.99$) but with hysteresis. The sensitivity was calculated as 2.26pm/kPa with a hysteresis error of 0.89pm.

Table 3.4 shows a summary of the samples tested for easier comparison.

Table 3.4: Summary of all the patch material samples tested.

Sample	Wavelength Stabilisation	Observations	Hysteresis Error (pm)	Sensitivity of Beginning Regions (pm/kPa)
Sample 1	In less than a minute	Small loading or unloading spikes	0.85	6.1pm/kPa when <10kPa
Sample 2	Some steps did not stabilise	large loading and unloading spikes	4.3 9.8	6.6pm/kPa when <2.5kPa 3.2pm/kPa when >2.5kPa
Sample 2a	Some steps did not stabilise	no clear wavelength shift to loading	-	-
Sample 2b	Some steps did not stabilise	large loading and unloading spikes	2.6	1.5pm/kPa
Sample 2c	Within a few seconds	large loading and unloading spikes	5.4 9.4	2.08pm/kPa when <2.5kPa 0.98pm/kPa when >2.5kPa
SYLGARD	Within a few seconds	No loading or unloading spikes	0.11	2.98pm/kPa when <12kPa
Dragon Skin	In less than a minute	Small loading or unloading spikes	0.89	2.26pm/kPa

When comparing the results from the sample tests to the specifications required for a tracheal contact pressure sensor (as described in Chapter 2.1.6), Sample 1 (SE1700), SYLGARD, and Dragon Skin exhibited desirable characteristics. The wavelength stabilised within a few seconds for these samples, while other samples took a few minutes. SYLGARD demonstrated the minimum hysteresis, and Dragon Skin showed a linear response. Additionally, these physical samples were not sticky like the UV-cured samples and returned to their original form after unloading. As a result, Sample 1 (SE1700), SYLGARD, and Dragon Skin were selected for further testing.

3.3.3.2 Batch and Repeatability Tests

Three pairs of SYLGARD rectangles, each measuring 7mm x 5mm x 1mm, were cut from a larger sample prepared in a petri dish to form three sensing samples. Each rectangle was manually measured with a ruler, which may introduce slight variations in size on the order of micrometres. This may influence results. Figure 3.19 shows the sensitivity results of all three sensing samples obtained using the procedure described in Section 3.2.2.2. It is worth noting that the FBG used in the first SYLGARD test (in Section 3.3.3.1, Figure 3.17) broke (this FBG was from Femto Fiber Tec, fibre was SMF28 Ultra, core 8.2 μ m, cladding 125 μ m, apodized, acrylate coating and recoated after FBG inscription), and an in-house FBG was created and used for these batch tests. In addition to this, as SYLGARD 184 is a two-part polymer, slight variations in the ratio can result in slightly different properties. All SYLGARD sensors were created with the recommended 10:1 ratio, however there may be variations of a few grams. Therefore, the sensitivities from these batch tests may not match the sensitivity measured in Section 3.3.3.1.

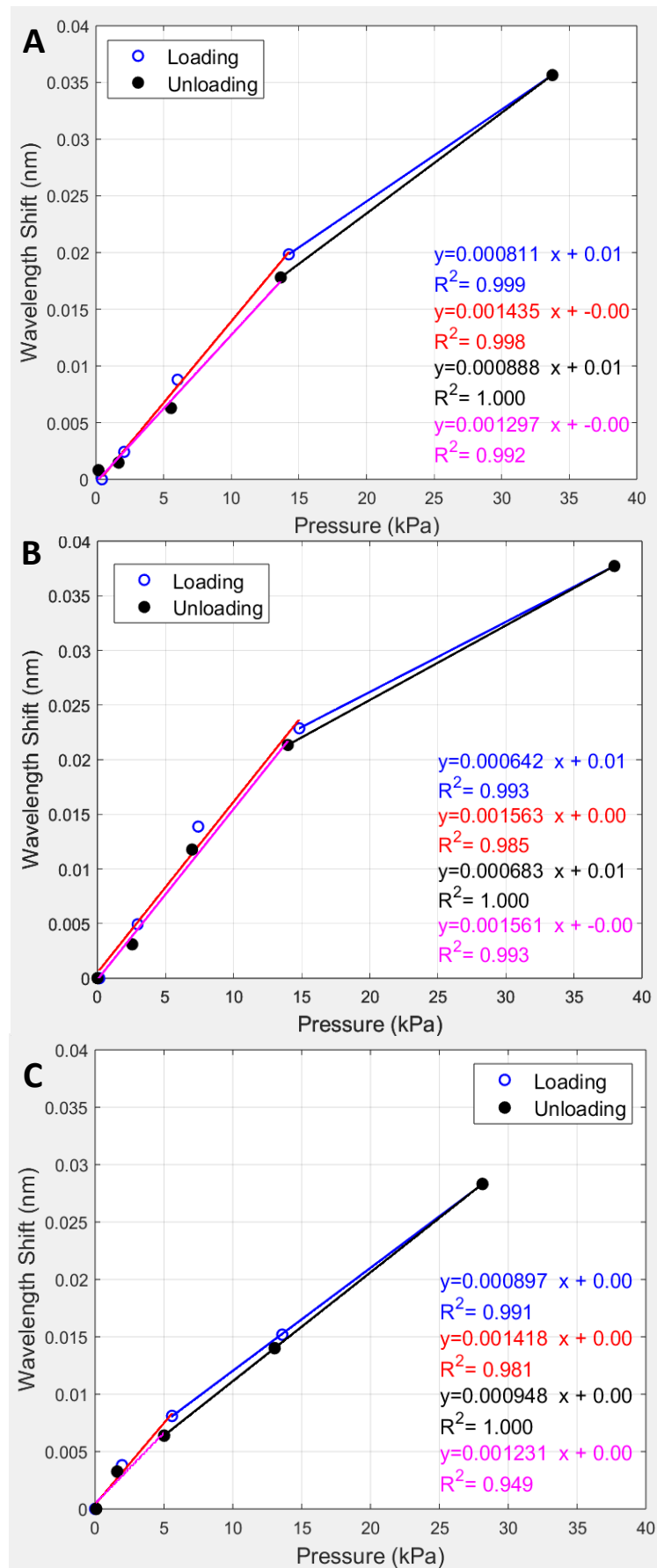


Figure 3.19: Sensitivity results of batch testing SYLGARD samples A: first pair of rectangles, B: second pair and C: a third pair of SYLGARD rectangles

Figures 3.19a, 3.19b, and 3.19c show that the loading and unloading curves vary slightly. Small hysteresis is present in these experiments, unlike that carried out in Section 3.3.3.1 (Figure 3.17b). The sensitivities calculated from Figure 3.19 are shown in Table 3.5.

Table 3.5: Sensitivity of the three sandwich sensors created from the batch SYLGARD sample.

Experiment	Sensitivity (pm/kPa)	Hysteresis Error (pm)
Figure 3.19a	1.37 when <15kPa	1.95
	0.85 when >15kPa	2.6
Figure 3.19b	1.56 when <15kPa	0.028
	0.66 when >15kPa	1.5
Figure 3.19c	1.32 when <5kPa	1
	0.92 when >5kPa	1.4

These sensitivities slightly differ and is worth noting that this could be due to the FBG being sandwiched and not embedded in these tests. A CV of 8.9% for pressures below 5kPa (indicating moderate repeatability) and a CV of 16.6% for pressures above 5kPa (indicating poor repeatability) was calculated. It is possible for these CV values to decrease when fibres are fully embedded.

Another experiment was carried out with the SYLGARD sample to tests sensor repeatability under cyclic loading. This was to assure that the wavelength returned to its original baseline after each loading and unloading cycle. The cycle was repeated 3 times. Figure 3.20 shows the loading and unloading cycle against time.

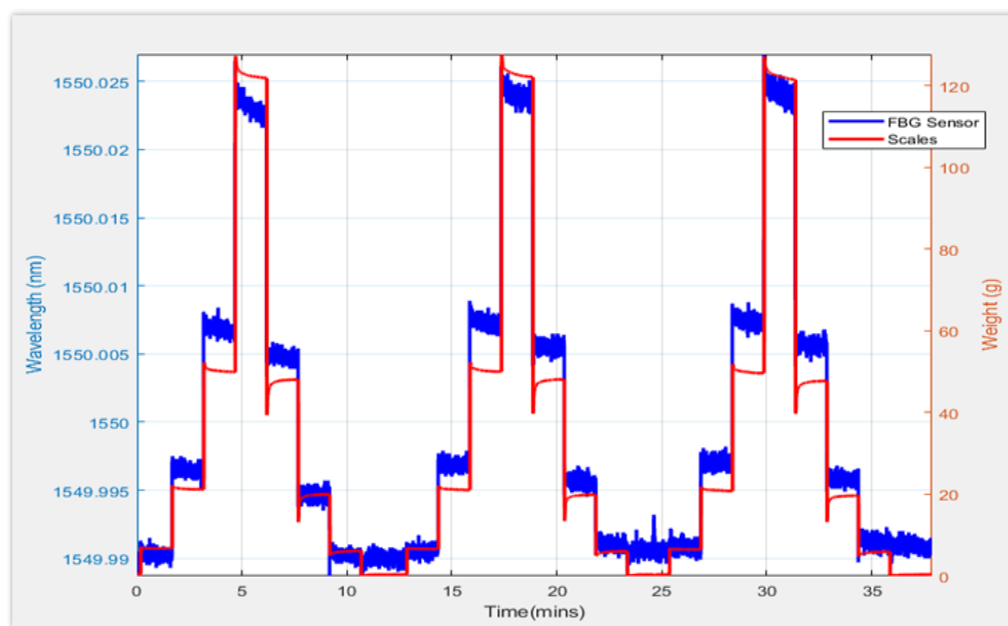


Figure 3.20: Cyclic results of testing SYLGARD sample.

It's clear from Figure 3.20 that the loading and unloading cycles of the 3 cycles are very similar. The wavelength always reaches baseline and doesn't take long to do so. Figure 3.21 shows the sensitivity calculated from each cycle to see if there is a change.

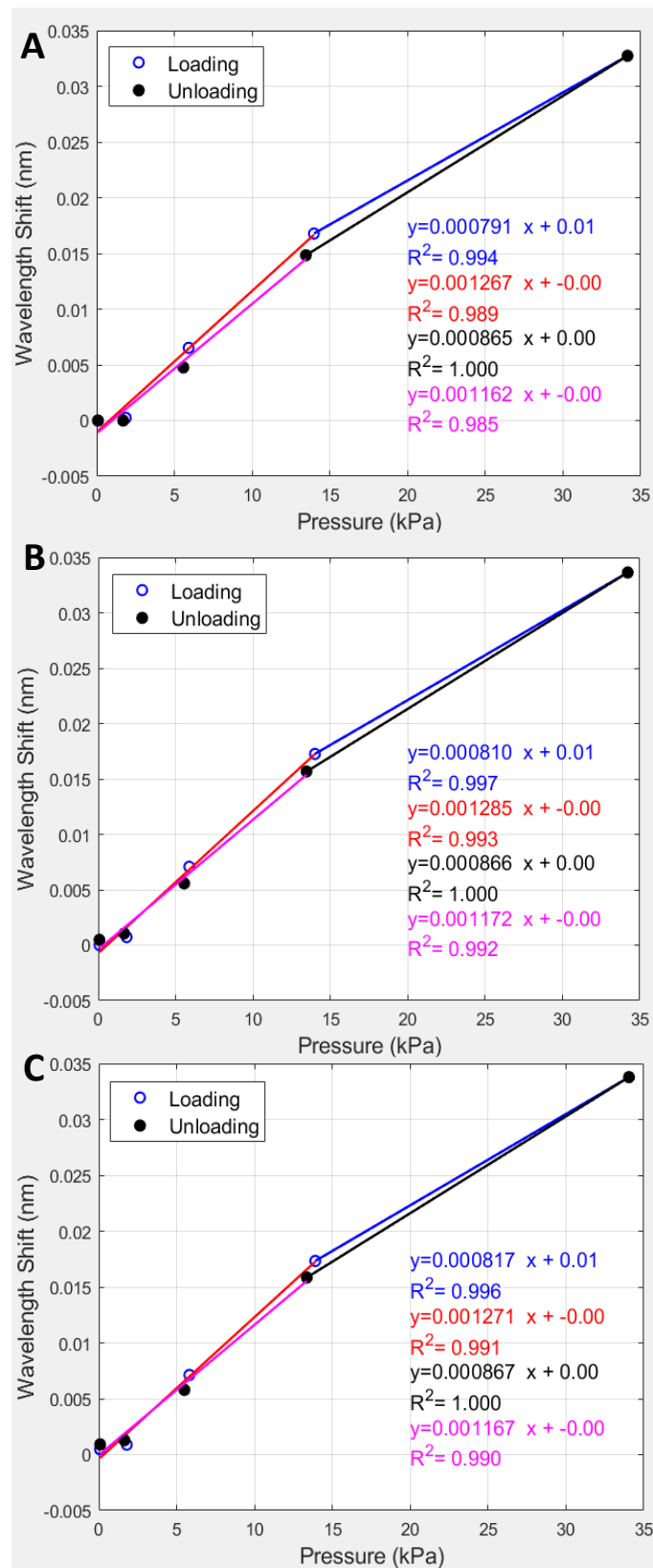


Figure 3.21: Pressure sensitivity of each cycle tested in Figure 3.20.

Like Figure 3.20, Figure 3.21 shows the sensitivity curves are almost identical. The sensitivities calculated from Figure 3.21 are shown in Table 3.6.

Table 3.6: Sensitivity of each cycle tested in the cyclic experiment for the SYLGARD sample.

Cycle	Sensitivity (pm/kPa)	Hysteresis Error (pm)
Figure 3.21a	1.21 when <15kPa 0.83 when >15kPa	1.4 9
Figure 3.21b	1.22 when <15kPa 0.84 when >15kPa	1.5 9.2
Figure 3.21c	1.22 when <15kPa 0.84 when >15kPa	1.4 9.3

With the sensitivities of each cycle being so similar, the CV for this sensor is 0.47% for the region below 15kPa, and 0.69% for the region above this. This indicates strong repeatability.

3.3.3.3 ABAQUS Model Simulations

After running the simulation with the pressures used experimentally, the model showed some deformation. Figure 3.22 shows the deformation of the patch under the plate load. Figure 3.22a shows the patch before a load is applied. Figure 3.22b shows the deformation of the patch after the load is applied.

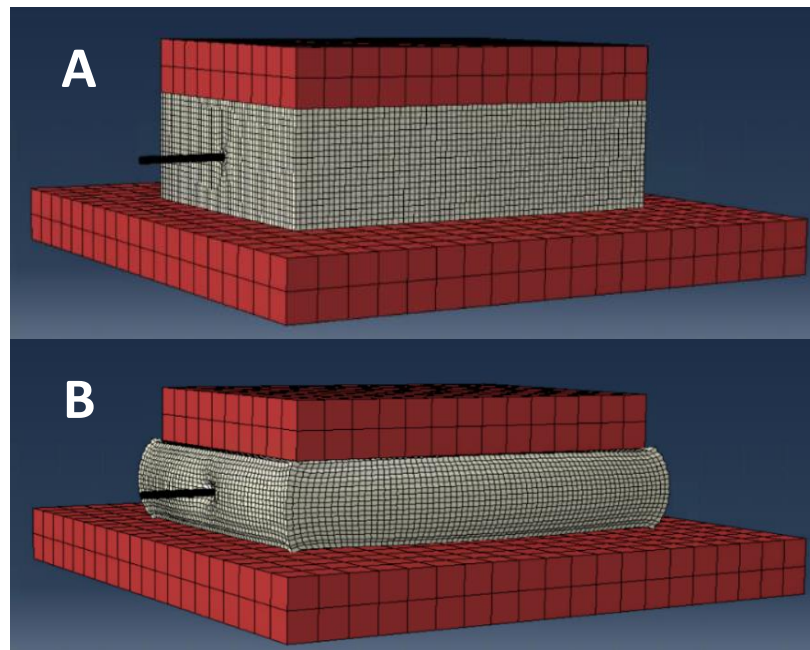


Figure 3.22: A: the model before deformation, red structures- surface (10mm x 10mm x 1mm) and plate (7mm x 5mm x 1mm), white structure- patch (7mmx5mmx2mm), dark green structure- fibre (125µm diameter x 10mm in length). B: the deformed model after running the ABAQUS simulation.

From Figure 3.22b, it is visible that the load applied caused the patch to expand outward and bulge. It can be seen that in the location where the fibre exits the patch, the expansion is limited. This is probably due to friction between the fibre and the patch restricting the expansion in this specific region.

3.3.3.4 Experimental Results vs Simulated Results

During the batch tests, three experiments were carried out on a large SYLGARD sample created in a petri dish. This sample was cut into rectangles of the same dimensions to make three sensing samples, results of these experiments are shown in Figure 3.19. In Figure 3.23, these experimental results are compared with simulated data that utilised the same loads as those used experimentally. Figure 3.23 shows the results for all three batch experiments (experimental, and simulated results).

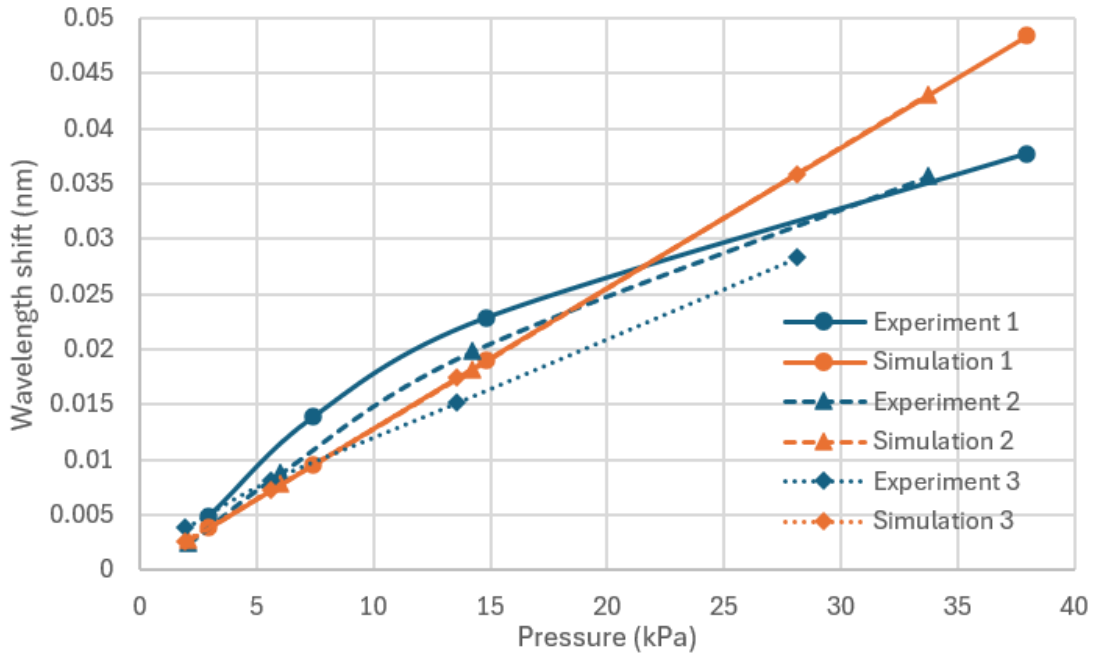


Figure 3.23: Experimental results and ABAQUS simulated results.

Figure 3.23 shows that the simulation has a linear response whereas experimentally the response is non-linear, this could be due to the limited number of material properties the model utilised (the material properties can be seen in Table 3.2). Alternatively, the non-linear results could be due to the sandwiching of the fibre into the patches. Additionally, for pressures above ~ 21 kPa, the simulated wavelength shift is larger than the experimental wavelength shift. This could be due to the model having full fibre to patch adhesion whereas in the experiments the fibre is sandwiched between the patch samples.

In the experiments, the average sensitivity for the SYLGARD sensors was 1.42pm/kPa for pressures <15kPa. For pressures above 15kPa, the sensitivity was 0.81pm/kPa. The FEA model results show a sensitivity of 1.27pm/kPa. This discrepancy could be due to a number of factors including the non-linearity of the experimental results, the few material properties incorporated in the model (as shown in Table 3.2), the weighing scales readings as they do not stabilise after each pressure increment, and favouring computational timing over result accuracy when meshing the model. Despite these differences, the Pearson's correlation coefficient between the simulated and experimental results indicates that there is a significant positive correlation ($r(12)=0.978$, $p < 0.001$). This means that if a factor causes the simulated results to change in some way, this factor will also cause the experimental results to change in the same way. This confirms that the model can be used to predict experimental changes.

3.3.4 Conclusion

Initial sandwich experiments were conducted with seven sample materials to evaluate their response to loading and unloading. For these experiments, two rectangles were cut from each sample, with an FBG placed between them. The scale setup described in Section 3.2.2.2 was utilised. Among the seven samples, three were selected for further testing to create a complete sensor by embedding a fibre into a patch made from these materials. The three chosen samples were Sample 1 (SE1700), SYLGARD, and Dragon Skin. They were selected due to their rapid stabilisation of the FBG during loading and unloading, non-sticky physical properties, and ability to return to their original form after unloading. These characteristics made them suitable for use as contact pressure sensors.

Batch tests were performed using the SYLGARD sample, where a large sample of SYLGARD was cured in a petri dish and multiple rectangles were cut and tested using the loading arm and scale setup. This aimed to assess whether the sensors could be batch-produced with consistent sensitivity across each sensing patch. Results indicated slight variations in sensitivity among the pairs of samples tested, with a CV of 8.9% for pressures below 5kPa (indicating moderate repeatability) and a CV of 16.6% for pressures above 5kPa (indicating poor repeatability). This is likely due to the nature of the sandwich test method and the FBG not being fully embedded in the fibre.

A repeatability test was also conducted with the SYLGARD sample to evaluate its response to cyclic loading. This was to ensure that the wavelength baseline returned to its original state after each loading and unloading cycle. The results were promising, showing that each cycle was almost identical, with a CV of 0.47% for pressures below 15kPa, and 0.69% for pressures above this, indicating strong repeatability. These findings suggest that SYLGARD has the potential to be a reliable and repeatable sensing material.

An FEA model was created in ABAQUS of the SYLGARD sample to determine the results from the batch experiments. The model, if working correctly could then be used to determine how to optimise the sensor further. The experimental pressure values from the batch experiments were used in the simulations. A comparison of simulated and experimental results was carried out. Results showed that the Pearson's correlation coefficient between the two results was found to be 0.978 with a p-value of <0.001. This shows high correlation between the simulated and experimental results and confirms that the model can be used to predict experimental changes with high certainty.

3.4 Conclusion

The development and testing of two contact pressure sensors using epoxy as an embedding material revealed several key findings. Results showed inconsistent fabrication and insufficient sensitivity to tracheal contact pressures. Consequently, it was evident that optimisation was necessary, including the need for a more sensitive material and a reliable fabrication method.

Subsequent tests on seven sample materials identified SE1700 (Sample 1), SYLGARD, and Dragon Skin as promising candidates due to their rapid stabilisation, non-sticky properties, and ability to revert to their original form post-loading. Batch tests on SYLGARD indicated moderate repeatability for pressures below 5kPa and poor repeatability for pressures above this. The repeatability cyclic loading tests however, showed strong repeatability. FEA modelling of SYLGARD confirmed high correlation between simulated and experimental results, affirming the model's ability to predict experimental changes with high certainty. These findings underscore the need for further optimisation and a new calibration method, ideally mimicking tracheal conditions, to enhance sensor performance and reliability for tracheal contact pressure applications.

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Chapter 4 – Embedded Sensor Testing and ABAQUS Embedded Sensor Model

4.1 Introduction

The first part of this chapter discusses creating fully embedded sensors from the materials chosen for further testing (Chapter 3.3.3.1). These embedded sensors are fabricated using a new mould, created by BlueFrog Design, that addresses the shortcomings of the open PFTE moulds (these shortcomings are described in Chapter 3.2.3). The second part of this chapter is about testing these embedded sensors on the loading arm and scale setup to evaluate their sensitivity and repeatability. The FEA model created in Chapter 3.3.2.5 was altered to match the new mould dimensions of the embedded sensors (the BlueFrog Design mould). The final part of this chapter investigates optimising the sensors for increased sensitivity and repeatability. The altered FEA model was also utilised in this chapter to determine optimisation approaches.

4.2 Fabrication Process

4.2.1 Introduction

As the fabrication process using the PTFE mould (described in Chapter 3.2.2.1) was unable to create repeatable sensors every time and resulted in sensors performing differently (due to difficulty ensuring the length of the fibre was running uniformly through the cured sample and due to the polishing of the patch before removing from the mould resulting in an uneven sensing surface), a new fabrication process needed to be employed. The ideal fabrication process must satisfy the following:

1. Result in a smooth uniform sensing surface once cured.
2. Have the fibre uniformly running through the cured sample.
3. Allow the sample to cure without bubbles.
4. Produce repeatable sensors every time.

4.2.2 3D Printing Sensors

Since Sample 1 (SE1700, Dow Corning) had the potential to be 3D printed, this new fabrication method could resolve the issues of curing in an open mould. It has potential to make sure all surfaces of the patch are smooth and parallel. To test the feasibility of 3D printing the sensors, Sample 1 (SE1700) was deposited using a custom-built 3D printer in the Additive Manufacturing Facility at The University of Nottingham. Figure 4.1 shows the custom 3D printer.



Figure 4.1: Custom-built 3D printer at Additive Manufacturing Facility at the University of Nottingham.

The printer was constructed with a movable stage, featuring two axes: one moving the printhead in the x-direction (PRO115LM, Aerotech) and the other in the y-direction (PRO115SLE, Aerotech), both controlled by A3200 controllers from Aerotech. The printhead (Pico Pulse, Nordson) deposited two layers of Sample 1 (SE1700) with a thickness of 0.25 mm per layer onto a piece of glass. The fibre was then placed in the centre, and an additional two layers were printed on top of the fibre creating a 1mm thick sensor. Since the printhead lacked heaters, the SE1700 was deposited at room temperature.

It became quickly apparent that this method of fabrication would not work for this sensing material. Figure 4.2 shows the resultant sensor.

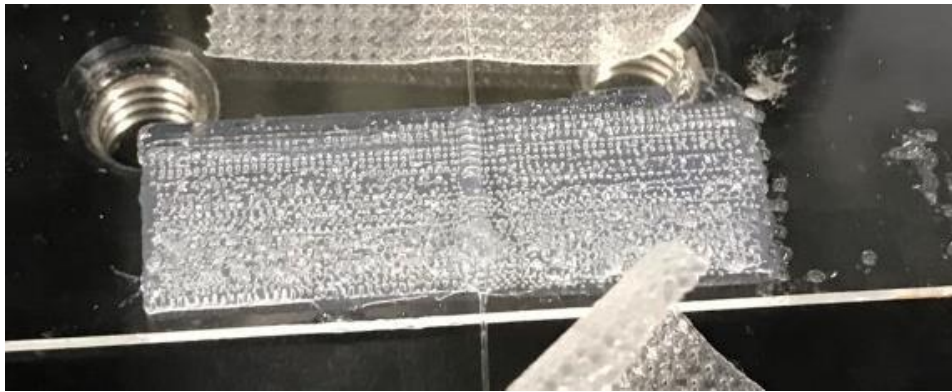


Figure 4.2: Bubbles produces when 3D printing Sample 1 (SE1700).

As shown in Figure 4.2, there are lots of bubbles in the sample. These bubbles persisted even when placed in a vacuum chamber for an hour. The sample viscosity was too high for the bubbles to release (as it had a gel-like consistency). In addition to this, there is a clear bulge on the surface where the fibre had been placed as the added deposit sat on top of the fibre. Due to these results, the 3D printing of sensors and the use of Sample 1 (SE1700) were put on hold (as removing the bubbles from Sample 1 (SE1700) proved difficult) while other fabrication methods and sensor materials were explored.

4.2.3 Embedded Sensor Fabrication Method

BlueFrog Design manufactured a bespoke injection mould where the inlet is located on the side of the sensor rather than the sensing surface (Figure 4.3a). This design ensures a smooth sensing surface with parallel faces and uniform thickness for each sensor. Figure 4.3b shows a schematic diagram created by BlueFrog of the casted sensor. It is worth noting that the change in sensor dimensions was necessitated by another project within the Optics and Photonics Research Group at the University of Nottingham. Since

these were preliminary tests to assess the repeatability of the sensors produced by this mould, the altered dimensions were used. If the results prove favourable, a mould with the desired dimensions can be created as part of the future work for this project.

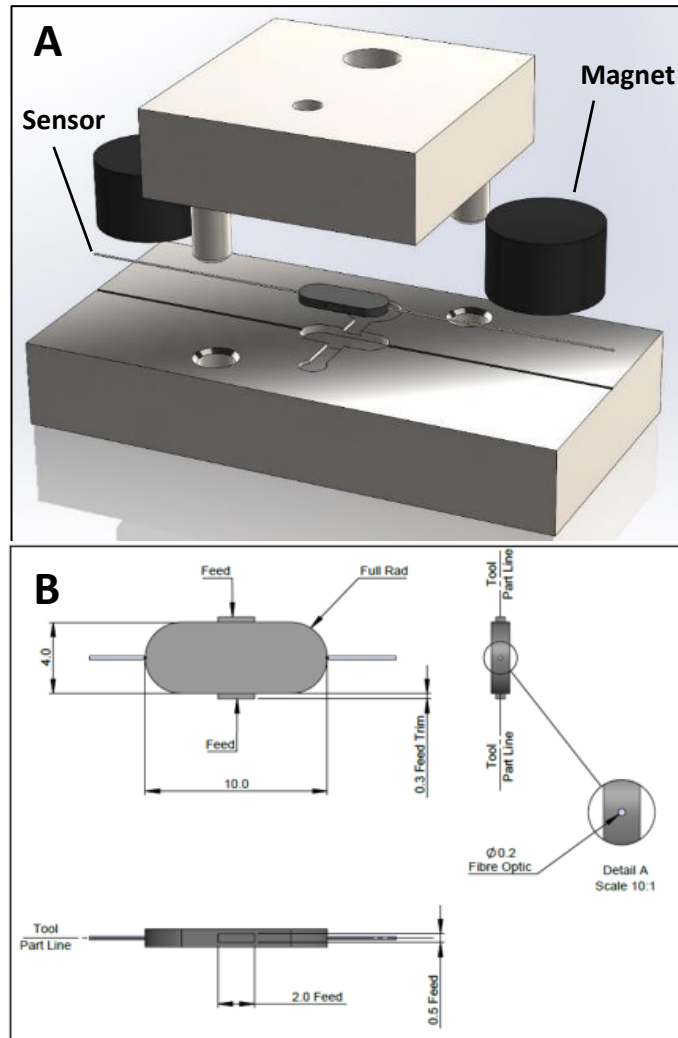


Figure 4.3: A: Mould design from BlueFrog Design. B: Dimensions of cast sensor using BlueFrog Design mould.

As shown in Figure 4.3a, the mould is created in such a way that the fibre is placed in the centre of the mould and is fixed in place with magnets preventing fibre movement. This addresses both concerns with the old PTFE open moulds that were used to create the epoxy sensors in Chapter 3.2.2.1. From Figure 4.3b, this mould creates sensors with a thickness of 1mm, a width of 4mm, and a length of 10mm.

Figure 4.4 shows the mould in use.

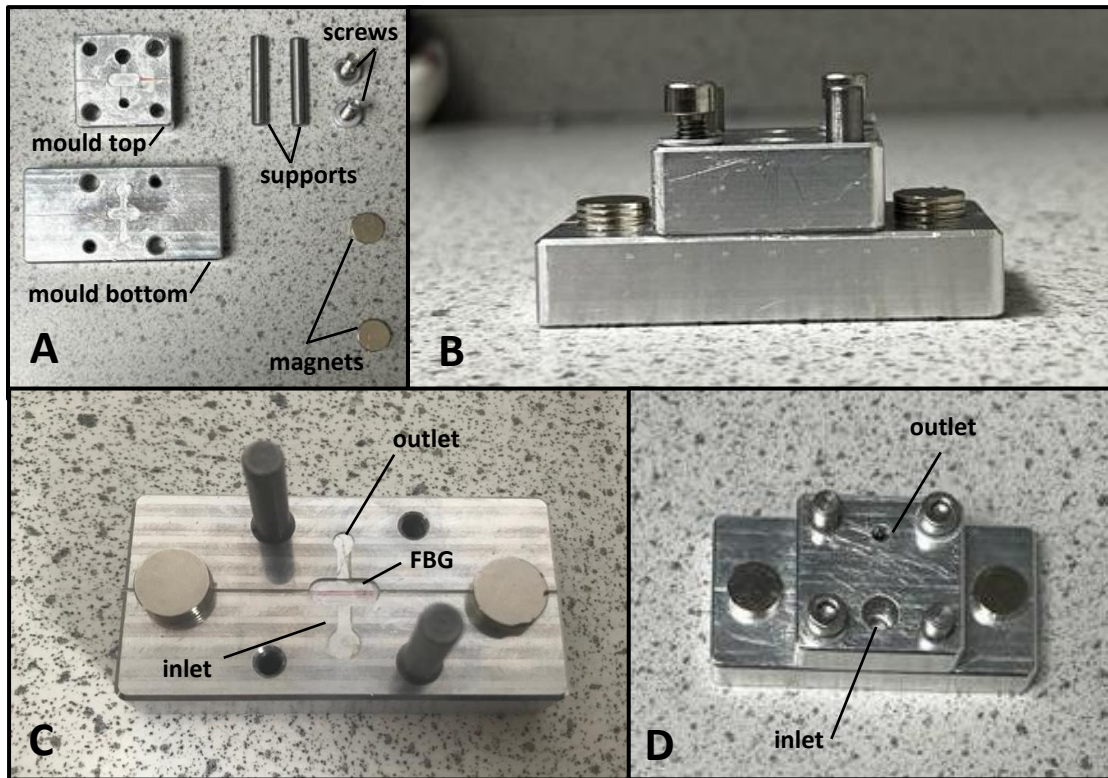


Figure 4.4: A: mould parts (mould bottom is 6cm x 3cm x 1.25cm, mould top is 3cm x 3cm x 1.25cm), B: mould assembled, C: fibre placed in mould for patch casting, D: assembled mould and casting solution placed in inlet.

This mould was used to create sensors out of SYLGARD and Dragon Skin. The following procedure was carried out:

1. The mould was taken apart and the inner parts of the mould (that form the sensor) were sprayed with mould release spray (Mould release spray ROC72021, Rocol), the fibre was cleaned with IPA Alcohol and optical fibre dustless cleaning tissue, and was placed in position so that the 3mm FBG used for contact pressure sensing was placed in the centre of the patch mould. As FBGs are not visible to the human eye, 3mm of the fibre was stripped upon fabrication of each of the 3mm FBGs and not recoated. The stripped regions of the fibre serve as markers for the location of the FBGs, helping ensure they were placed in the centre of the mould. Once they were placed in the centre of the mould, the fibre was kept in place with strong magnets. The mould was then reassembled.
2. The patch material solution was made according to their directions-
 - For SYLGARD: the two-part solution was mixed at a 10:1 ratio.
 - For Dragon Skin: the two-part solution was mixed at a 1:1 ratio.

3. The mixed solution was placed into a vacuum de-gassing chamber (26L DS-26S, Easy Composites) for approximately 15 minutes until no bubbles remained.
4. Using a pipette, the solution was then gravity fed into the mould through the large inlet at the top of the mould (Figure 4.4d).

By pouring the solution into this inlet, it is directed through the channel on the side of the sensor (as shown in Figure 4.4c). This process ensures that the sensor patch has smoothly cast surfaces, free from bubbles. Additionally, the use of an injection mould guarantees that the sensor will replicate the exact dimensions and features of the mould once cured. Since the mould's surfaces are parallel, the resulting sensors also have parallel surfaces.

Once these steps were completed the curing procedure varied for each solution.

For the SYLGARD sensor, after the solution was poured into the mould, the mould was placed in the oven (Model FD 23, Binder) for 1 hour at a temperature of 80°C. The mould was then removed and left overnight at room temperature.

For the Dragon Skin sensor, the mould was left overnight at room temperature.

4.2.4 Embedded Sensor Fabrication Results and Discussion

After leaving to cure overnight at room temperature, the mould was taken apart and the sensor was carefully removed. Figure 4.5 shows the sensors created from SYLGARD and Dragon Skin.



Figure 4.5: Cured sensors from SYLGARD (top) and Dragon Skin (bottom). Sensors are 10mm in length and 4mm in width.

As shown in Figure 4.5, the two sensors look almost identical, they have a smooth sensing surface (devoid of bubbles and both surfaces are parallel) with similar shape and size. The dimensions were checked with a micrometre. Measurements all matched those of the mould confirming that the mould creates reproducible sensors.

4.2.5 Conclusion

The new mould from BlueFrog Design shows that suitable sensors can be made without further refinement, meaning they can be used straight out of the mould. It is easy to create these sensors and with the use of a vacuum chamber, sensors can be made without the presence of bubbles. Two sensors were created using this new mould, one from SYLGARD and one from Dragon Skin. Both the created sensors had the same dimensions when measured with a micrometre and did not require any polishing to achieve uniform surfaces.

4.3 Embedded Sensor Testing

4.3.1 Introduction

The SYLGARD and Dragon Skin sensors, developed using the BlueFrog mold, exhibit promising potential but require further testing to evaluate their performance accurately. It is essential to establish a more sensor specific target sensitivity. With the sensor dimensions yielding an area of $3.65 \times 10^{-5} \text{ m}^2$, a 1g change in load translates to a pressure of 0.26kPa, which falls short of the required resolution specified in Chapter 2.1.6. Therefore, the sensor needs to detect a 0.1g change in load, equivalent to 0.026kPa, and this meets the target specification. Considering the optical interrogator's minimum resolution of 0.1pm and the minimum pressure to capture at 0.026kPa, the ideal sensitivity calculates to 0.1 pm per 0.026 kPa, which is 4 pm/kPa. This sensitivity ensures precise measurement of the required pressures within the defined range. Higher sensitivities exceeding 4 pm/kPa may inadvertently amplify unwanted noise or artifacts, potentially compromising measurement accuracy.

4.3.2 Methods

4.3.2.1 Experimental Testing

Three experiments were carried out for each sensor. The first was carried out the day after the sensor was fabricated (once the sensor had cured), the second was carried out a week later, and the third the week after. This was to see how the sensor performed over time.

The same loading arm and scale experimental setup described in Chapter 3.2.2.2 and shown in Figure 3.4 was used for testing the two embedded sensors. The loading arm was moved onto the sensors until a load of approximately 50g was applied (resulting in approximately 15kPa). Although the typical pressures used clinically to create a gas seal during intubation is 2-3kPa, over-inflation can occur (due to reasons covered in Chapter 2.1.4), and therefore it is important to make sure the sensors retain their sensitivity to pressures above this 2-3kPa range.

As the surface area of the sensor has changed from the 7mm x 5mm sandwich tests in Chapter 3 (Chapter 3.3.2.2) to the dimensions shown in Figure 4.3b, the area of the new

sensor is now the area of a 6mm x 4mm rectangle, plus the area of a 2mm radius circle. This results in a total area of $3.65 \times 10^{-5} \text{m}^2$.

4.3.2.2 ABAQUS Model

Since the shape of the sensor changed, another FEA ABAQUS model was created, this new model would be used to compare experimental results to the simulated ones and be utilised to further optimise the sensors.

a) Model design and material properties

As the experimental methods are similar to those previously carried out, the model created is similar to the previous model described in Chapter 3.3.2.5. The main difference is the dimensions of the SYLGARD part of the model. The fibre was also made longer, and the surface was made bigger. The dimensions used for the embedded sensor model are displayed in Figure 4.3. Table 4.1 shows the list of parameters used in the model and their values. The coefficient of friction throughout the model was set to 0.64, which is the coefficient of friction between a steel and rubber surface (*Coefficient of Friction*, Engineers Edge).

Table 4.1: Material properties and part dimensions of ABAQUS embedded sensor model

Model Part	Young's Modulus (MPa)	Poisson's Ratio	Dimensions
Fibre Core	72000 (Feng et al., 2010)	0.17 (Wierzbna & Kosmowski, 2003)	9µm (diameter) x 12mm
Fibre Cladding	72000 (Feng et al., 2010)	0.17 (Wierzbna & Kosmowski, 2003)	125µm (diameter) x 12mm
SYLGARD Patch	2 (Dow Corning)	0.495 (Müller et al., 2019)	4mm x 6mm rectangle with two 2mm radius semicircles on either side x 1mm thick
Steel Surface	200000 (<i>Young's Modulus of Steel</i> , AmesWeb)	0.3 (<i>Poisson's ratio</i> , Engineering Toolbox)	14mm x 14mm x 1mm
Steel Plate	200000 (<i>Young's Modulus of Steel</i> , AmesWeb)	0.3 (<i>Poisson's ratio</i> , Engineering Toolbox)	4mm x 6mm rectangle with two 2mm radius semicircles on either side x 1mm thick

Again, as pressure in ABAQUS relies on the area of the surface the load is applied to, the plate was also changed to match the size of the patch.

The boundary conditions and constraints were the same as the ones used in the 7mmx5mmx2mm ABAQUS model described in Chapter 3.3.2.5. The uniform load applied on the plate of the model was a pressure, the pressures used in this model were the experimental pressures calculated from the experiments described in Chapter 4.3.2.1. Figure 4.6 shows the created embedded sensor model with the boundary conditions and load.

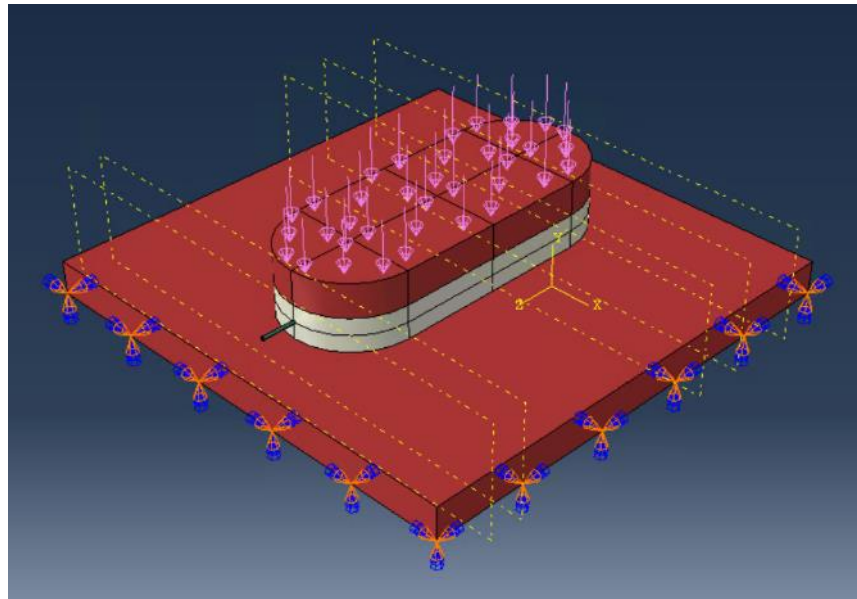


Figure 4.6: ABAQUS model showing load and boundary conditions.

The red parts correspond to the material properties of Steel (surface: 14mm x 14mm x 1mm, plate: 10mm x 4mm x 1mm). The white part corresponds to the material properties of SYLGARD 184 (patch: 10mm x 4mm), and the dark green parts correspond to the material properties of the fibre (125 μ m x 12mm).

b) Refining model mesh

When refining the mesh of the model, Table 3.3 was taken into consideration. Therefore, the meshing of this model has similar meshing to the previous model where the surface and plate were both meshed at 0.5mm, the SYLGARD patch was meshed at 0.08mm and the fibre was meshed at 0.01mm. Figure 4.7 shows the meshing of the embedded sensor model.

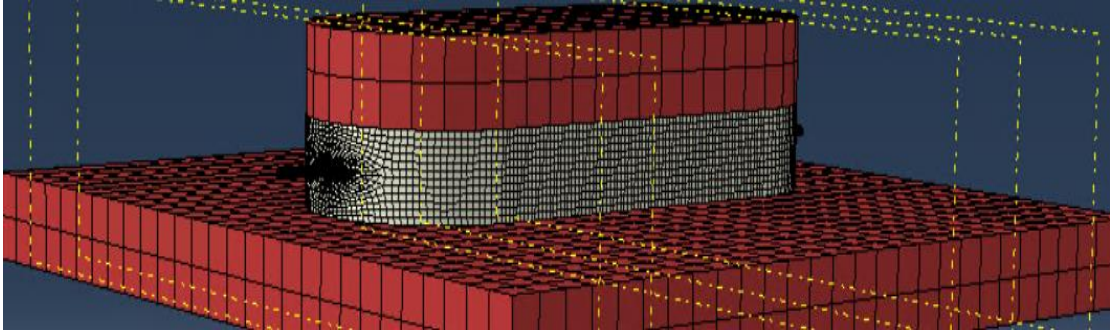


Figure 4.7: Meshed embedded sensor model; each red grid box for the mesh of the surface and plate is 0.5mm by 0.5mm. Each white grid box for the mesh of the patch is 0.08mm by 0.08mm. Each dark green grid box for the mesh of the fibre is 0.01mm by 0.01mm.

c) Stress and strain outputs

For the analysis of the model results the stress and strain values of each node along the core were exported. This method is described in Chapter 3.3.2.5c and was employed again for this model.

4.3.2.3 Statistical Analysis

To assess the accuracy of the sensor, linear regression and test-retest repeatability was evaluated using CV values (by using equation 3.4). All final sensor sensitivities are presented as mean $\pm$ SD.

Pearson's correlation coefficient (PCC) was calculated to evaluate the relationship between experimental and simulated results using Microsoft Excel (Excel 2021 v16.0, Microsoft). Statistical analysis with unpaired, two-tailed t-tests were also conducted, and differences were considered significant when the p-value was ≤ 0.05 .

4.3.3 Results and Discussion

4.3.3.1 Experimental Testing

Experiments were carried out for both the SYLGARD and Dragon Skin embedded sensors and repeated twice. Figure 4.8 shows the results for the SYLGARD sensor.

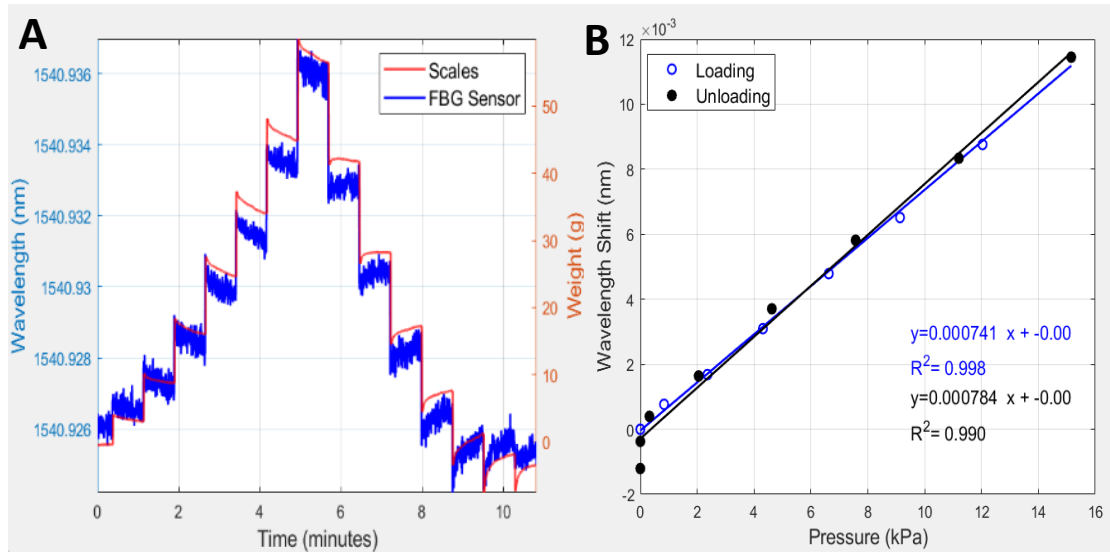


Figure 4.8: Experimental results for 1mm SYLGARD. A: wavelength vs time (from interrogator) and weight vs time (from scale), B: calibration curve.

Figure 4.8a shows that the number of loading steps and unloading steps of the scale are not equal. The unloading cycle has two fewer steps than the loading cycle and this is shown in both the scale and FBG response. This is most likely due to the nature of the loading arm setup, although the loading arm is programmed to incrementally apply and release the load, it may encounter resistance during loading as it presses onto the weighing scales. During unloading there is no resistance to the loading arm movement. This results in fewer steps in the unloading cycle to reach back to the 0kPa baseline. In addition to this, from Figure 4.8a, the sensor baseline at the end of the experiment falls slightly from the starting baseline (by 1pm). This is also observed in the repeated experiments. This could be due to hysteresis. Figure 4.8b shows a linear response. The results of the repeated tests also show this linear response. The CV for this sensor is 6.22%, which shows moderate repeatability. It is worth noting that the sample size is very small and therefore for a more accurate CV, further testing would need to be carried out. When taking the average of the sensitivities of the three experiments, the sensitivity of the SYLGARD sensor is $0.82 \pm 0.051\text{pm/kPa}$ with a hysteresis error of $0.26 \pm 0.34\text{pm}$.

Figure 4.9 shows the results for the Dragon Skin sensor.

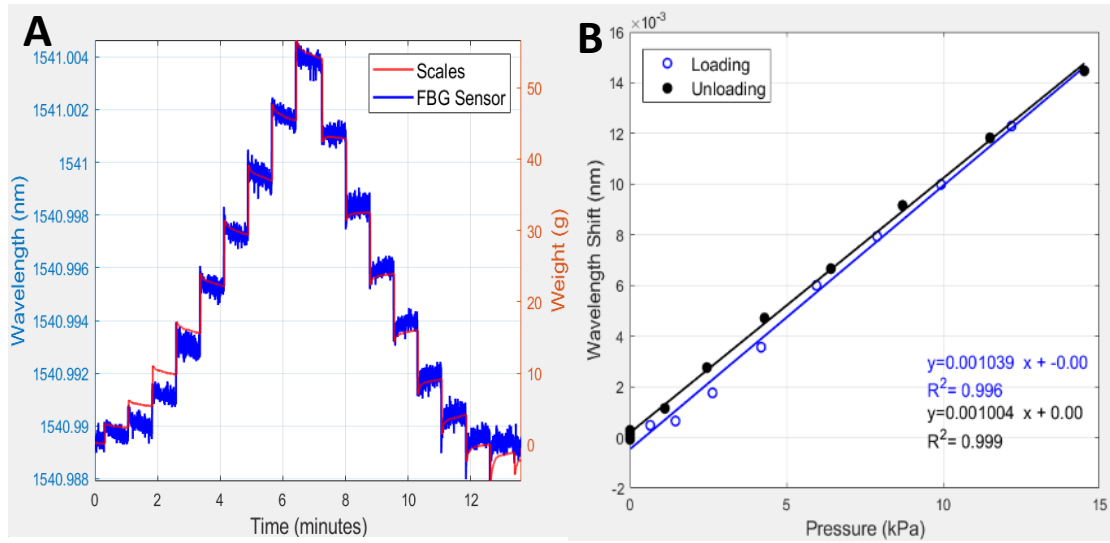


Figure 4.9: Experimental results for 1mm Dragon Skin. A: wavelength vs time (from interrogator) and weight vs time (from scale), B: calibration curve.

Figure 4.9a shows the same phenomena as in Figure 4.8a, where the number of steps in the unloading cycle is fewer than the loading cycles. However, for this sensor the baseline remains constant at 1540.989nm, and this is also observed in the repeated experiments. Figure 4.9b shows a linear response and the repeated tests also show this linear response. The CV for this sensor is 5.54% which also indicates moderate repeatability. When taking the average of the sensitivities from the three experiments, the sensitivity of the Dragon Skin sensor is $0.99 \pm 0.055\text{pm/kPa}$ with a hysteresis error of $0.4 \pm 0.1\text{pm}$.

When comparing the sensitivities of both the SYLGARD and Dragon Skin sensors, Dragon Skin shows a slightly higher sensitivity than the SYLGARD sensor (by 0.17pm). When comparing these results to the target specification, the sensors show a desirable linear response with low hysteresis. However, the repeatability needs improving in order to show good repeatability, and the sensitivity of the sensors were significantly below the ideal sensitivity of 4pm/kPa. Therefore, optimisation is required to better align the sensor performance with the application requirements.

4.3.3.2 ABAQUS Model

After running the simulation with the pressures used in the SYLGARD experiments above (Chapter 4.3.3.1), the model showed some deformation. Figure 4.10 shows the sensor before and after the pressure was applied.

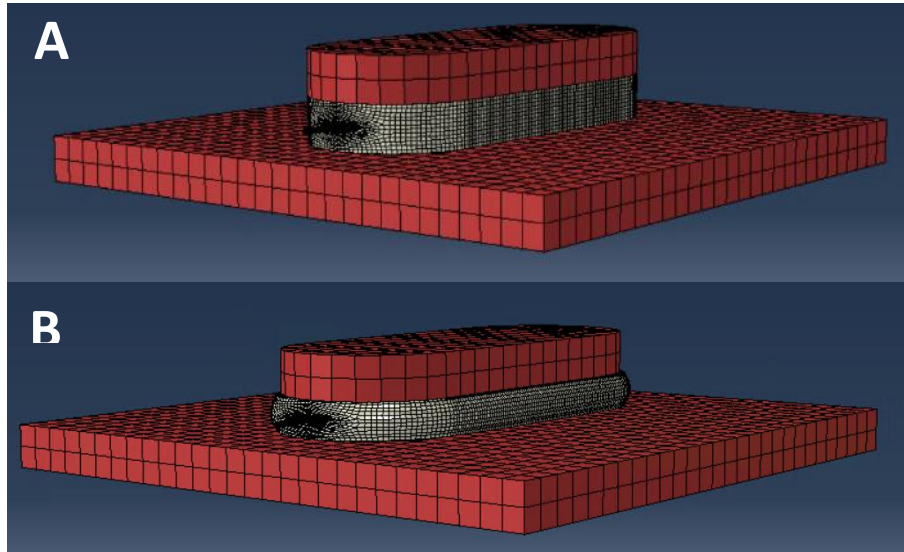


Figure 4.10: A: the embedded sensor model before deformation. B: the deformed model after running the ABAQUS simulation and the pressure is applied. Scale- each red grid box for the mesh of the surface and plate is 0.5mm by 0.5mm.

Figure 4.10b shows the patch expand causing it to bulge.

4.3.3.3 Experimental Results vs Simulated Results

The calculated pressures from the three SYLGARD experiments were used in the model. The simulated wavelength shift results and experimental wavelength shift results were then compared. Figure 4.11 shows the experimental results and the simulated results for all three SYLGARD experiments.

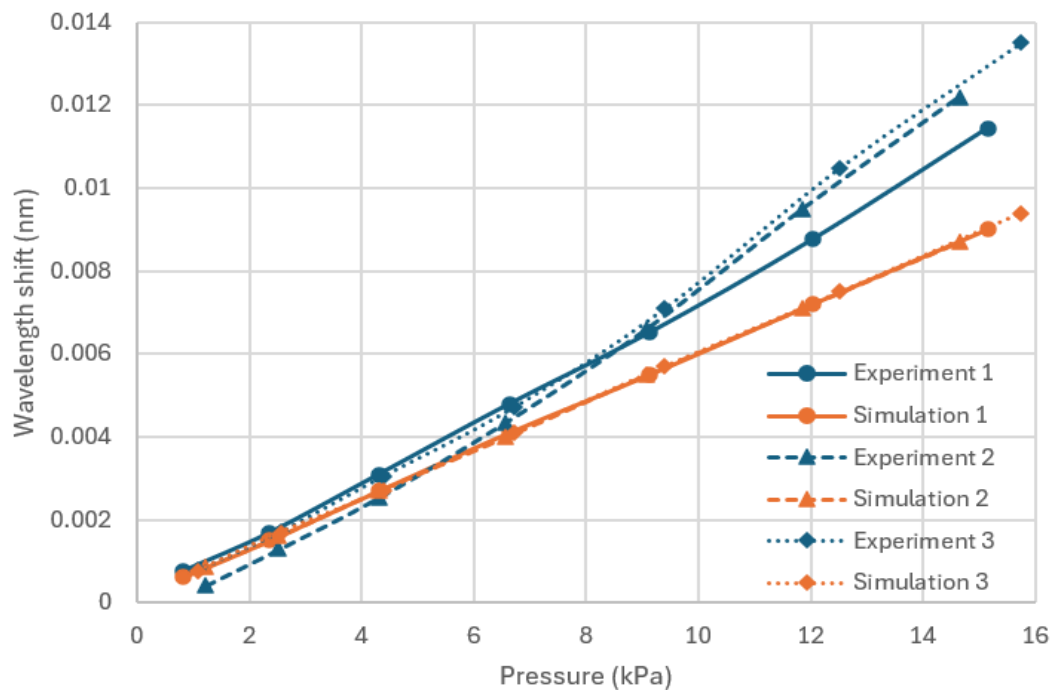


Figure 4.11: 1mm SYLGARD Experimental results and ABAQUS simulated results.

Figure 4.11 shows that the wavelength shift observed experimentally is larger than the simulated wavelength shift. This is different to what was observed during the sandwich tests, where the wavelength shift of the simulations were larger than those experimentally observed (Chapter 3.3.3.4). In the sandwich tests the fibre was not bound to the patch and therefore this observation was expected. However, for these experiments the fibre is bound to the SYLGARD patch during sensor fabrication. Therefore, the simulated results should be closer to the experimental results. Due to this assumption, it is expected that the experimental results will be more highly correlated with the simulated results than with the sandwich tests. In the experiments the sensitivity of the 1mm SYLGARD sensor was 0.82 ± 0.051 pm/kPa, the model results show a sensitivity of 0.59pm/kPa. This discrepancy could be due to a number of factors including the non-linearity of the experimental results, the few material properties incorporated in the model (as shown in Table 3.2), the weighing scales readings as they do not stabilise after each pressure increment and favouring computational timing over result accuracy when meshing the model. Despite these discrepancies, the PCC between the simulated and experimental results indicates that there is a significant positive correlation ($r(21) = 0.993$, $p < 0.001$). This shows higher correlation between the simulated and experimental results than those observed in the sandwich experiments ($r(12) = 0.978$, $p < 0.001$). This also confirms that the model can be used to predict experimental changes with high certainty.

When comparing the model sensitivity results to the target specification, it is evident that the sensitivity falls significantly below the ideal sensitivity of 4 pm/kPa, mirroring the trends observed in the experimental results. Optimisation of the sensor's sensitivity is required in order for it to better meet the target specification for a tracheal contact pressure sensor.

4.3.4 Conclusion

Three experiments were carried out with the two sensors that were created using the BlueFrog mould. These tests were done using the loading arm and scale setup. Results showed that both SYLGARD and Dragon Skin sensors had a linear loading and unloading response with slight hysteresis observed (0.26pm for SYLGARD and 0.4pm for Dragon Skin). Although overall the Dragon Skin sensor showed a higher sensitivity,

the sensitivities of each individual repeat experiment for both of the sensors, SYLGARD and Dragon Skin, varied slightly causing a CV of 6.22% and 5.54% respectively. This is thought to be due to the scale taking too long to stabilise. Another reason for this difference could be due to the sensor material curing further over time causing the sensor to have slightly different material properties to when it last performed (this is mentioned on the Product Information Sheet of SYLGARD 184 and the Technical Datasheet for Dragon Skin).

When comparing these results to the target specification, the sensors show a desirable linear response with low hysteresis. However, the repeatability needs improving as it currently shows moderate repeatability, and the sensitivity of the sensors were significantly below the ideal sensitivity of 4pm/kPa indicating optimisation of the sensors are required.

Another ABAQUS model was created with the dimensions of this SYLGARD sensor and the pressures used in the model were those used in the three repeat experiments. It was hypothesised that the PCC of the experimental results against the simulated results would be higher than the PCC of the sandwich tests. This is because the fibre is now fully bound to the patch during fabrication. Results showed that the experimental wavelengths shifts were overall larger than those simulated with a PCC of 0.993 and a p-value of <0.001. This shows higher correlation than the sandwich model and confirms that the model can be used to predict experimental changes with high certainty. This model will be used to carry out simulations to see how the sensor can be optimised.

4.4 Optimisation of Embedded Sensor Sensitivity Using ABAQUS Model

4.4.1 Introduction

Now that sensors sensitive to low pressures (below 15 kPa) have been created, the next step is to optimise sensitivity, repeatability, and minimise hysteresis to better meet the target specifications: achieving a sensitivity of ideally 4 pm/kPa, good repeatability with CV values below 5%, and minimal hysteresis. This optimisation will be guided by simulations using the embedded sensor ABAQUS model described in Chapter 4.3.2.2. The goal is to identify design adjustments that can lead to increased simulated sensitivity, which can then be validated experimentally.

To achieve this, the following parameters will be investigated:

- 1) Sensor Thickness: Increasing the thickness of the sensor to assess its impact on sensitivity and hysteresis.
- 2) Anchors: Adding anchors to the sensor to induce axial strain on the fibre, potentially enhancing sensitivity and reducing hysteresis.

By systematically varying these parameters, the goal of this chapter is to identify configurations that enhance the sensor's performance, achieving higher sensitivity and better repeatability in line with the target specifications.

4.4.2 Sensor Thickness

The first parameter that was altered in the embedded sensor ABAQUS model was the thickness of the sensor. This was investigated based on the hypothesis that by increasing the sensor's thickness, it is expected that the material will deform more under the same load resulting in increased strain along the fibre and a larger wavelength shift.

4.4.2.1 Modelling Methods

To test how thickness affects sensitivity, the embedded model was altered to have a 2mm and 3mm thick patch. A pressure of 1kPa, 5kPa, and 10kPa was applied to the 1mm model described in Chapter 4.3.2.2 as well as the new 2mm and 3mm models.

For these new models, the same model parameters as those displayed on Table 4.1 were used; except the thickness was changed from 1mm to 2mm and 3mm.

4.4.2.2 Modelling Results and Discussion

Once the pressures were simulated, the models showed some deformation. Figure 4.12 shows both the 2mm and 3mm model deformation.

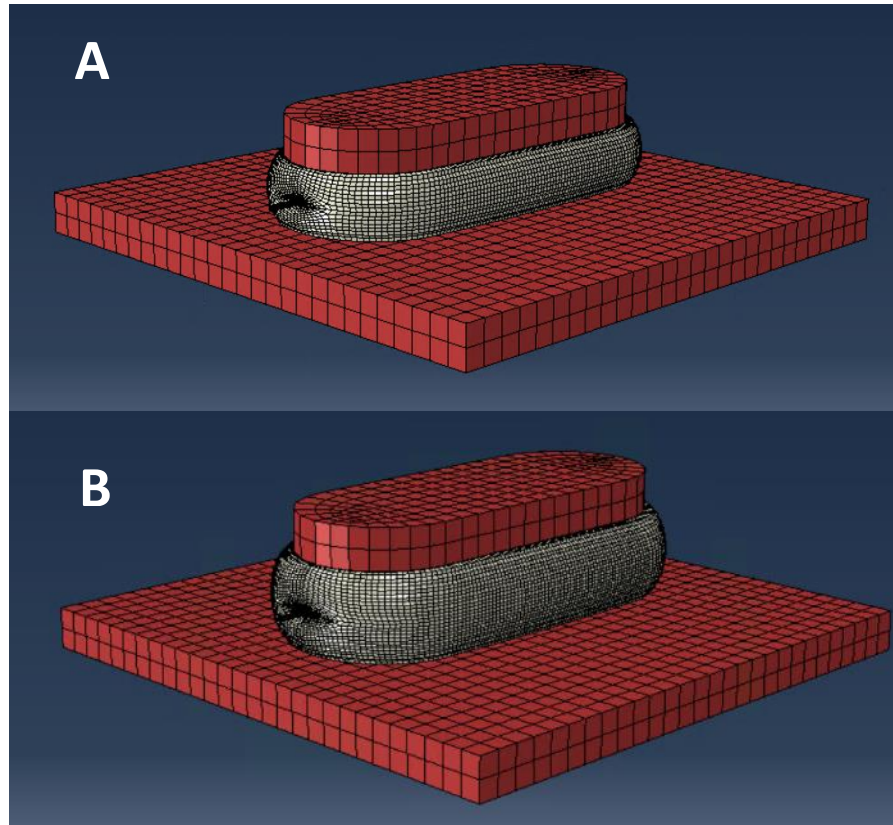


Figure 4.12: A: the deformed 2mm model after running the simulation with 5kPa pressure applied. B: the deformed 3mm model after running the simulation with 5kPa pressure applied. Scale- each red grid box for the mesh of the surface and plate is 0.5mm by 0.5mm

In Figures 4.12a and 4.12b it is clearly visible that in the location where the fibre exits the patch, the expansion is limited (Like in Figures 3.22b and 4.10b). This is probably due to friction between the fibre and the patch restricting the expansion of the patch in this specific region. It is worth investigating methods to prevent this from occurring as it is the pull of the patch in this region that causes the increased axial strain along the fibre. If the friction is limiting this expansion, then it is also limiting the axial strain induced.

Figure 4.13 shows the simulation results for all three models.

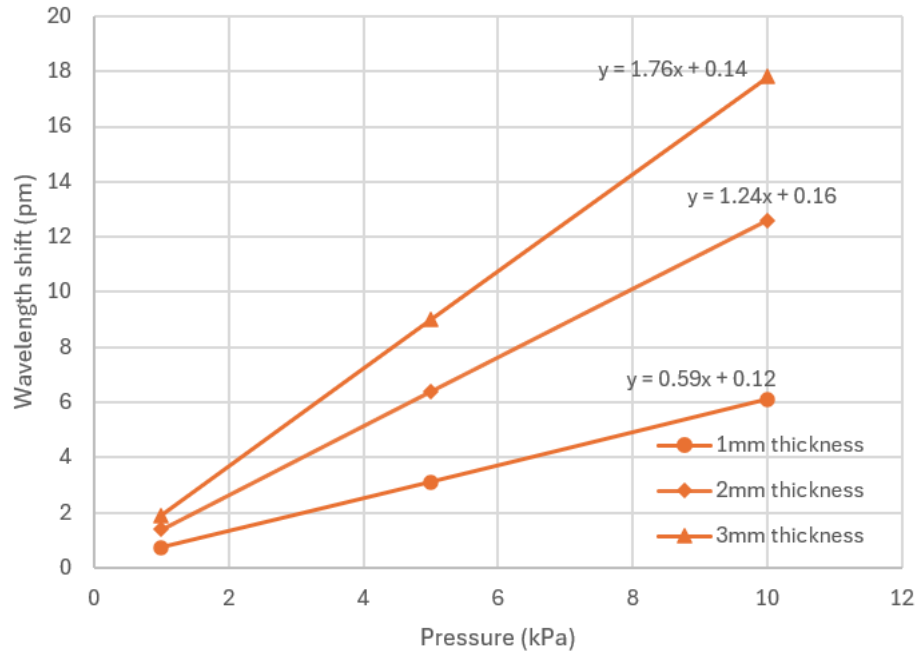


Figure 4.13: Simulated results for the 1mm, 2mm and 3mm SYLGARD embedded sensor model with their sensitivities.

Figure 4.13 suggests that by adding a millimetre to the thickness so that the sensor is 2mm thick, the sensitivity doubles (from 0.59pm/kPa to 1.24pm/kPa) and by adding two millimetres to the thickness so that the sensor is 3mm thick, the sensitivity approximately triples (from 0.59pm/kPa to 1.76pm/kPa). For elastomer materials like SYLGARD, a thicker sensor can undergo more significant deformation under the same applied pressure. This increased deformation leads to greater strain on the fibre and a larger wavelength shift, enhancing sensitivity. It is important to note that these results specifically represent the SYLGARD material as the model contains SYLGARDs material properties and therefore it cannot be assumed that the Dragon Skin will also double in sensitivity by increasing its thickness to 2mm.

It is also important to note that although optimising sensor sensitivity is necessary, the sensor needs to remain as small as possible as its application requires it to be placed on the ETT cuff. Although the simulated sensitivity of the 3mm sensor is higher than that of the 2mm sensor, it is difficult to tell how it will perform on the ETT cuff. A larger sensor could create passageways for aspirate to flow into the lungs. For this reason, only a sensor that is 2mm thick will be created for further experimentation.

4.4.2.3 Fabrication Process Methods

BlueFrog Design created a new mould for the fabrication of a sensor that is 2mm thick with all other dimensions remaining the same. This new 2mm thick mould is shown in Figure 4.14.

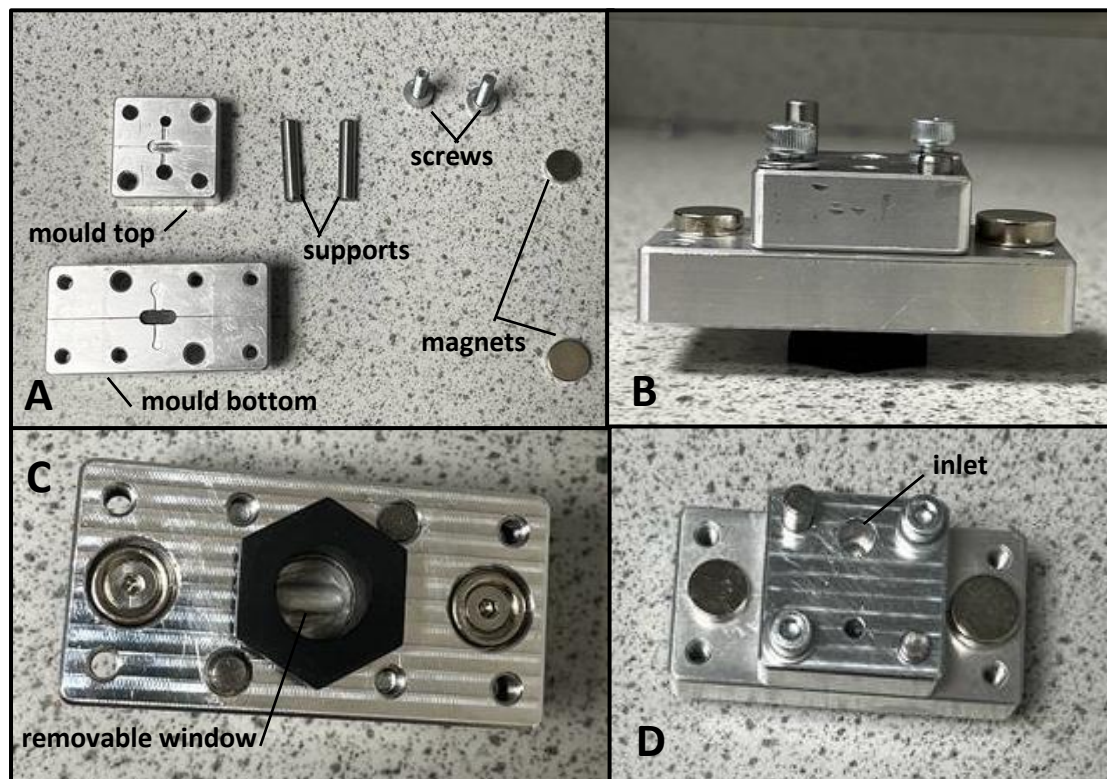


Figure 4.14: A: mould parts, B: mould assembled, C: mould from the bottom showing removable window into the inside of the mould, D: assembled mould and casting solution placed in large hole.

Overall, the new mould is similar to the 1mm sensor mould except that it contains a window on the underside. This allows the user to see the solution in the mould making sure that the whole space is filled with the casting solution. The window also allows for UV curing if needed.

Using this new mould, two sensors were created, one from SYLGARD and the other from Dragon Skin. The fabrication method used to create these sensors was the same as that described in Chapter 4.2.3. However, the curing procedure for the 2mm SYLGARD sensor changed slightly. This is because when using the 1mm SYLGARD sensors' curing directions, the sensor didn't fully cure. To cure the 2mm SYLGARD sensor, once the solution was placed into the mould the mould was then placed into the oven for 4 hours at 80°C then removed and left to cure overnight at room temperature. For the 2mm Dragon Skin sensor, the procedure remained the same- it was left to cure overnight at room temperature.

4.4.2.4 Fabrication Process Results and Discussion

The cured sensors were then removed from the mould. Figure 4.15 shows the cured 2mm sensors against the 1mm sensors for comparison.

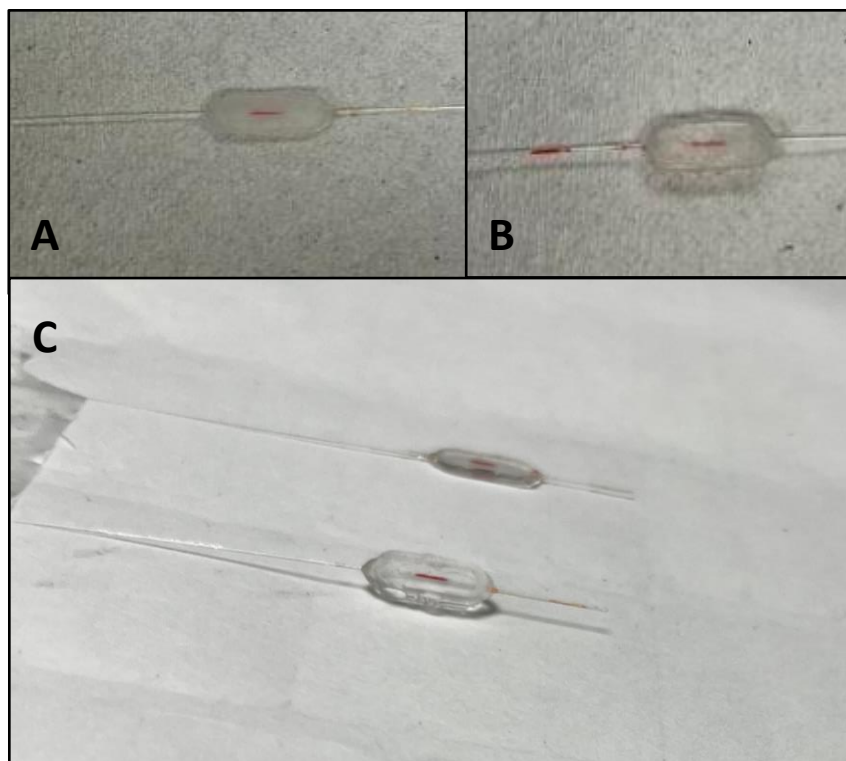


Figure 4.15: A: 1mm Dragon Skin, B: 2mm SYLGARD, C: 1mm SYLGARD (back) and 2mm SYLGARD (front) viewed from the side, sensors had a length of 10mm and a width of 4mm.

As shown in Figures 4.15a and 4.15b, the 2mm and 1mm sensors look almost identical. The difference in their thickness can be seen in Figure 4.15c. The dimensions were checked with a micrometre. Measurements all matched those of the mould confirming that the mould creates reproducible sensors.

4.4.2.5 Experimental Methods

Now the 2mm SYLGARD and Dragon Skin embedded sensors have been created, they are ready to be tested. In order for the results to be comparable to their 1mm sensor counterparts, the experimental setup is the same as that described in Chapter 4.3.2.1.

4.4.2.6 Experimental Results and Discussion

Experiments were carried out for both the 2mm SYLGARD and Dragon Skin embedded sensors. Experiments were then repeated two more times. Figure 4.16 shows the results for the 2mm SYLGARD sensor.

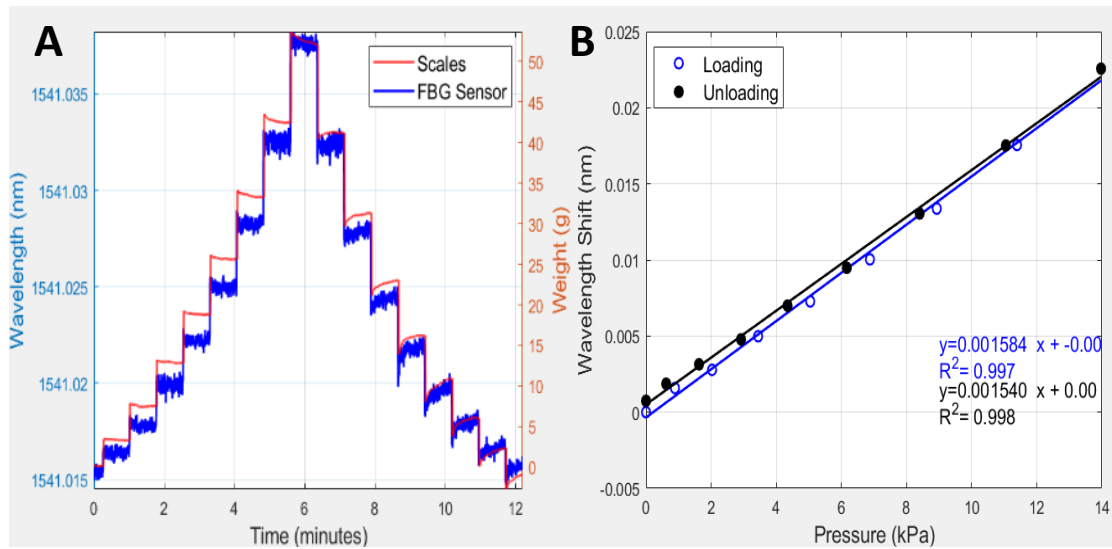


Figure 4.16: Experimental results for 2mm SYLGARD. A: wavelength vs time (from interrogator) and weight vs time (from scale), B: calibration curve.

Figure 4.16a shows that the loading and unloading cycles have the same number of steps and is no longer uneven. This is shown in both the scale and FBG response. In addition to this, there are no loading and unloading spikes visible from either the scales or the sensor, indicating a smooth response to pressure. Despite these improvements, the baseline at the end of the experiment is 0.75pm away from the starting baseline. This is also observed in the repeated experiments. Figure 4.16b shows a strong linear response ($R^2 = 0.99$) and the results of the repeated tests also show this. The CV for the 2mm SYLGARD sensor is 6.02% over the three experiments which shows moderate repeatability. The overall sensitivity (by averaging) of the 2mm SYLGARD has increased to $1.6 \pm 0.096\text{pm/kPa}$ with a hysteresis error of $1.49 \pm 0.89\text{pm}$.

Figure 4.17 shows the results for the 2mm Dragon Skin sensor.

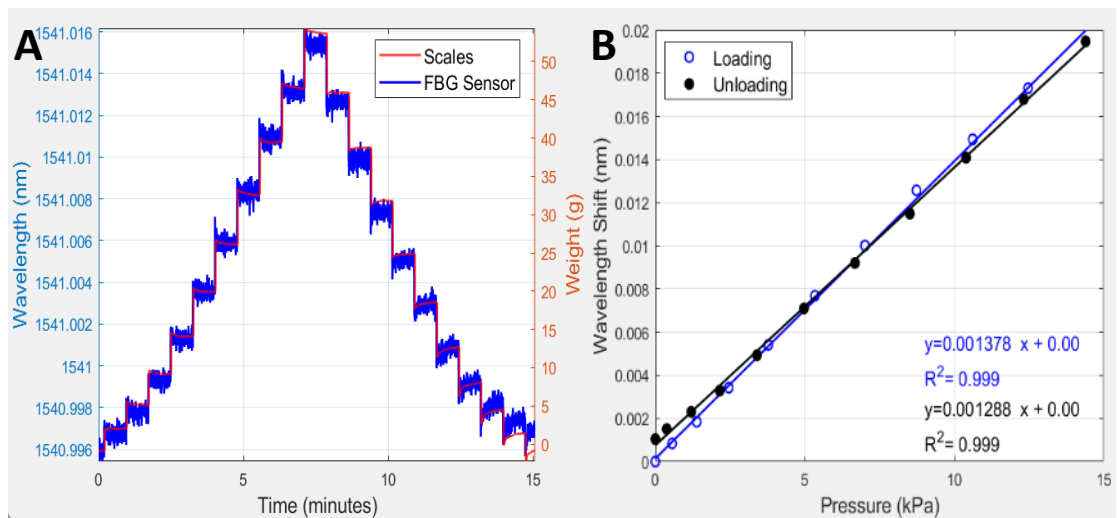


Figure 4.17: Experimental results for 2mm Dragon Skin. A: wavelength vs time (from interrogator) and weight vs time (from scale), B: calibration curve.

Like Figure 4.16a, Figure 4.17a shows that the loading and unloading cycles of the scale have the same number of steps in both the scale and FBG response. A smooth response to pressure is also observed. It is also shown in Figure 4.17a that the baseline at the end of the experiment (1540.9964nm) isn't where the starting baseline is (1540.9959nm). This is also observed in the repeat experiments. Figure 4.17b shows a strong linear response ($R^2 = 0.99$) and the results of the repeated tests also show this linear response. The sensitivities of the sensor over the three experiments are similar with a CV of 3.40% which indicates good repeatability. The sensitivity of the 2mm Dragon Skin sensor has increased to $1.36 \pm 0.046\text{pm/kPa}$ with a hysteresis error of $0.84 \pm 0.65\text{pm}$. It is also worth noting for these experiments that the hysteresis error is low with experiment 2 in particular showing very low hysteresis of 0.088pm.

These results show that although the 2mm SYLGARD sensor has higher sensitivity to the Dragon Skin sensor, the Dragon Skin sensor performed showed better repeatability than the SYLGARD sensor with a CV of 3.40%. When comparing the hysteresis error values for the two sensors, Dragon Skin shows slightly lower hysteresis error (0.84pm compared to 1.49pm of SYLGARD).

4.4.2.7 2mm Results vs 1mm Results

Having completed the 2mm sensor tests, results can be compared to their 1mm sensor counterparts. Table 4.2 shows a comparison of the 1mm and 2mm sensor sensitivities.

Table 4.2: 1mm SYLGARD and Dragon Skin sensitivity against the 2mm SYLGARD and Dragon Skin sensitivity.

1mm SYLGARD	2mm SYLGARD
$0.82 \pm 0.051 \text{ pm/kPa}$	$1.6 \pm 0.096 \text{ pm/kPa}$
1mm Dragon Skin	2mm Dragon Skin
$0.99 \pm 0.055 \text{ pm/kPa}$	$1.36 \pm 0.046 \text{ pm/kPa.}$

From Table 4.2, it is clear the sensitivity increased for both sensors when the thickness was increased to 2mm. It is visible that the SYLGARD sensitivity approximately doubled when making the sensor 2mm. This was predicted by the ABAQUS 2mm model and further indicates its credibility. For the 2mm Dragon Skin sensor, although the sensitivity increased it did not increase as much as the SYLGARD sensor did (an increase of 0.37pm opposed to an increase of 0.78pm for SYLGARD), this is thought

to be due to its material properties. As predicted with the 2mm model results in Chapter 4.4.2.2, although the increase in sensitivity is due to material thickness, the material properties play a key role in determining how much the sensitivity will increase.

When comparing the hysteresis error, the SYLGARD sensors had a hysteresis error of 0.26pm for the 1mm sensor and 1.49pm for the 2mm sensor. The 1mm Dragon Skin sensor had a hysteresis error of 0.4pm and the 2mm Dragon Skin has a hysteresis error of 0.84pm. The increase in sensor thickness has increased the effect of hysteresis on both sensors.

Lastly, the CV for the SYLGARD sensor decreased from 6.22% as a 1mm sensor to 6.02% as a 2mm sensor and the CV for the Dragon Skin sensor decreased from 5.54% as a 1mm sensor to 3.40% as a 2mm sensor.

When comparing these results to the target specification, the sensors show a desirable linear response with some hysteresis and the repeatability of the Dragon Skin sensor shows good repeatability. However, the repeatability needs improving for the SYLGARD sensor in order to show good repeatability. Additionally, the sensitivity of the sensors are still significantly below the ideal sensitivity of 4pm/kPa. Further optimisation is required to better align the sensor performance with the application requirements.

4.4.2.8 2mm Experimental Results vs Simulated Results

The pressures used in the three 2mm SYLGARD experiments were input into the 2mm ABAQUS model and the resultant wavelength shifts were simulated. Figure 4.18 shows the experimental and simulated results.

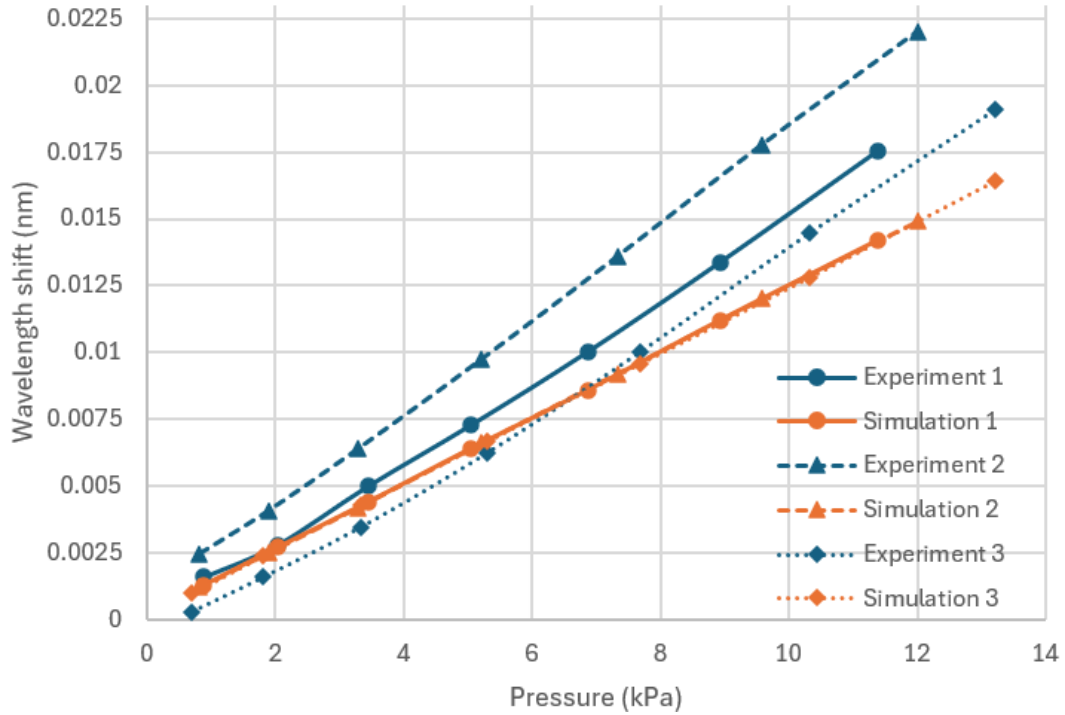


Figure 4.18: 2mm SYLGARD experimental results and ABAQUS simulated results.

Figure 4.18 shows that the wavelength shift observed experimentally is larger than the simulated wavelength shift. When comparing the 2mm simulations to the 1mm simulations (in Chapter 4.3.3.3), the 2mm sensor simulations have larger discrepancies to the experimental results. In the experiments the sensitivity of the 2mm SYLGARD sensor was 1.6 ± 0.096 pm/kPa, the model results show a sensitivity of 1.24pm/kPa. As mentioned previously, this discrepancy could be due to a number of factors including the non-linearity of the experimental results, the few material properties incorporated in the model (as shown in Table 3.2), the weighing scales readings as they do not stabilise after each pressure increment and favouring computational timing over result accuracy when meshing the model. Despite these discrepancies, the PCC between the simulated and experimental results indicates that there is a significant positive correlation ($r(23)= 0.977$, $p < 0.001$). This shows lower correlation between the simulated and experimental results than those observed in the 1mm embedded

experiments ($r(21) = 0.993$, $p < 0.001$). However, these still show high correlation and certainty between the results.

4.4.3 Changes in Material Properties

The next parameter that was altered in the embedded sensor ABAQUS model was the material properties of the sensor patch. This was done to see what material properties would result in a more sensitive sensor. The material properties that were altered were the Poisson's Ratio of the patch material as well as its Young's Modulus.

4.4.3.1 Modelling Methods

For these simulations, a pressure of 0.005MPa was applied to the 1mm embedded model described in Section 4.3.2.2. The same model parameters as those displayed on Figure 4.1 were used except the following parameters were changed:

a) Poisson's Ratio

When testing to see how Poisson's Ratio affects sensitivity, the patch in the 1mm embedded model was altered to have Poisson's Ratio of 0.05, 0.1, 0.15, 0.2, 0.25, 0.3, 0.35, 0.4, and 0.45.

b) Young's Modulus

When testing to see how the Young's Modulus affects sensitivity, the patch in the 1mm embedded model was altered to have a Young's Modulus (in MPa) of 0.5, 1, 10, 50, 100, 500, and 1000.

4.4.3.2 Modelling Results and Discussion

a) Poisson's Ratio

Once the simulations were complete the results were collected and plotted in Figure 4.19 to show how the material's Poisson's Ratio affects the sensitivity.

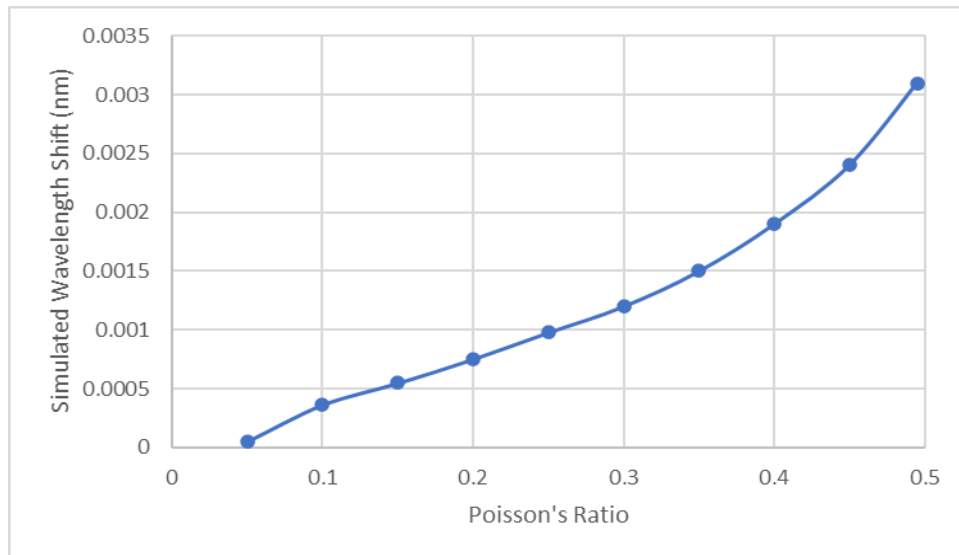


Figure 4.19: Graph showing how material Poisson's Ratio affects wavelength shift for a load of 5kPa.

ABAQUS does not allow simulations to run if materials have a Poisson's Ratio more than 0.495. Dragon Skin has a Poisson's Ratio of 0.4999 (Sparks *et al.*, 2015). From Figure 4.19, larger Poisson's Ratios cause larger wavelength shifts. This is due to the fact that the Poisson's ratio determines the expansion of a material perpendicular to the loading direction. The larger the Poisson's ratio of a material the more the material will expand when a load is compressing it, and therefore the more longitudinal strain it will apply on the embedded fibre. The Poisson's Ratio of a material cannot exceed 0.4999 (Mott & Roland, 2009). From Figure 4.19, it would be reasonable to assume that polymers with a Poisson's Ratio even closer to 0.4999, like Dragon Skin, may result in a slightly larger wavelength shift for the same pressure applied. Interestingly, from the experiments carried out and displayed in Figures 4.8 and 4.9 for the 1mm sensors, Dragon Skin shows a higher sensitivity than SYLGARD.

b) Young's Modulus

Table 4.3 shows how the Young's Modulus of a material effects the wavelength shift. The row with the black border is the Young's Modulus of SYLGARD, the row with the orange border is the Young's Modulus of Dragon Skin (1.11MPa) (Sandoval *et al.*, 2019). Figure 4.20 shows a plot of how the material's Young's Modulus affects the sensitivity.

Table 4.3: Simulated results for different Young's Modulus values for the embedded patch material.

Young's Modulus (MPa)	Wavelength Shift (nm)
0.5	0.0030
1	0.0031
2	0.0031
10	0.0034
50	0.0035
100	0.0033
500	0.0022
1000	0.0016

Table 4.3 shows that the wavelength shift experienced increases as the Young's Modulus of the patch increases to around 50MPa, and then decreases beyond that (as the Young's Modulus increases). The Young's Modulus of both SYLGARD and Dragon Skin are below 50MPa. The Young's Modulus of rubber can range from 1MPa to 100MPa (*Young's Modulus of Materials*, Bestech Australia), and therefore it may be possible to find a material with a Poisson's Ratio closer to 0.4999 and with a Young's Modulus of around 50MPa for optimum sensitivity.

The effect of the materials Young's Modulus on the simulated wavelength shift can be seen in Figure 4.20

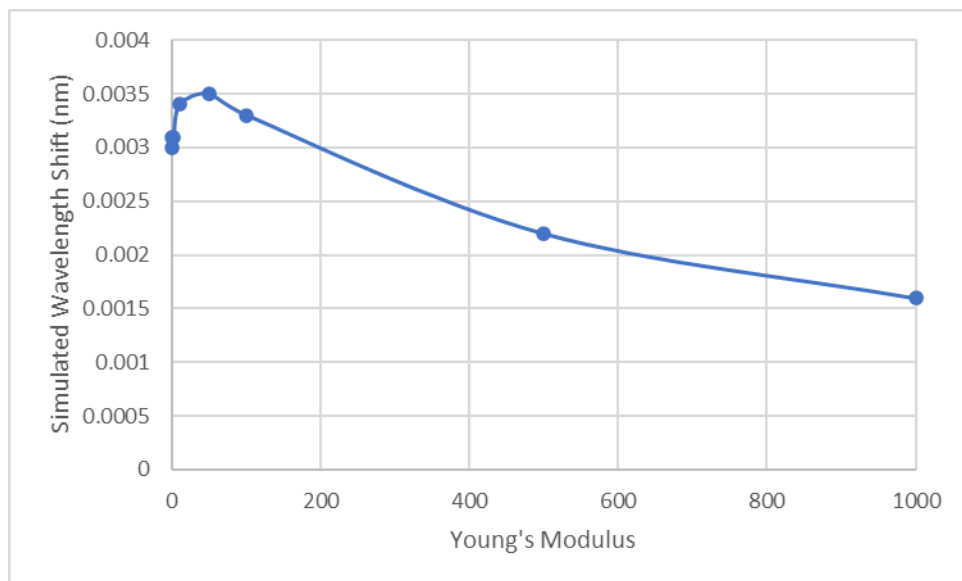


Figure 4.20: Graph showing how material Young's Modulus affects wavelength shift for a load of 5kPa.

Figure 4.20 demonstrates that as the Young's Modulus increases, the wavelength shift decreases. This is expected because a larger Young's Modulus indicates a less deformable material. Conversely, one might assume that a smaller Young's Modulus would lead to more material deformation and, therefore, a larger wavelength shift. However, Figure 4.20 shows this is not the case, as the wavelength shift decreases when the Young's Modulus falls below 50 MPa.

A lower Young's Modulus indicates greater deformability, meaning the material undergoes significant deformation when stress is applied, expanding along the fibre. If the material is too deformable, friction between the material and the fibre constrains its expansion. As the Young's Modulus increases, the material becomes less yielding, and under the same stress, friction imposes fewer constraints on expansion, resulting in greater strain on the fibre and a larger wavelength shift. At 50 MPa, the material's deformability surpasses the constraining effects of friction, leading to the maximum wavelength shift. Beyond this point, the material becomes increasingly rigid, limiting expansion and exerting less strain on the fibre.

This phenomenon is seen in the model deformation Figures (Figures 3.22b, 4.10b, and 4.12) where patch expansion is restricted at the interface between the fibre and the patch. When considering methods to increase sensitivity, it is crucial to address this region and explore ways to enhance the effect of expansion (induced strain) along the fibre.

4.4.4 Conclusion

The ABAQUS model was used to explore how variations in sensor thickness affect the wavelength shift. Simulations indicated that increasing the sensor thickness significantly enhances sensitivity: doubling for 2mm thick sensors and tripling for 3mm thick sensors. These findings were validated experimentally, with both SYLGARD and Dragon Skin 2mm sensors showing improved sensitivity, and SYLGARD's sensitivity approximately doubling. Additionally, Dragon Skin exhibited good repeatability (CV of 3.40%). However, hysteresis increased for both 2mm sensors, measuring 0.84pm for Dragon Skin and 1.49pm for SYLGARD. Despite these enhancements, the sensitivities of both materials remain below the ideal 4 pm/kPa, achieving 1.36 pm/kPa for Dragon Skin and 1.6 pm/kPa for SYLGARD, highlighting the need for further optimisation.

Material properties such as Poisson's Ratio and Young's Modulus were also investigated for their impact on sensitivity. Results showed that larger Poisson's Ratios lead to greater wavelength shifts due to increased longitudinal strain on the fibre. The study found that materials with a Young's Modulus around 50 MPa provided the highest sensitivity before becoming too rigid to deform significantly under pressure. Additionally, addressing expansion constraints at the fibre-patch interface, where friction limits the material's deformation, is crucial. Optimising this region to enhance the induced strain along the fibre has the potential to further improve sensor sensitivity.

4.5 Optimisation of Embedded Sensor Sensitivity Using Epoxy Anchors

4.5.1 Introduction

After looking into how to optimise the sensor using the model. Further testing was done to see if the sensor could be optimised by adding epoxy anchors at either end of the polymer patches. The theory behind this was to create slight resistance to the patch expanding when a load is applied. As seen in the FEA model results (shown in both Figure 4.12 and 4.20) when the patch expands there is resistance to expansion at the fibre-patch interface around the egress of the fibre. If an epoxy anchor is placed in this location, this resistance to expansion is negated as the patch will push on the anchors. The anchors will resist the expansion however, as they are strongly bonded to the fibre, this will cause an increase in induced strain along the fibre causing a larger wavelength shift. Figure 4.21 shows what these anchors are.

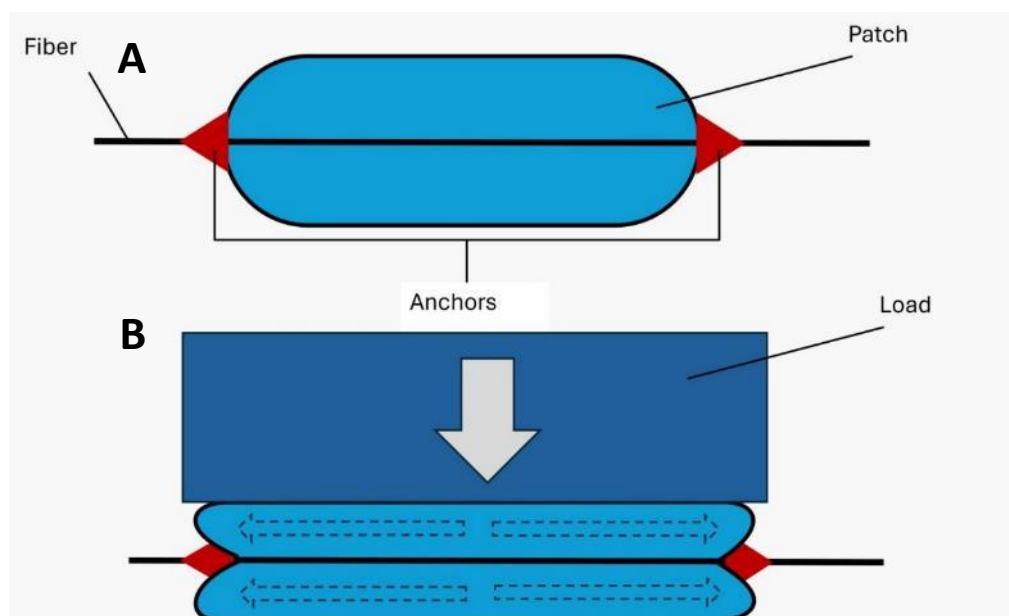


Figure 4.21: A: Diagram of anchors (in red) placed on either side of the patch. B: Showing what happens when a load is applied to the patch with anchors. Anchors limiting expansion and inducing increased strain along the fibre.

To test this theory out, anchors were placed on all four of the sensors (the 1mm and 2mm SYLGARD sensors and the 1mm and 2mm Dragon Skin sensors).

4.5.2 Fabrication Methods

To apply the anchors onto the four embedded sensors, a fibre tip was placed into some epoxy. This resulted in a small droplet of epoxy gathered at the tip of the fibre. Using the fibre tip, the small ball of epoxy was carefully placed on either end of the patch. The sensor was then placed under a UV torch to cure for 30 minutes. During this process it was important that the diameter of the anchors was less than the thickness of the patch. This was to ensure that when applying the load to the patch, the patch would still act as it should and compress vertically.

4.5.3 Fabrication Results and Discussion

Once the anchors were cured the UV torch was removed. Figure 4.22 shows the sensors once the anchors had cured.

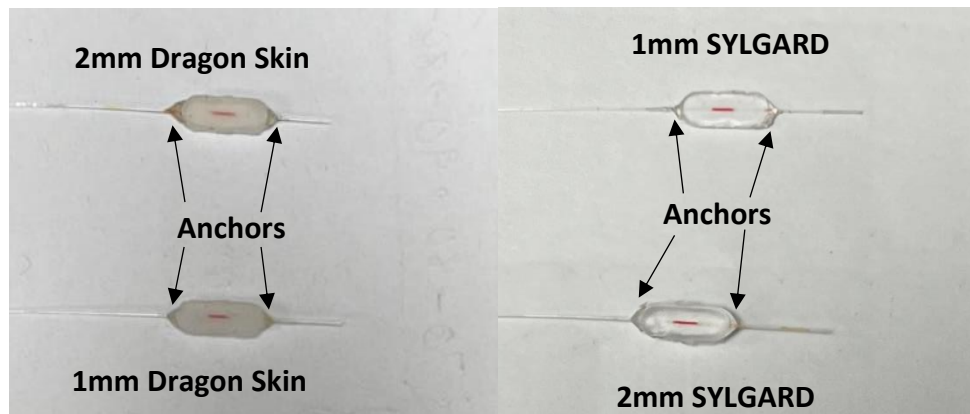


Figure 4.22: All four sensors with epoxy anchors attached to them. It is quite difficult to see the anchors on the SYLGARD sensors as the cured epoxy looks very similar to the cured SYLGARD patch.

As shown in Figure 4.22, the sizes of the epoxy anchors vary from sensor to sensor. It is unclear whether the size of these anchors effects the sensitivity of the sensor. However, as these tests are proof of concept, a more detailed method of how to apply the epoxy anchors, how large they should be, and how to get them the same size every time should be carried out as part of future work.

4.5.4 Experimental Methods

For the results to be comparable to the ‘unanchored’ 1mm and 2mm SYLGARD and Dragon Skin sensors, the experimental setup is the same as that described in Chapter 4.3.2.1. It is worth mentioning that after completing the anchored 1mm SYLGARD and Dragon Skin sensor experiments and the anchored 2mm Dragon Skin experiments, the

scales broke before testing the anchored 2mm SYLGARD sensor. New weighing scales was purchased (S5i-6, GRAM) and the anchored 2mm SYLGARD sensor was tested on these new scales. These new weighing scales had a capacity of 6kg, with a resolution of 0.5g and used a load cell rather than EMFC (which was the working mechanism of the previous weighing scales).

4.5.4.1 Repeat Experiments

For these tests, three experiments were carried out with each sensor to see how it performs. These tests are similar to the ones performed for the ‘unanchored’ 1mm and 2mm experiments.

4.5.4.2 Cyclic Test

For these experiments, a test involving three loading and unloading cycles was carried out to see if each cycle had the same sensitivity as the others. If these cycles result in different sensitivities, the sensor is unreliable. These tests were carried out to see how these sensors would perform in an experiment where the pressures vary over time.

4.5.4.3 Statistical Analysis

The same statistical methods were incorporated as those used in Chapter 4.3.2.3.

4.5.5 Experimental Results and Discussion

Experiments were carried out for the 1mm SYLGARD and Dragon Skin embedded sensors as well as the 2mm SYLGARD and Dragon Skin embedded sensors.

4.5.5.1 Repeat Experiments

Experiments were carried out three times for each sensor. Figure 4.23 shows the results for the first anchored 1mm SYLGARD sensor experiment.

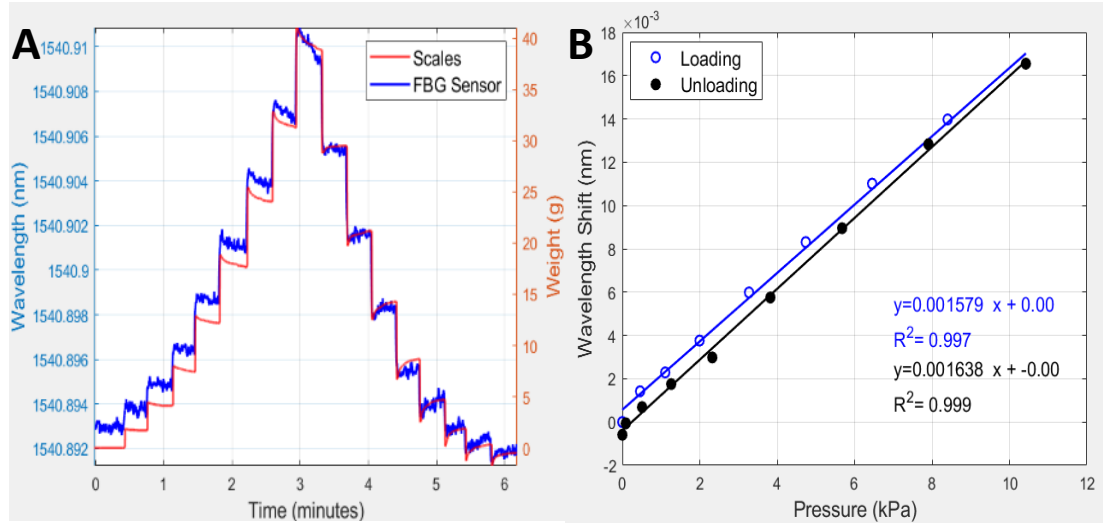


Figure 4.23: Experimental results for anchored 1mm SYLGARD. A: wavelength vs time (from interrogator) and weight vs time (from scale), B: calibration curve.

Figure 4.23a shows that the starting wavelength (1540.8931nm) is different from the ending wavelength (1540.8919nm). This is observed in two of the three experiments. The repeat experiments have a CV of 0.62% indicating high repeatability. The hysteresis error across all experiments is also shown to be low averaging at 0.39pm. The overall sensitivity (by averaging) of the anchored 1mm SYLGARD sensor is $1.61 \pm 0.01\text{pm/kPa}$ with a hysteresis error of $0.39 \pm 0.25\text{pm}$.

Figure 4.24 shows the results for the first anchored 1mm Dragon Skin sensor experiment.

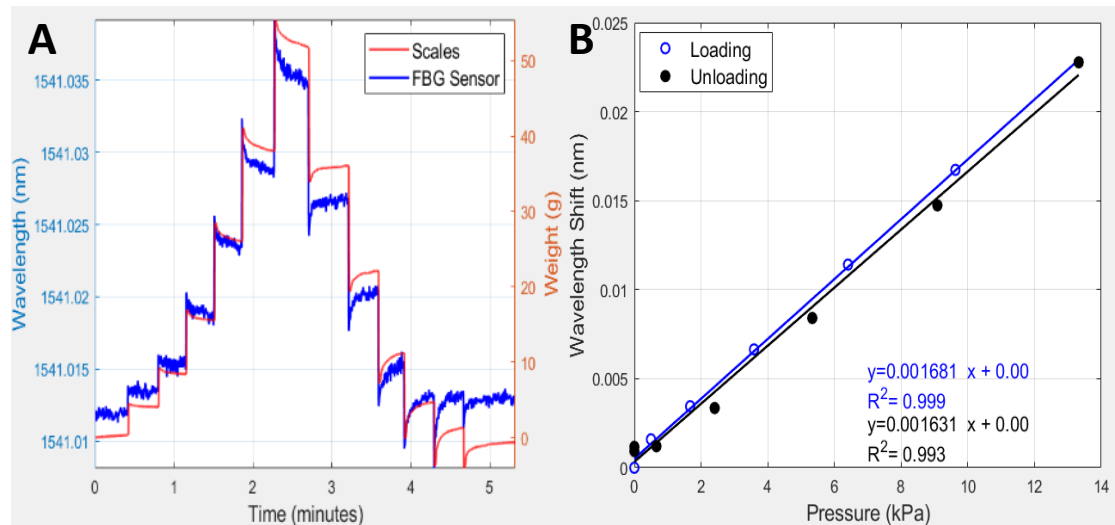


Figure 4.24: Experimental results for anchored 1mm Dragon Skin. A: wavelength vs time (from interrogator) and weight vs time (from scale), B: calibration curve.

Figure 4.24a shows that's the starting wavelength (1541.0121nm) is slightly different from the ending wavelength (1541.0129nm). This is also seen throughout the repeat experiments. The CV across all three experiments was 3.26%. This indicates good repeatability. The sensitivity of the anchored 1mm Dragon Skin is $1.62 \pm 0.053\text{pm/kPa}$ with a hysteresis error of $0.92 \pm 0.22\text{pm}$.

Figure 4.25 shows the results for the first anchored 2mm SYLGARD sensor experiment.

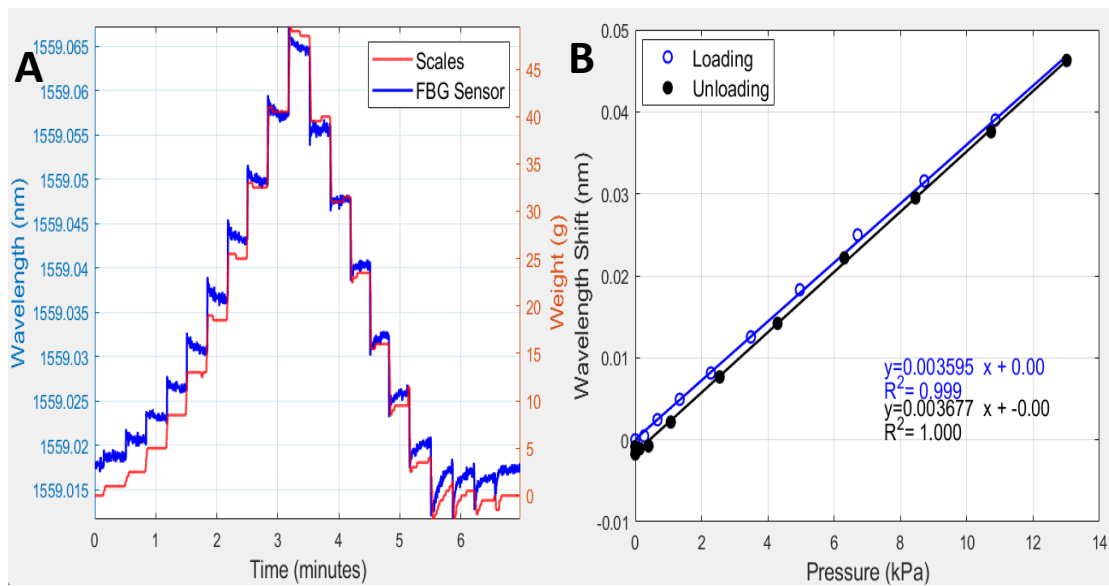


Figure 4.25: Experimental results for anchored 2mm SYLGARD. A: wavelength vs time (from interrogator) and weight vs time (from scale), B: calibration curve.

Figure 4.25a shows that the starting wavelength is similar to the ending wavelength. This was also seen in the repeat experiments. As this experiment was performed with the new scales, it can be seen that the scales settle significantly faster than the old scales (as seen in other experiments). This is thought to be due to these scales utilising a load cell as their mechanism for measurement whereas the old scale used EMFC. All three experiments showed a similar sensitivity with a CV of 1.13%. This indicates good repeatability. The sensitivity of the anchored 2mm SYLGARD sensor is $3.67 \pm 0.042\text{pm/kPa}$ with a low hysteresis error of $0.52 \pm 0.49\text{pm}$.

Figure 4.26 shows the results for the first anchored 2mm Dragon Skin sensor experiment.

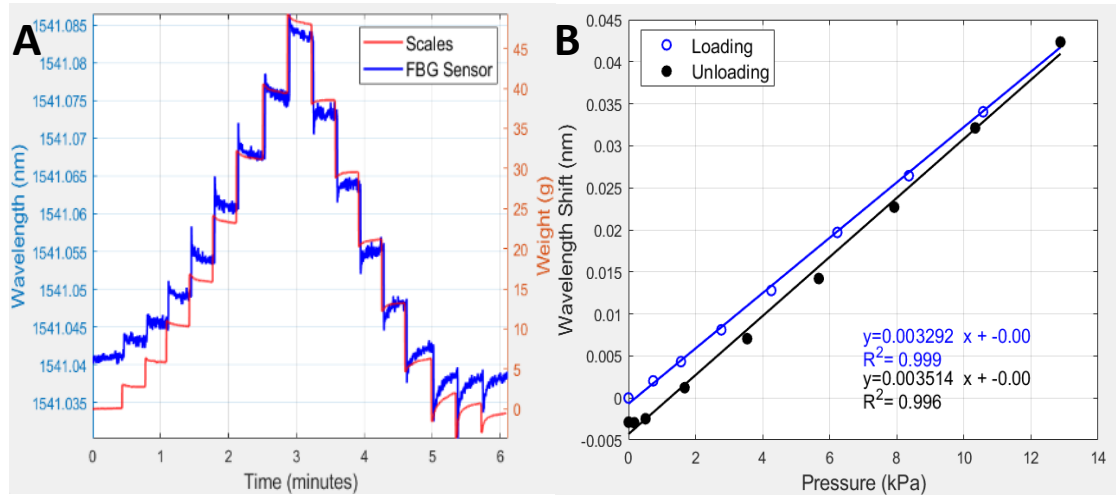


Figure 4.26: Experimental results for anchored 2mm Dragon Skin. A: wavelength vs time (from interrogator) and weight vs time (from scale), B: calibration curve.

Figure 4.26a shows that the starting wavelength (1541.0407nm) is slightly different from the ending wavelength (1541.0391nm). This was not seen in the repeat experiments. The sensitivities across all three experiments were similar with a CV of 1.75%. This indicates good repeatability. The sensitivity of the anchored 2mm Dragon Skin sensor is $3.47 \pm 0.061\text{pm/kPa}$ with a hysteresis error of $1.4 \pm 1.3\text{pm}$.

4.5.5.2 Cyclic Test

For these experiments each sensor underwent three cycles of loading and unloading.

Figure 4.27 shows the anchored 1mm SYLGARD sensor's response during the experiment. The time series for each experiment looks similar to Figure 4.27 and therefore will not be shown here. The calibration curves however, will be shown.

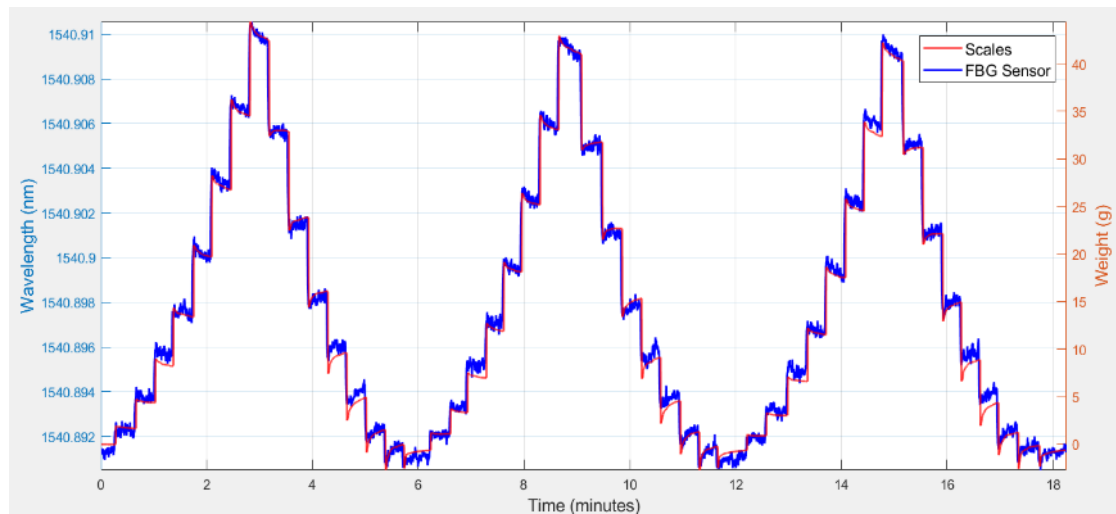


Figure 4.27: Experimental results for anchored 1mm SYLGARD, wavelength vs time (from interrogator) and weight vs time (from scale)

Figure 4.28 shows the calibration curves for each of the loading and unloading cycles.

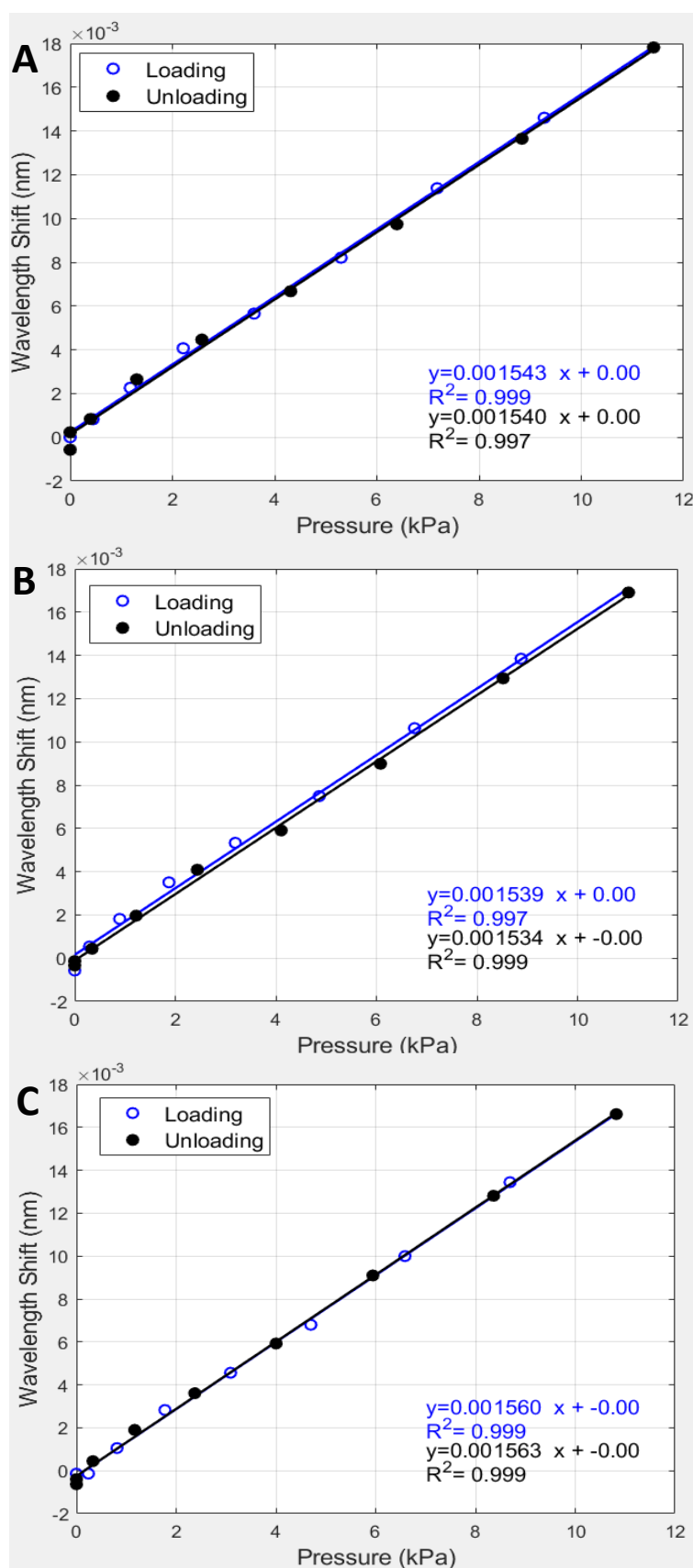


Figure 4.28: Calibration curves for the anchored 1mm SYLGARD sensor. A: calibration curve for the first cycle, B: calibration curve for the second cycle, and C: calibration curve for the third cycle.

As shown in Figure 4.27 the response of the sensor closely follows the scale response throughout all three loading and unloading cycles with the ending wavelength returning to baseline. Figure 4.28 shows that the response from this sensor is strongly linear with very little hysteresis. Each loading and unloading cycle of the experiment resulted in a similar sensitivity with a CV of just 0.75%. This indicates strong repeatability. From this experiment, the anchored 1mm SYLGARD sensitivity is $1.55 \pm 0.012 \text{ pm/kPa}$ with a very low hysteresis error of $0.04 \pm 0.01 \text{ pm}$.

Figure 4.29 shows the calibration curves for each of the loading and unloading cycles of the anchored 1mm Dragon Skin sensor.

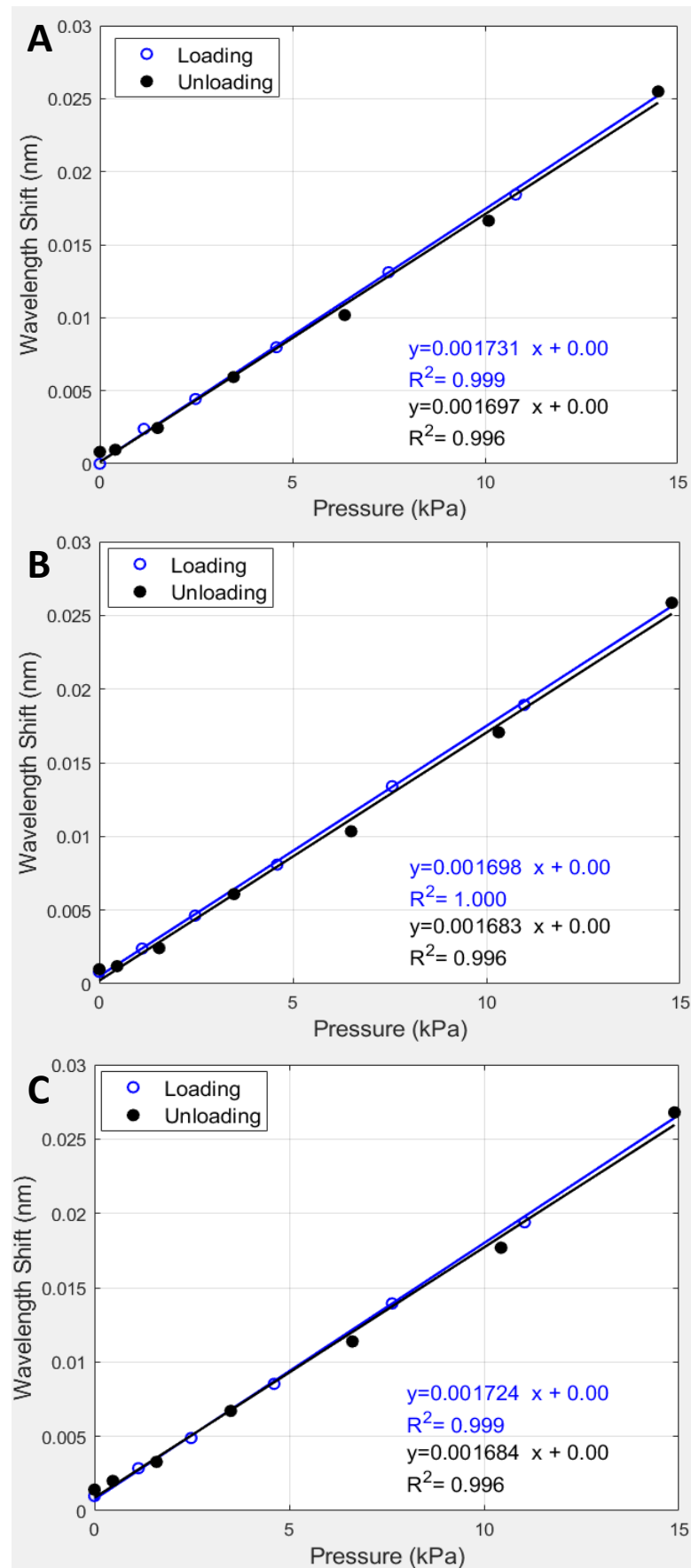


Figure 4.29: Calibration curves for the anchored 1mm Dragon Skin sensor. A: calibration curve for the first cycle, B: calibration curve for the second cycle, and C: calibration curve for the third cycle.

Like the anchored 1mm SYLGARD sensor, Figure 4.29 shows that the response from this sensor is also strongly linear. Each loading and unloading cycle of the experiment resulted in similar sensitivities with a CV of just 0.59%. This indicates strong repeatability. This experiment shows that the anchored 1mm Dragon Skin sensor has a sensitivity of $1.7 \pm 0.01 \text{ pm/kPa}$ with a hysteresis error of $0.43 \pm 0.19 \text{ pm}$.

When comparing the two 1mm anchored sensors, the hysteresis error of the anchored 1mm Dragon Skin sensor is larger than that calculated for the anchored 1mm SYLGARD sensor. However, the CV for the Dragon Skin is smaller than that of the SYLGARD sensor.

When comparing these results to the target specification, the sensors show a desirable linear response with very little hysteresis and good repeatability. However, the sensitivity of the sensors are still significantly below the ideal sensitivity of 4 pm/kPa .

Figure 4.30 shows the calibration curves for each of the loading and unloading cycles for the 2mm anchored SYLGARD sensor.

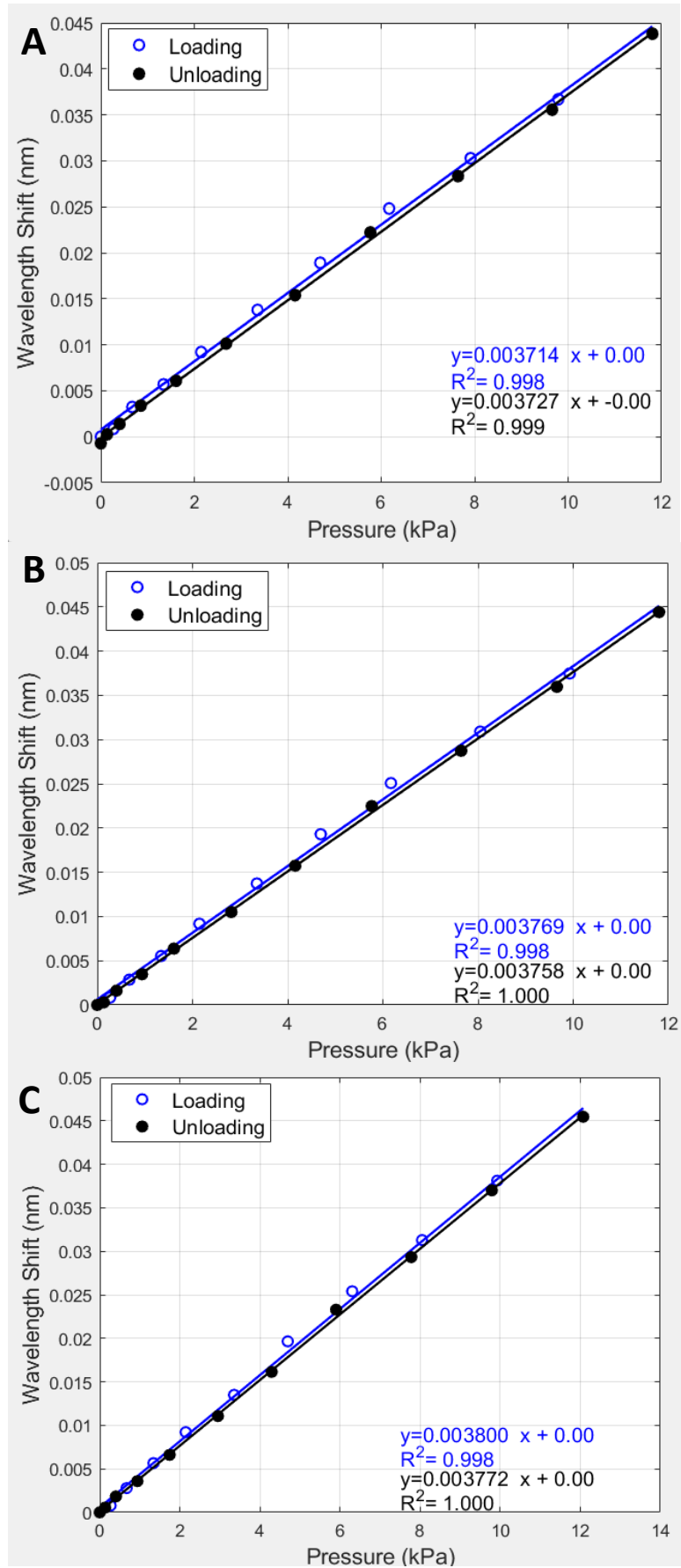


Figure 4.30: Calibration curves for the anchored 2mm SYLGARD sensor. A: calibration curve for the first cycle, B: calibration curve for the second cycle, and C: calibration curve for the third cycle.

Figure 4.30 shows that the response from this sensor is also strongly linear. Each loading and unloading cycle of the experiment resulted in similar sensitivities with a CV of just 0.81%. This indicates strong repeatability. This experiment shows that the anchored 2mm SYLGARD sensor has a sensitivity of $3.75 \pm 0.03\text{pm/kPa}$ with a low hysteresis error of $0.21 \pm 0.12\text{pm}$.

Figure 4.31 shows the calibration curves for each of the loading and unloading cycles for the anchored 2mm Dragon Skin sensor.

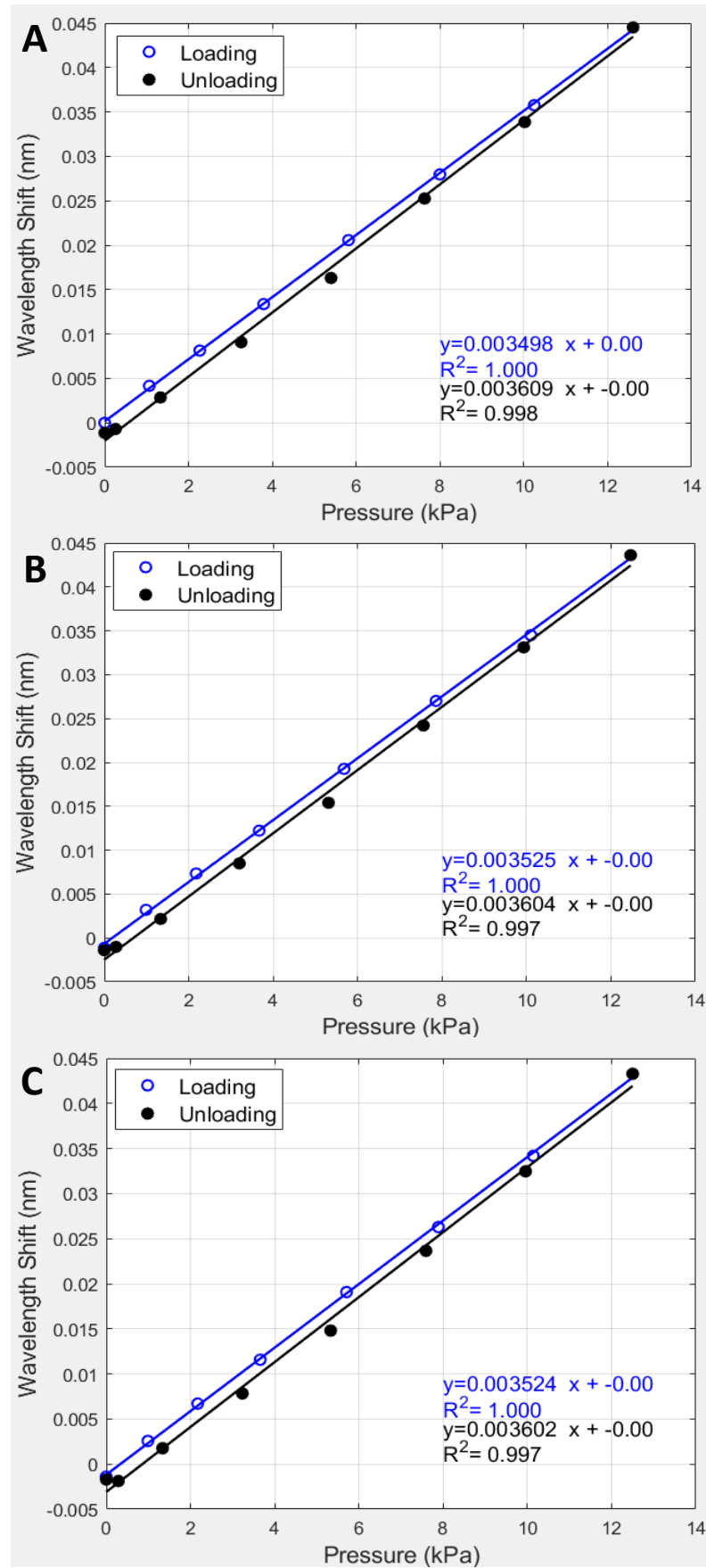


Figure 4.31: Calibration curves for the anchored 2mm Dragon Skin sensor. A: calibration curve for the first cycle, B: calibration curve for the second cycle, and C: calibration curve for the third cycle.

Figure 4.31 shows that the response from this sensor is also strongly linear. Each loading and unloading cycle of the experiment resulted in a similar sensitivity with a CV of just 0.16%. This indicates strong repeatability and is smaller than the CV of all other sensors. From these experiments, the sensitivity of the anchored 2mm Dragon Skin sensor is $3.56 \pm 0.0058\text{pm/kPa}$ with a hysteresis error of $1.12 \pm 0.24\text{pm}$.

When comparing the two anchored 2mm sensors, like for the anchored 1mm sensors, the hysteresis error calculated for the anchored 2mm Dragon Skin sensor ($1.12 \pm 0.24\text{pm}$) is larger than that calculated from the anchored 2mm SYLGARD sensor ($0.21 \pm 0.12\text{pm}$). However, the CV for the Dragon Skin (0.16%) is smaller than that of the SYLGARD sensor (0.81%). It is worth noting that as the scales used for the anchored 2mm SYLGARD experiments are different from the scales used for the anchored 2mm Dragon Skin experiments, these results may not be directly comparable and there is a possibility that some discrepancies are due to the scales being different. CV values for both sensors showed good repeatability.

When comparing these results to the target specification, the sensors show a desirable linear response and good repeatability with the 2mm anchored SYLGARD showing low hysteresis as well. Additionally, the sensitivity of the sensors are not far from the 4pm/kPa ideal sensitivity (with 3.75pm/kPa for the anchored 2mm SYLGARD and 3.56pm/kPa for the anchored 2mm Dragon Skin). These sensors have been optimised for tracheal contact pressure measurement and offer a resolution of 0.026pm/kPa for the anchored 2mm SYLGARD, and a resolution of 0.028pm/kPa for the anchored 2mm Dragon Skin.

4.5.6 Anchored vs Unanchored Results

Before comparing anchored sensors to ‘unanchored’ sensors, the ‘unanchored’ 2mm SYLGARD sensor was retested on the new scales. This would allow for a better comparison of the ‘unanchored’ and anchored 2mm SYLGARD sensors.

Figure 4.32 shows the results for the ‘unanchored’ 2mm SYLGARD sensor on the new scales.

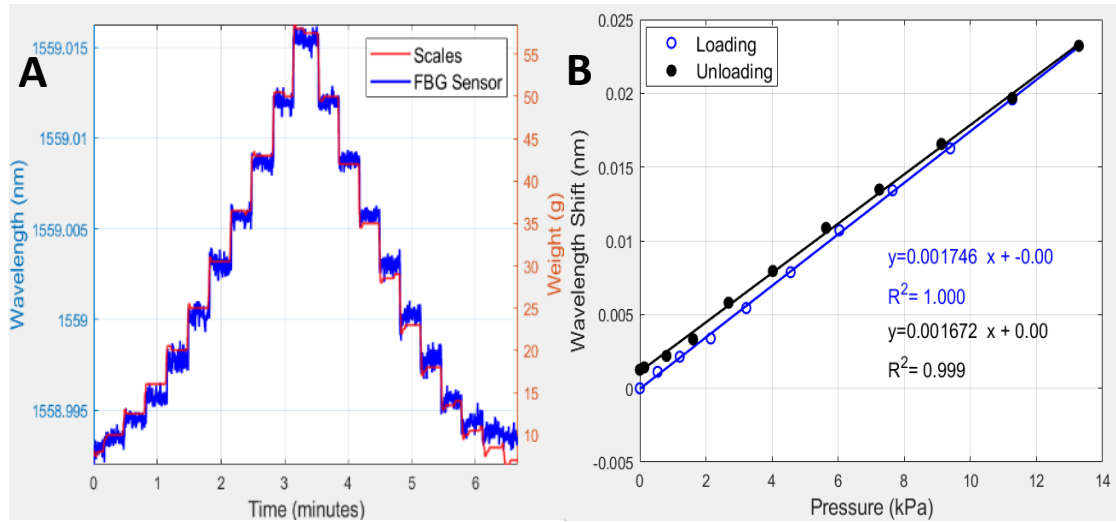


Figure 4.35: Experimental results for unanchored 2mm SYLGARD on the new scales. A: wavelength vs time (from interrogator) and weight vs time (from scale), B: calibration curve.

Results showed that the sensitivity of the sensor through all three experiments was similar with a CV of 0.90%. This indicates strong repeatability. The sensitivity of the ‘unanchored’ 2mm SYLGARD sensor is $1.69 \pm 0.015 \text{ pm/kPa}$ with a hysteresis error of $0.74 \pm 0.3 \text{ pm}$. The results from the ‘unanchored’ 2mm SYLGARD sensor using the new scales are not too far off the results of the same sensor on the old scales (which was calculated as $1.6 \pm 0.096 \text{ pm/kPa}$, as shown in Chapter 4.4.2.6). This slight discrepancy could be due to the different scales used in the experiments.

Having completed all the experiments, the anchored sensor results can be compared to the ‘unanchored’ sensor results. For the comparison, the sensitivities from the repeat experiments were utilised. Table 4.4 shows a comparison of the anchored and ‘unanchored’ sensor sensitivities in pm/kPa.

Table 4.4: Comparison of all sensor’s sensitivities

Unanchored 1mm SYLGARD	Anchored 1mm SYLGARD
$0.82 \pm 0.051 \text{ pm/kPa}$	$1.61 \pm 0.01 \text{ pm/kPa}$
Unanchored 1mm Dragon Skin	Anchored 1mm Dragon Skin
$0.99 \pm 0.055 \text{ pm/kPa}$	$1.62 \pm 0.053 \text{ pm/kPa}$
Unanchored 2mm SYLGARD	Anchored 2mm SYLGARD
$1.69 \pm 0.015 \text{ pm/kPa}$	$3.67 \pm 0.042 \text{ pm/kPa}$
Unanchored 2mm Dragon Skin	Anchored 2mm Dragon Skin
$1.36 \pm 0.046 \text{ pm/kPa}$	$3.47 \pm 0.061 \text{ pm/kPa}$

From Table 4.4 it is clear the sensitivity increased for all sensors when epoxy anchors were added onto the sensor. The 1mm SYLGARD sensitivity approximately doubled. This means that by adding anchors onto the 1mm SYLGARD sensor, the sensitivity increased to around the same amount as it did when the sensor was made with a thickness of 2mm without anchors (as described in Chapter 4.4.2.7). The 1mm Dragon Skin sensitivity increased by 0.63pm/kPa by adding anchors. This increase is larger than the increase experienced when the 1mm Dragon Skin sensor was turned into a 2mm Dragon Skin sensor (which was an increase of approximately 0.37pm/kPa). This shows that by adding anchors onto the 1mm sensors, the sensitivity can be increased further than it did when 2mm versions of the sensors were created.

The 2mm SYLGARD sensitivity increased by more than double (an increase of approximately 1.97pm/kPa) when anchors were placed on it. When comparing the overall optimisation of the SYLGARD sensor from an 'unanchored' 1mm sensor to an anchored 2mm sensor, the sensitivity increased by approximately 2.85pm/kPa (a 348% increase in sensitivity). The anchored 2mm Dragon Skin sensitivity increased by more than double its 'unanchored' version (an increase of approximately 2.11pm/kPa) and shows the biggest increase in sensitivity across all sensors caused by the addition of anchors. When calculating the overall optimisation of the Dragon Skin sensor from an 'unanchored' 1mm sensor to an anchored 2mm sensor, the sensitivity increased by approximately 2.48pm/kPa (a 250% increase in sensitivity).

4.5.7 Conclusion

Epoxy anchors were placed on the embedded sensors to create slight resistance allowing for increased induced axial strain along the FBG as the anchors pull the fibre when the patch expands into them. Four sensors were created with these added anchors, the 1mm SYLGARD and Dragon Skin sensors as well as the 2mm SYLGARD and Dragon Skin sensors. The sensors were then tested using the scale and loading arm setup so that the results would be comparable to the 'unanchored' sensors. Results showed a strong linear response from all the sensors. Results also showed that by adding anchors onto the sensors, the sensitivity increased for all the sensors. The anchored 2mm SYLGARD sensor achieved a sensitivity of $3.67 \pm 0.042\text{pm/kPa}$ with a low hysteresis error of $0.52 \pm 0.49\text{pm}$, and the anchored 2mm Dragon Skin sensor achieved

a sensitivity of $3.47 \pm 0.061 \text{ pm/kPa}$ with a hysteresis error of $1.4 \pm 1.3 \text{ pm}$. These sensitivities are now close to the ideal target sensitivity of 4 pm/kPa and provide a suitable and appropriate resolution for tracheal contact pressure sensing.

In addition to this, the CV for three of the sensors also decreased. For the 1mm SYLGARD sensor, it decreased from 6.22% to 0.62% by adding the anchors. For the 1mm Dragon Skin sensor, CV decreased from 5.54% to 3.26% by adding anchors. For the 2mm Dragon Skin sensor, the CV decreased from 3.4% to 1.75% by adding anchors. However, for the 2mm SYLGARD sensor, the CV increased from 0.9% to 1.13%. All these anchored sensors show good repeatability. It is worth reiterating that for these CV values the sample size is small and for a more accurate CV, more repeat experiments need to be carried out. It is also worth mentioning that the unanchored repeat experiments were carried out over a period of a few weeks whereas the anchored experiments were carried out over a period of a few days. This could affect the CV values and could possibly be a reason for the unanchored CV values being higher than the anchored CV values. Although all sensors had cured before experiments were carried out, it is mentioned in the Product Information Sheet of SYLGARD 184 that the material continues to cure over a period of 7 days.

Further experiments were carried out to test the repeatability of each of the sensors. To do this, each sensor underwent cyclic loading with a total of three loading and unloading cycles. The sensitivity of each cycle was calculated, and the CV was calculated. Results showed that all four sensors had very little variation with a CV of 0.75% for the anchored 1mm SYLGARD sensor, 0.59% for the anchored 1mm Dragon Skin sensor, 0.81% for the anchored 2mm SYLGARD sensor, and 0.16% for the anchored 2mm Dragon Skin sensor. When looking at the hysteresis, results showed that all four sensors had a strongly linear response with low hysteresis. For the anchored 1mm SYLGARD sensor, the hysteresis observed was the lowest of the four sensors ($0.39 \pm 0.25 \text{ pm}$). As it is possible that the hysteresis observed is due to the non-linearity of the scale, the performance of all four sensors showed great potential for performing linearly for low pressures with little to no hysteresis. However, more experiments need to be carried out over time to test sensor reliability.

4.6 Conclusion

In conclusion, this chapter has explored the development, and optimisation of sensors fabricated using the moulds created by BlueFrog Design. These sensors have shown promising results, both in terms of their repeatability and sensitivity. The BlueFrog mould allows for the creation of sensors in a repeatable fashion, free from defects like bubbles. Two distinct polymers, SYLGARD and Dragon Skin, were employed, and the resulting sensors showed uniform surfaces without the need for further refining; allowing sensors to be used straight out the mould. Initial experiments showed that both the SYLGARD and Dragon Skin sensors exhibited linear responses, with low hysteresis error ($0.26 \pm 0.34\mu\text{m}$ and $0.4 \pm 0.1\mu\text{m}$ respectively). Additionally, the sensitivity of the sensors were $0.82 \pm 0.051\mu\text{m/kPa}$ for SYLGARD and $0.99 \pm 0.055\mu\text{m/kPa}$ Dragon Skin which were far below the target sensitivity of $4\mu\text{m/kPa}$. Results showed that the SYLGARD sensor had a CV of 6.22% and the Dragon Skin sensor had a CV of 5.54%. These suggest moderate repeatability, with the potential for further optimisation.

The integration of the FEA model yielded valuable insights into sensor optimisation. It was demonstrated that altering the thickness of the sensor patch significantly influenced sensitivity, with thicker patches resulting in higher sensitivity. Notably, the model predicted that for the SYLGARD sensor, increasing the thickness to 2mm would approximately double its sensitivity. Additionally, simulations indicated that materials with Poisson's Ratios close to 0.5 exhibited higher sensitivity. The model also highlighted the impact of friction at the fibre-patch interface indicating that addressing this constraint, where friction restricts material deformation, is crucial for further optimisation of sensor sensitivity.

To test the effect thickness had on sensitivity, 2mm thick sensors were created from both polymers using a new mould from BlueFrog Design. The 2mm thick sensors were tested under the same experimental setup as the 1mm sensors. Results showed that the sensitivity of both sensors increased. In addition to this, the SYLGARD sensitivity doubled in line with FEA predictions.

To address the friction constraints on the patch expansion in the fibre-patch interface epoxy anchors were incorporated onto either side of the sensors. These anchors would pull on the fibre when the patch pushes on the anchors during expansion increasing the strain along the fibre. Results showed an increase in sensitivity for all the sensors. With

the anchored 2mm sensors in particular approaching the 4pm/kPa target sensitivity. The anchored 2mm SYLGARD sensor achieved a sensitivity of 3.67 ± 0.042 pm/kPa with a low hysteresis error of 0.52 ± 0.49 pm, and the anchored 2mm Dragon Skin sensor achieved a sensitivity of 3.47 ± 0.061 pm/kPa with a hysteresis error of 1.4 ± 1.3 pm. Furthermore, the CV values improved indicating good repeatability for all sensors.

Additional cyclic loading experiments for the four anchored sensors reinforced the robustness of these sensors, as they displayed minimal variation in sensitivity and hysteresis, with CV values all below 1% and as low as 0.16%. These results indicate that the sensors hold great promise for applications involving low pressures, exhibiting linear behaviour with low hysteresis.

The addition of anchors to the sensors clearly shows an improvement in both sensitivity and repeatability, however, it would be interesting to investigate if anchors could be optimised for even better sensitivity and repeatability results. It may be worth modelling the anchors to determine if the shape, size, or material of the anchors affect sensitivity results.

Furthermore, it is essential to note that further experiments are warranted to assess the long-term reliability of these sensors, as well as to expand the sample size for more accurate CV calculations. Moreover, the continuing curing process of the polymer materials over the course of a month may impact sensor sensitivity over time.

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Chapter 5 – Embedded Sensor Tube Testing

5.1 Introduction

After optimizing the embedded sensors for enhanced sensitivity and repeatability, in alignment with the target specifications for a tracheal contact pressure sensor in Chapter 4.4, the focus now shifts towards testing these optimized anchored sensors for tracheal applications. In Chapter 4, the calibration method employed (the loading arm and scale setup) involved calibrating the sensors on a hard, flat surface. The trachea however is a soft and curved surface. It cannot be assumed that as these sensors perform well on a hard and flat surface, they will also perform well on a soft and curved surface like the trachea. To bridge this gap in sensing environments, this chapter discusses a novel calibration setup featuring a rigid and curved surface, a load cell, and an ETT. This new setup simulates the pressures experienced in the trachea as well as the method of pressure application (inflation of the ETT cuff). This would allow for the identification of any challenges that the sensor may face when placed in the trachea and determine its ability to accurately measure pressures on the tracheal wall. If the sensor performs well in this setup, this setup can then be used to calibrate the sensors for tracheal use.

5.2 Calibration Tube

5.2.1 Calibration Tube Design

To create the calibration tube, a tube made from polypropylene with an inner diameter of 18mm and a load cell (outer diameter- 9.5mm, button diameter- 3mm, capacity- 1000g, rated output- 1mV/V, LCM100, Futek) were utilised. The 18mm diameter tube was chosen as it had a diameter comparable to an average adult trachea. Adult tracheas have an inner diameter range of 15mm and 25mm in males and 10mm to 21mm in females (Breatnach *et al.*, 1984). This diameter is important as it allows the curvature in the calibration tube to be like that of the trachea. This means that when the ETT cuff is placed in the calibration tube, it can simulate pressure values experienced in the trachea.

For the setup, the load cell is enclosed in 3D printed casing to keep securely in place. This also allowed for the load cell button to align with the calibration tube wall. A 3 mm hole was drilled into the side of the tube and the tube was secured onto the 3D printed casing with super glue. This allows the load cell button to protrude through the hole on the side of the tube and enabled direct measurement of the cuff contact pressure. Figure 5.1 shows the calibration tube design.

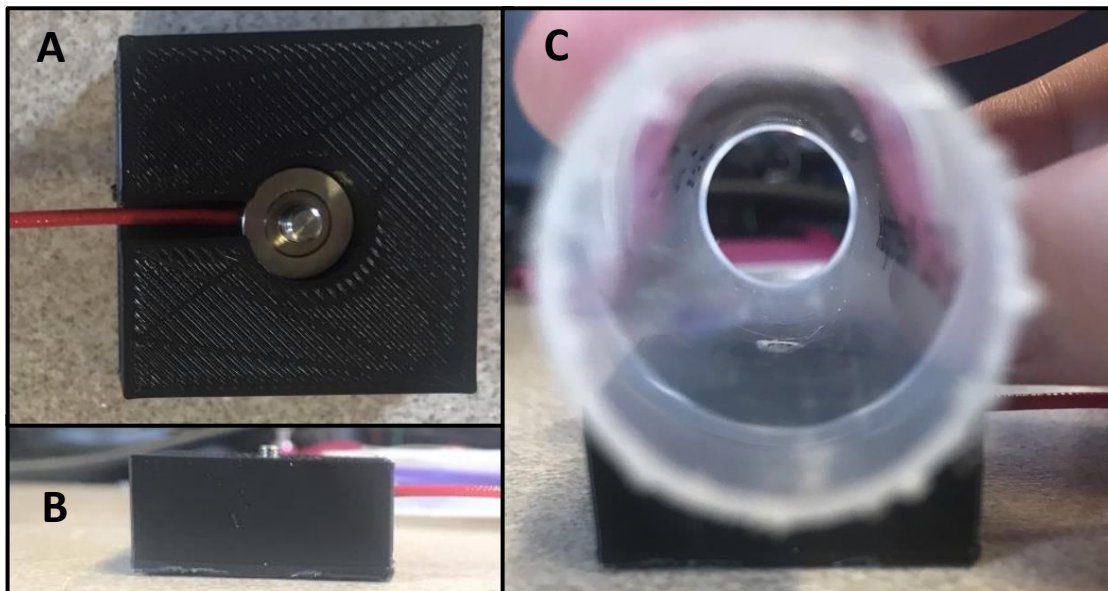


Figure 5.1: A: load cell secured inside 3D printed casing (4cm x 4cm x 2cm). B: side view of 3D printed casing with load cell inside, the load cell button is visibly sticking out of the casing. C: the 18mm inner diameter tube secured on top of the structure with super glue. The hole in the bottom of the tube is visible and the load cell just peeks through it so that it isn't protruding and creates a smooth surface in line with the tube walls.

5.2.2 Load Cell Circuitry

The load cell itself is equipped with a Wheatstone bridge arrangement, which generates both positive and negative voltage outputs in response to a load. This voltage differential, signifying the deviation from the load cell's equilibrium state, is the measurement signal.

The change in differential voltage is captured by an instrumentation amplifier (integrated circuit INA118P). The instrumentation amplifier enhances the measurement accuracy by amplifying the difference between the positive and negative voltage outputs from the load cell's Wheatstone bridge.

The output of the instrumentation amplifier is then sent to a low pass filter (LPF) using a $47\mu\text{F}$ capacitor, and a $15\text{k}\Omega$ resistor. Connected to the LPF is a Data Acquisition Card (DAQ) (USB-6210, National Instruments). This DAQ card incorporates an Analogue to Digital Converter (ADC), which effectively converts the analogue voltage signal into a digital format.

The DAQ, which serves as the interface between the amplified load cell output and the PC, connects via USB, enabling real-time monitoring of acquired data. Figure 5.2 shows a block diagram of the electrical components.

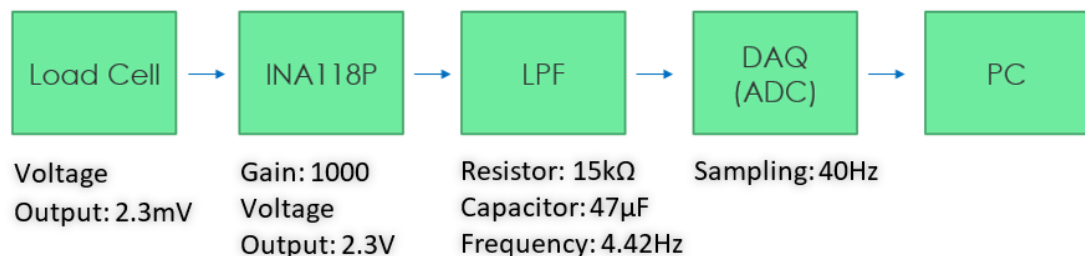


Figure 5.2: block diagram of the electrical components and values used to process the signal from the load cell (LCM100) and display the signal on the PC.

Figure 5.3 shows the load cell circuit with all the components.

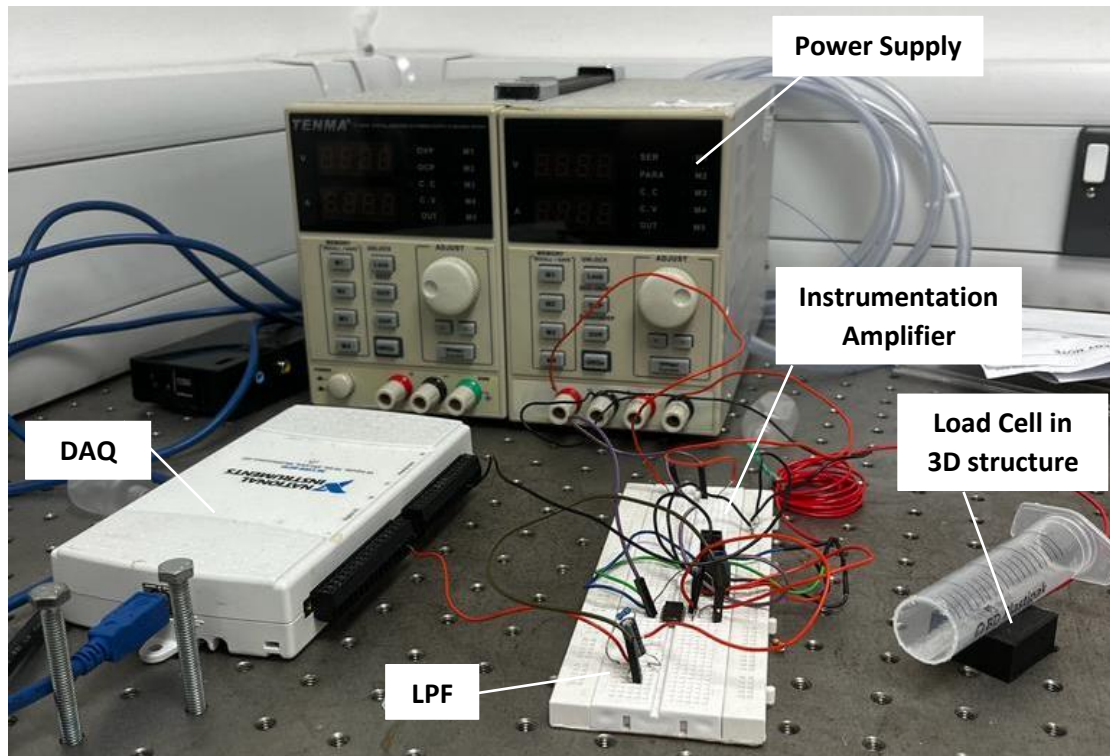


Figure 5.3: Load cell circuit and components.

To test the signal from the load cell a quick experiment was carried out using the calibration tube.

5.2.2.1 Experimental Methods

The calibration tube, load cell circuit, and an ETT were utilised for this experiment. The ETT cuff was placed in the calibration tube so that it was on top of the load cell. A syringe pump (AL-1000, World Precision Instruments) was connected to the ETT cuff and a manometer (HD750, Extech) using a T-shape connector. The manometer was used in this experiment to confirm that the intracuff pressures were low (up to 10kPa). This was to determine how the load cell responds to low pressures in this range. Figure 5.4 shows the experimental setup.

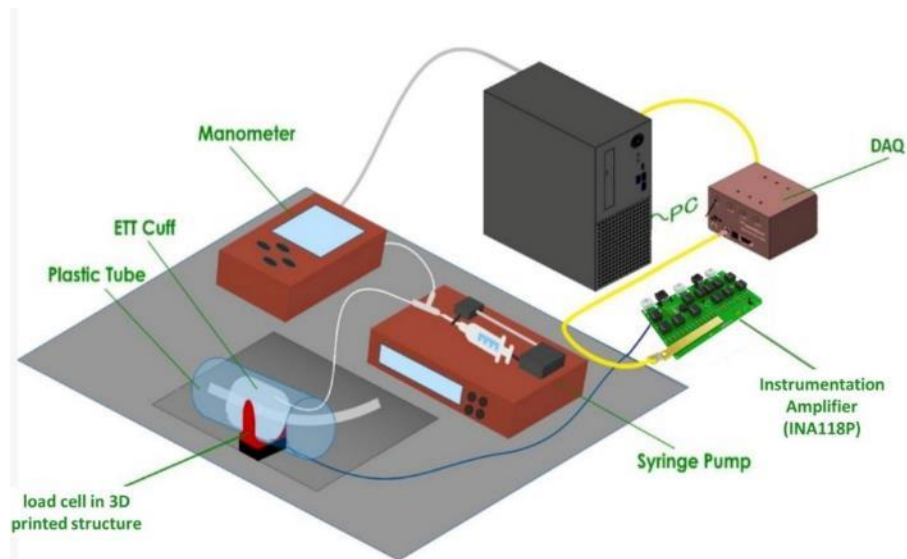


Figure 5.4: Experimental setup for load cell testing.

For this experiment, the syringe pump inflated the cuff in 2ml increments until a total of 8ml was inside the cuff. Each 2ml increment was held until the manometer stabilised. Once 8ml was reached, the air was removed from the cuff in the same 2ml decrements. The manometer response was recorded using the accompanying Extech software, and the load cell response was recorded using LabView. It is worth mentioning, that to prevent aliasing of the signal, the ADC was set to sample at a frequency of 40Hz. As the output voltage from the load cell was just 2.3mV, The INA118P provided a gain of 1000 to amplify the signal to 2.3V.

5.2.2.2 Experimental Results

Figure 5.5 shows the LCM100 load cell response.

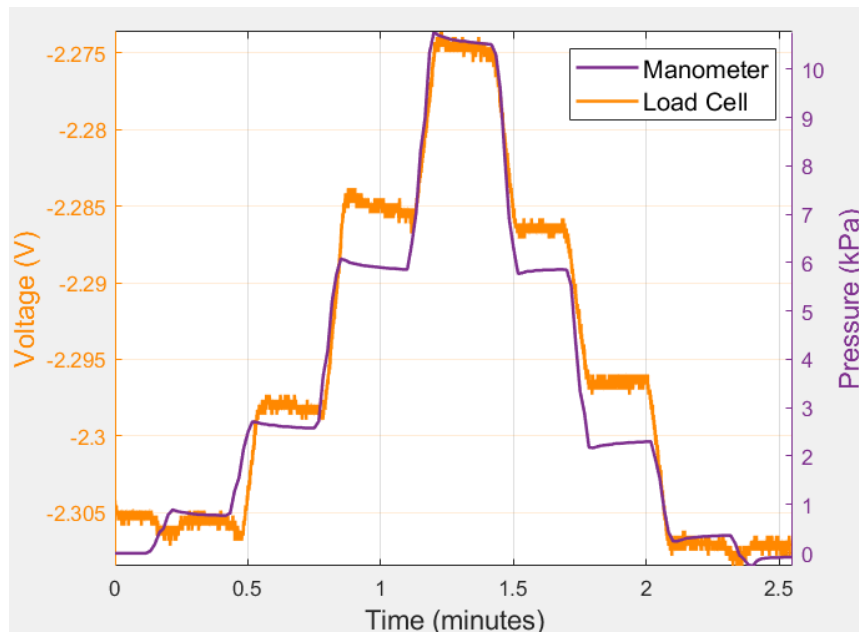


Figure 5.5: LCM100 load cell response.

From Figure 5.5, it is clear that the new LCM100 load cell is sensitive to these loads and experiences little noise.

A Fast Fourier Transform (FFT) of the signal was used to investigate whether any other noise sources are present. As the sampling rate is 40Hz, an FFT at 40Hz will not show if there is 50Hz noise present. To make sure there is no 50Hz noise in the signal, an FFT was done on the load cell signal at an ADC sampling rate of 150Hz. Figure 5.6 shows the FFT at 150Hz.

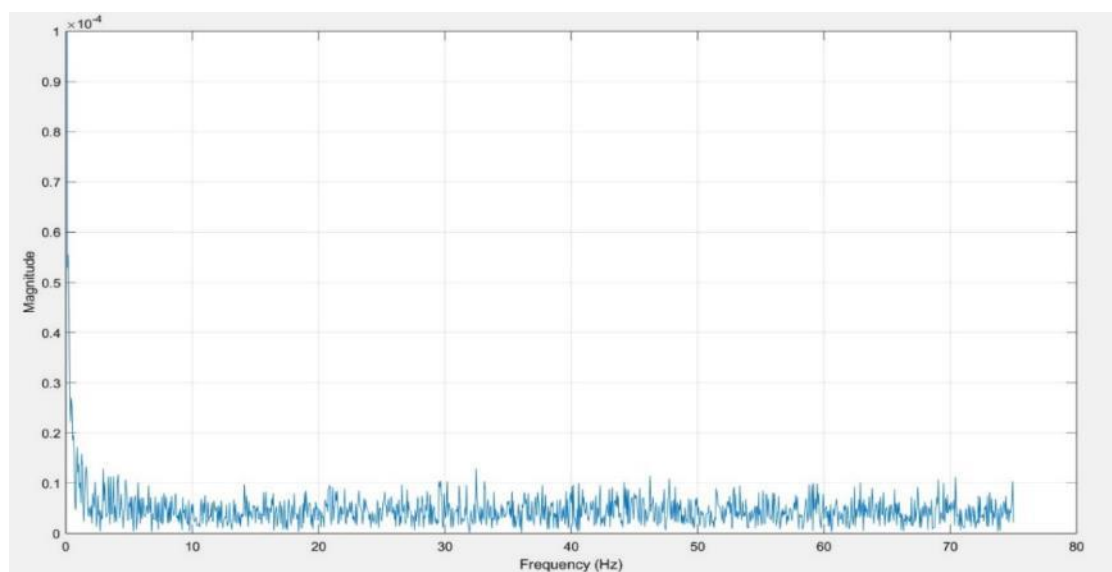


Figure 5.6: FFT of LCM100 load cell signal at 150Hz sampling rate showing no peak at 50Hz.

Figure 5.6 shows that there is no dominant 50Hz noise peak present in the signal. This indicates that the load cell noise has been minimised and is working well.

5.2.3 Load Cell Calibration

Before the sensors are calibrated in the calibration tube, the load cell needs calibrating. This allows for the conversion of millivolts to grams to use for pressure calculations.

5.2.3.1 Methods

To calibrate the load cell, it was placed on the scale and loading arm setup described in Chapter 3.2.2.2. The loading arm moved onto the load cell button until 15g was recorded on the scales. As the diameter of the load cell button is 3mm, the area is $7.07 \times 10^{-6} \text{m}^2$.

5.2.3.2 Results

Figure 5.7 shows the response from both the load cell and the scales.

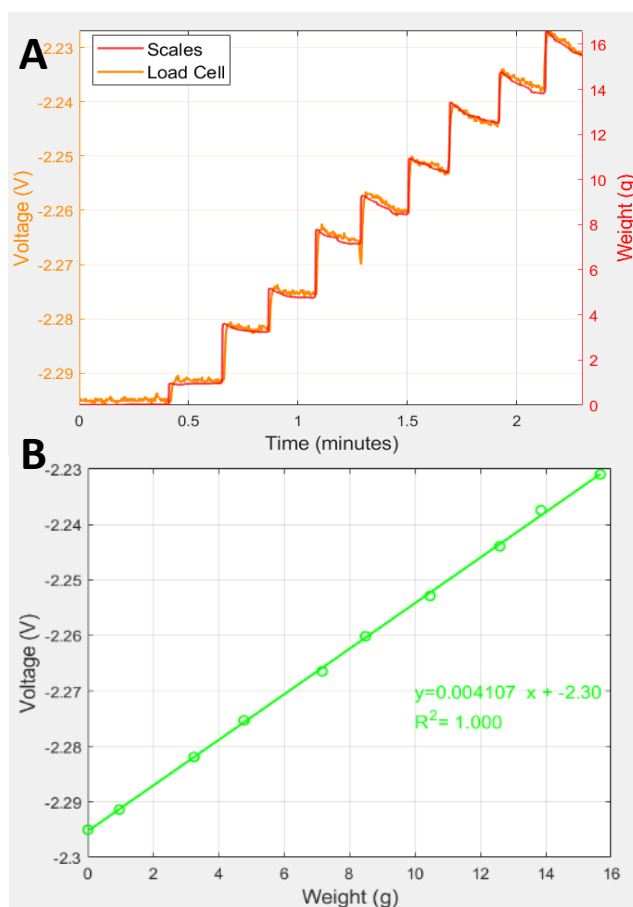


Figure 5.7: Load cell calibration, A: voltage from load cell against weight from scales. B: calibration curve.

From Figure 5.7b, the load cell sensitivity is 4.1mV per gram.

5.2.4 Load Cell vs Manometer

To make sure the load cell is performing well in the calibration tube, an experiment was carried out to compare it to the manometer. It is hypothesised that as the calibration tube is a uniform cylinder with no crevices, the pressure calculated by the load cell should be similar to the intracuff pressure readings on the manometer.

5.2.4.1 Methods

For this experiment, the setup described in Chapter 5.2.2.1 was utilised.

It is worth noting that the position of the load cell button is important as slight changes in the height can result in different pressures. If the load cell protrudes into the tube too much the pressure readings on the load cell will be greater than those of the manometer. This is because it disrupts the even distribution of pressure within the tube, creating a localised pressure point at the button. If the load cell is below the tube surface the pressure readings on the load cell will be lower than those of the manometer. The best position for the load cell button is flush with the tube surface (as shown in Figure 5.8) so that the results are most like that of the manometer. This can be checked with a ruler. When placing a ruler into the tube there should be a reading on the load cell. If there is no reading shown, then the load cell is below the surface of the tube and if the ruler cannot move around the tube freely then the button has protruded into the tube too much.

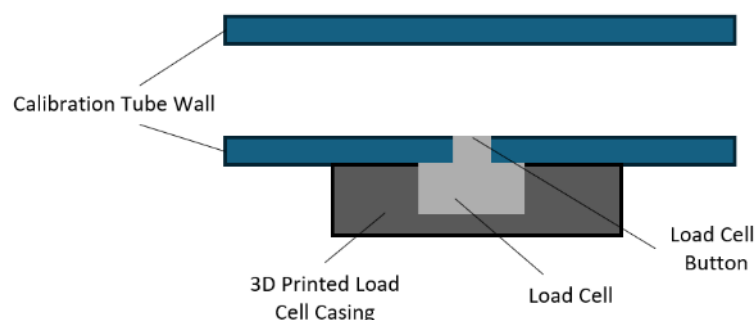


Figure 5.8: Diagram of how the load cell is incorporated in calibration tube setup. Note the load cell button flush with the tube wall.

The experimental procedure involved placing the ETT cuff in the calibration tube so that it is on top of the load cell button and inflating the cuff with the syringe pump in 2ml increments until the total amount of air in the cuff was 10ml. Then 2ml of air was

removed from the cuff in the same increments until all the air was removed from the cuff.

To calculate pressure from the load cell readings, the load cell calibration curve (Figure 5.7b) was used. The change in voltage from the beginning of the experiment is calculated (voltage change from baseline) and each 4.1mV change equates to a load of 1g on the load cell. The mass is then converted to Newtons by using Equation 3.2. Equation 3.1 is then used to calculate the pressure on the load cell, where the area is the area of the load cell button. The load cell has a button diameter of 3mm. This means the calculated area of the load cell button is $7.07 \times 10^{-6} \text{m}^2$.

5.2.4.2 Results

Figure 5.9 shows the load cell response compared to the manometer response

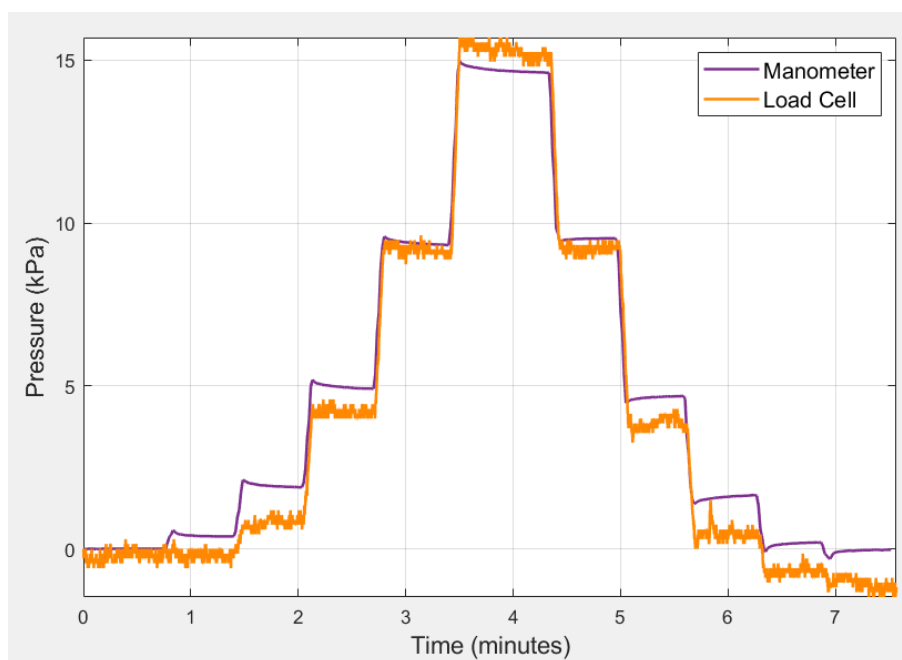


Figure 5.9: Load cell pressure readings vs manometer pressure readings in calibration tube.

From Figure 5.9, it is clear that the load cell follows the manometer response well for pressures above 5kPa. When looking at the first 2ml increment, the load cell does not sense the cuff. This is most likely due to the cuff not being inflated enough to be in contact with the wall (or the load cell button). At the end of the experiment, the load cell experiences a zero drift as the pressure at the end goes below 0Pa. Zero drift can be caused by temperature variations, hysteresis or vibrations during the experiments.

It seems from this experiment that the load cell and manometer readings are similar but not exact; and for pressures below 5kPa, the pressures that the load cell experiences are slightly lower than those inside the cuff.

5.2.5 Conclusion

A new method of calibrating the embedded sensors was created involving a calibration tube and a load cell that is sensitive to small loads of up to 1000g. This method was created to evaluate how the sensors would perform in an environment with a curved surface rather than a flat surface. The tube calibration setup was tested to check its performance against the manometer. Results show that the setup is working well and that the load cell follows the manometer closely for pressures above 5kPa. Results also show that for the first pressure increment, the load cell does not register pressure whereas the manometer does, showing that the cuff is inflating and not yet inflated enough to come in contact with the calibration tube wall.

5.3 Embedded Sensor Tube Tests

5.3.1 Introduction

To gain insights into the sensors' performance in a trachea-like setting, the optimised embedded sensors need to be tested in the calibration tube setup. This setup simulates the pressures applied to the tracheal wall during ETT cuff inflation, enabling the identification of any challenges the sensors may encounter in the tracheal environment.

5.3.2 Methods

The experiments with the calibration tube utilised the anchored embedded sensors described in Chapter 4.5.3 as they were most sensitive and exhibited a lower CV than the 'unanchored' sensors (as mentioned in Chapter 4.5.6).

This experiment was carried out using the anchored 1mm SYLGARD sensor and the anchored 1mm and 2mm Dragon Skin sensors. Unfortunately, the anchored 2mm SYLGARD sensor broke and therefore will not be tested in these experiments. If the results of these initial experiments prove promising, then the calibration tube setup will be used to calibrate all sensors including a newly created anchored 2mm SYLGARD sensor.

For these experiments the tube calibration setup in Chapter 5.2.2.1 was used with the anchored embedded sensors and optical interrogator added to the setup. Figure 5.10 shows the experimental setup.

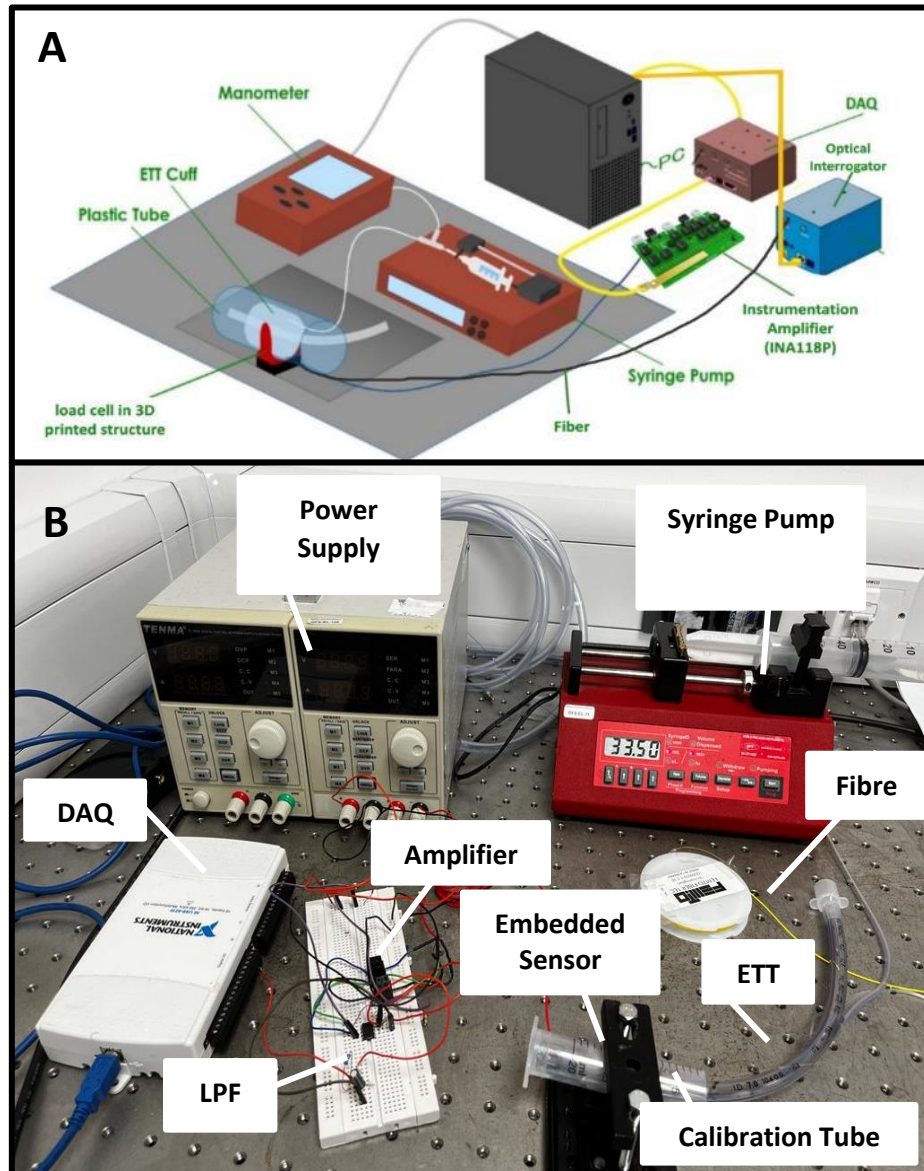


Figure 5.10: A: Diagram of the tube calibration experimental setup. B: Tube calibration experimental setup in lab.

To make sure the calibration tube does not move during the experiment, it was screwed in place (as shown in Figure 5.10 Calibration Tube).

For these experiments, the anchored embedded sensor was placed at the center of the ETT cuff, with another PVC cuff placed over it to create a double cuff. This method, which was also used in Correia *et al.*'s in-vivo experiments (Correia *et al.*, 2022), ensures the sensor is protected when placed in the trachea. Direct placement in the trachea without protection could pose a risk to the patient and make retrieval difficult if the sensor breaks. Therefore, using a double cuff in this experiment mimics the likely clinical application. The outer cuff, which had the same diameter as the ETT cuff, held

the sensor in place without needing to anchor it to the ETT tube. Figure 5.11 shows the double cuff and the sensor in between the two cuffs.

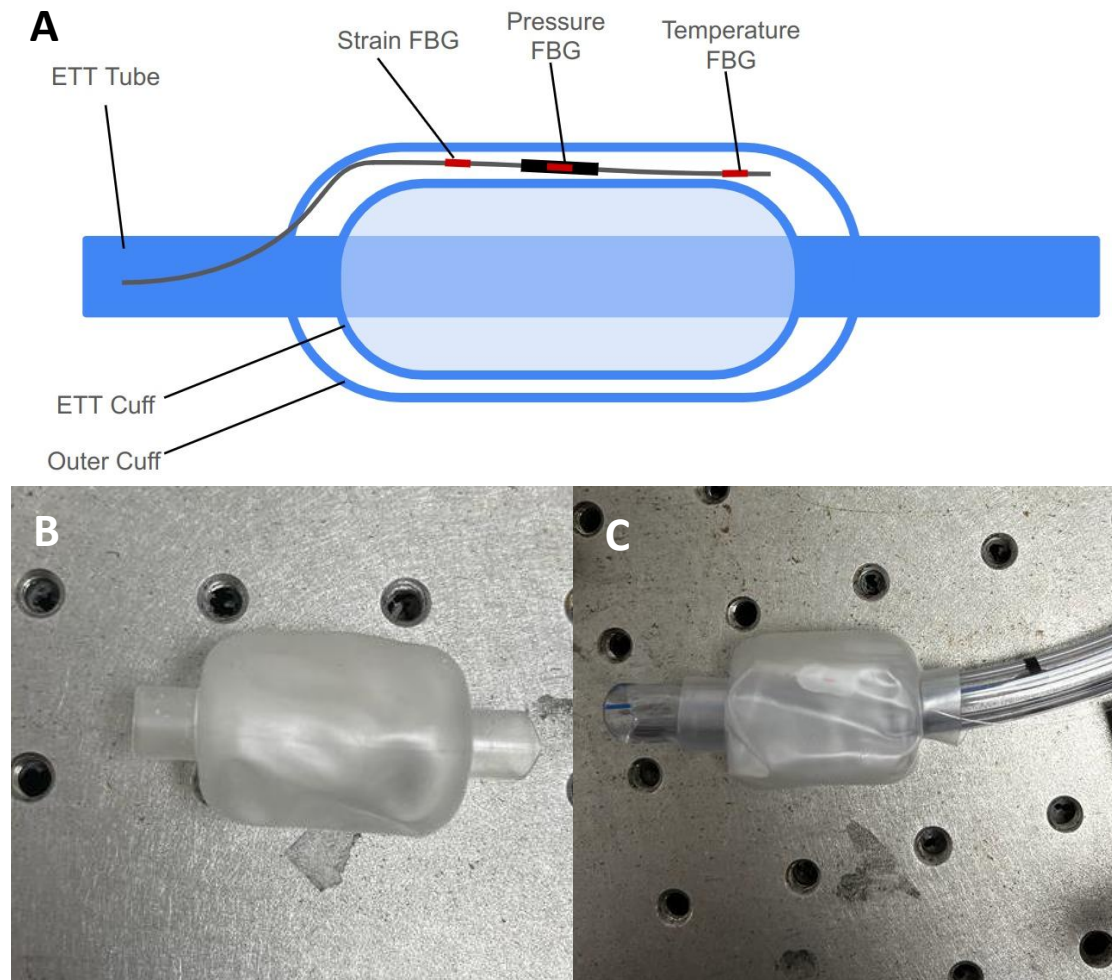


Figure 5.11: A: diagram of how the sensor is deployed on the ETT cuff. The sensor is free floating. B: the outer cuff used to place on top of the embedded sensor. C: the embedded sensor in between the double cuff.

Once the embedded sensor was placed on the ETT cuff with the double cuff over it, the deflated ETT cuff was positioned in the calibration tube on top of the load cell, ensuring the sensor aligned with the load cell. Using a syringe pump, the cuff was then inflated with 2 ml increments of air until a pressure of approximately 9 kPa was indicated on the manometer, reaching a total volume of around 8-10 ml of air. This pressure was achieved more quickly for the 2mm thick sensors compared to the 1mm sensors. The cuff was held at this pressure before being deflated in the same 2 ml increments. Each experiment was conducted three times, and the manometer pressure, load cell pressure, and FBG wavelength shift were recorded. Sensitivity was calculated using the pressure values from the load cell.

5.3.3 Results and Discussion

Experiments were carried out with the anchored 1mm SYLGARD and the 1mm and 2mm Dragon Skin sensors. Figure 5.12 shows the results for the first anchored 1mm SYLGARD sensor experiment.

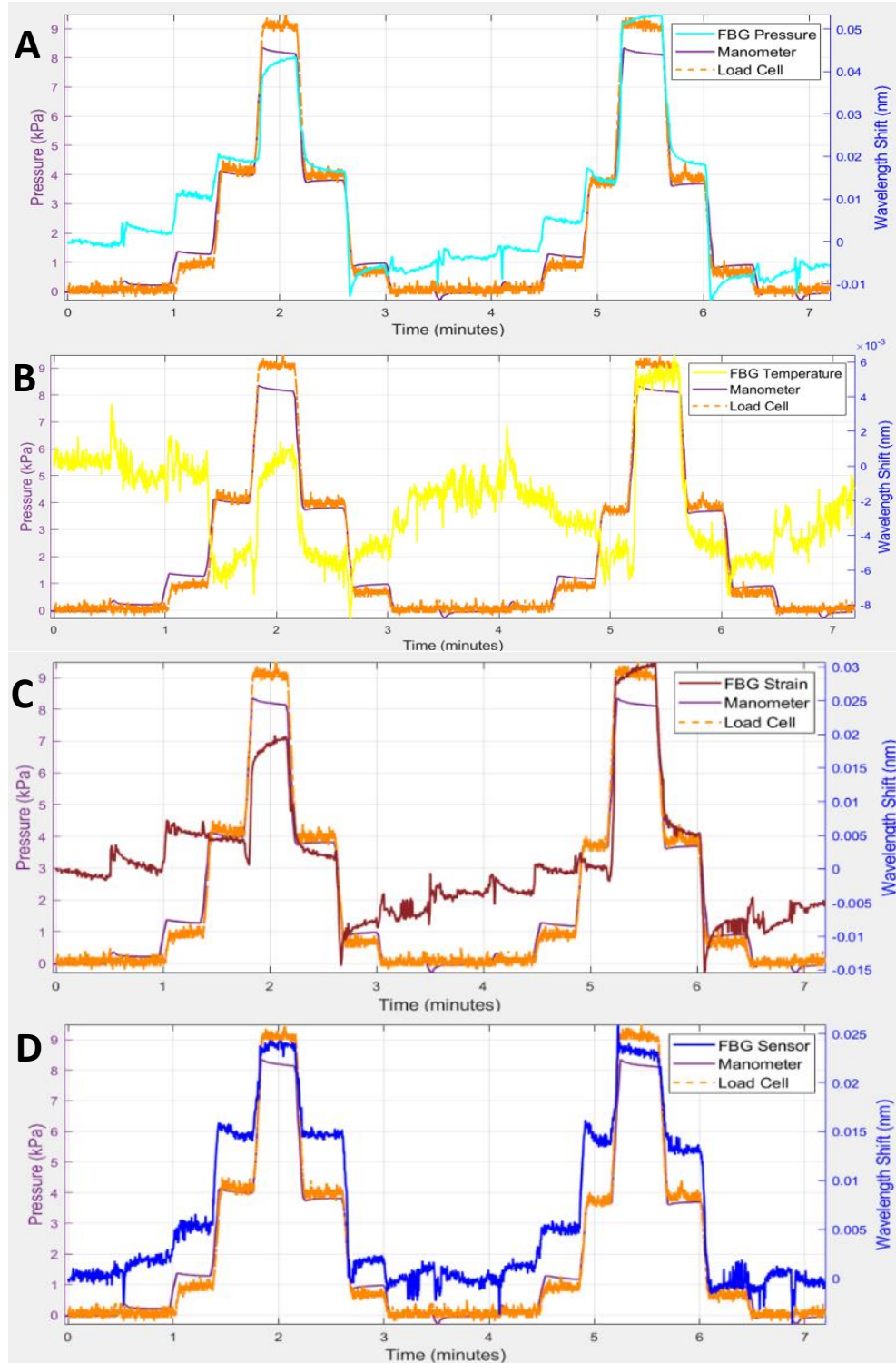


Figure 5.12: FBG response for anchored 1mm SYLGARD in calibration tube. A: raw response of pressure FBG (located inside anchored patch). B: raw response of temperature FBG. C: raw response of strain FBG. D: response after the strain FBG was compensated for (pressure minus strain).

Figure 5.12b illustrates strain observed in the FBG used for temperature compensation, likely caused by the cuff's expansion exerting tension on the fiber. Additionally, Figure 5.12c indicates a distinct response between the temperature FBG, positioned 30mm from the pressure FBG, and the strain FBG located 15mm away. This discrepancy suggests varying strain levels across different parts of the cuff. Given the strain experienced by the temperature FBG, it cannot effectively compensate for temperature changes affecting the strain FBG.

Since both reference FBGs are affected by temperature and strain, the contact pressure was determined by subtracting the values of the strain FBG (closest to the anchored embedded sensor, Figure 5.12c) from the raw pressure response (Figure 5.12a). This compensation process resulted in the creation of Figure 5.12d.

These observations (where the temperature FBG experiences strain) were seen throughout the repeat experiments and the magnitude of strain observed in each experiment differed. An example of this can be seen in Figure 5.12c where the first cycle shows a wavelength shift of 0.02nm for a pressure of 8kPa, and the next cycle shows a wavelength shift of 0.03nm for the same 8kPa pressure. Interestingly, this difference in strain (of the 2 cycles) can be seen in both the temperature response and the raw pressure response and is assumed to be due to sensor movement during inflation and deflation. When subtracting the strain response from the pressure response, these differences are removed and both cycles have the same magnitude as well as the same starting and ending wavelength.

Figure 5.13 shows the anchored 1mm SYLGARD sensor response for the second and third repeat experiments.

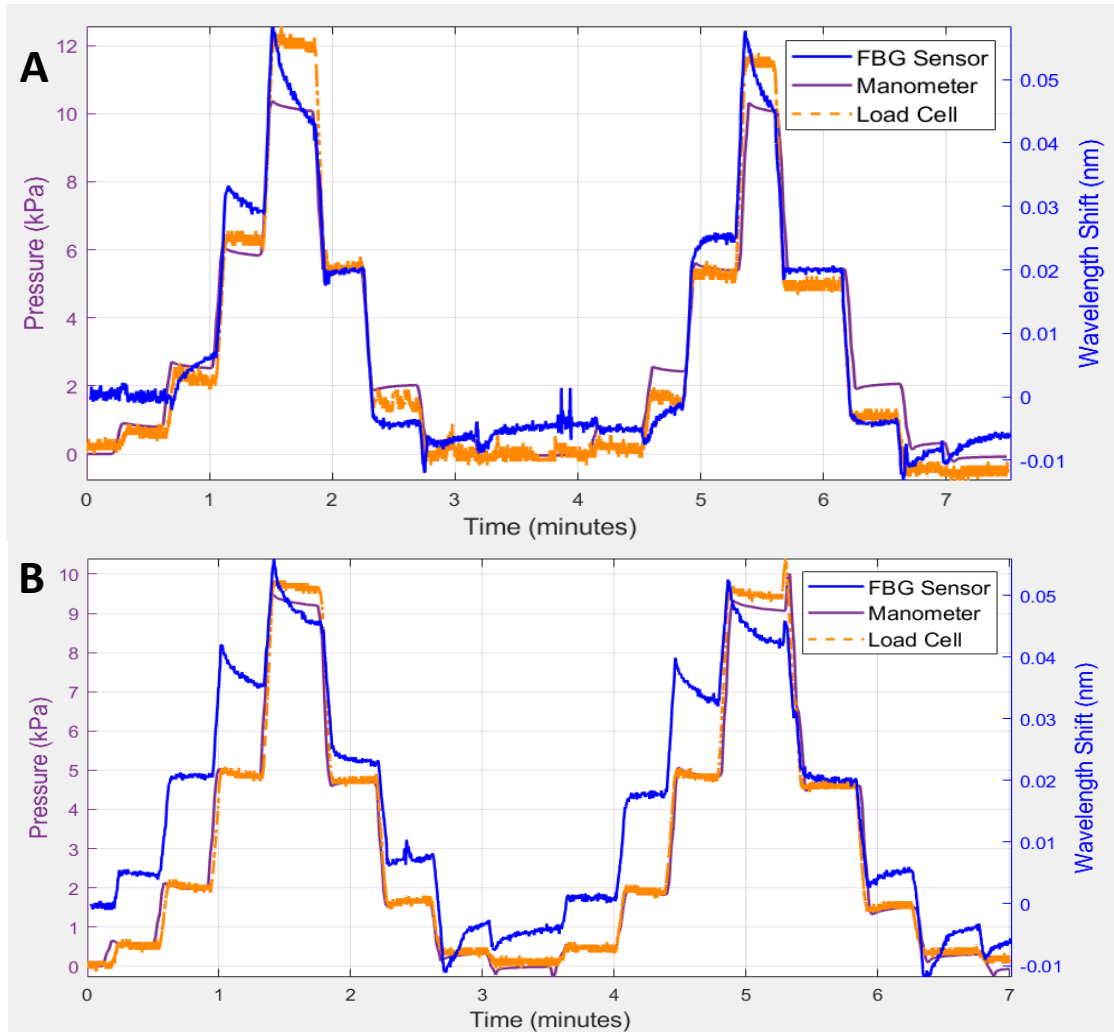


Figure 5.13: Pressure response after strain and temperature compensation using same method as used for Figure 5.12d. A: first anchored 1mm SYLGARD repeat experiment, B: second anchored 1mm SYLGARD repeat experiment.

When observing the response from the three experiments (Figures 5.12d, 5.13a, and 5.13b), it is clear the responses are different. Figure 5.14 shows the calculated sensitivity for both cycles of the first experiment. As the results were not linear, a piecewise linear plot was created.

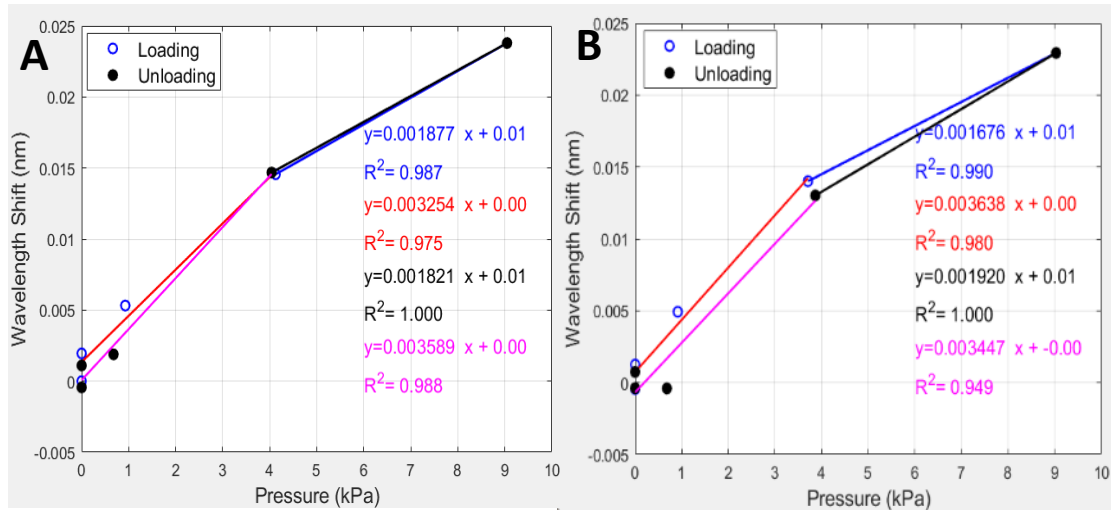


Figure 5.14: A: sensitivity for cycle 1 of the first anchored 1mm SYLGARD calibration tube experiment. B: sensitivity for cycle 2.

It's important to acknowledge that for pressures above 4kPa a line of best fit was drawn with only two data points and therefore this sensitivity may not capture the true behaviour of the sensor for these pressures. However, the primary goal is to assess the sensor's performance in the new environment and to assess how each loading and unloading cycle differs; not specifically to determine the sensitivity for calibration. Additionally, provided the results of these experiments are promising, more experiments with more data points will be taken to provide a more accurate sensitivity reading for calibration purposes.

Table 5.1 shows the calculated sensitivity for all three experiments.

Table 5.1: sensitivities of each of the three anchored 1mm SYLGARD experiments as well as the sensitivity of each loading and unloading cycle.

Experiment	Sensitivity of First Cycle (pm/kPa)	Sensitivity of Second Cycle (pm/kPa)
1	<4kPa: 3.42 >4kPa: 1.85	<4kPa: 3.54 >4kPa: 1.80
2	4.08	4.79
3	<2kPa: 8.82 >2kPa: 3.95	<2kPa: 10.393 >2kPa: 3.98

Table 5.1 shows that the sensitivity varies for each experiment and even every loading and unloading cycle. However, interestingly the sensitivity throughout the experiments show a much higher sensitivity than when these sensors were calibrated on a flat surface (in Chapter 4.5.5.1 the anchored 1mm SYLGARD showed a sensitivity of 1.61pm/kPa).

Figure 5.15 shows the results for the first anchored 1mm Dragon Skin embedded sensor experiment.

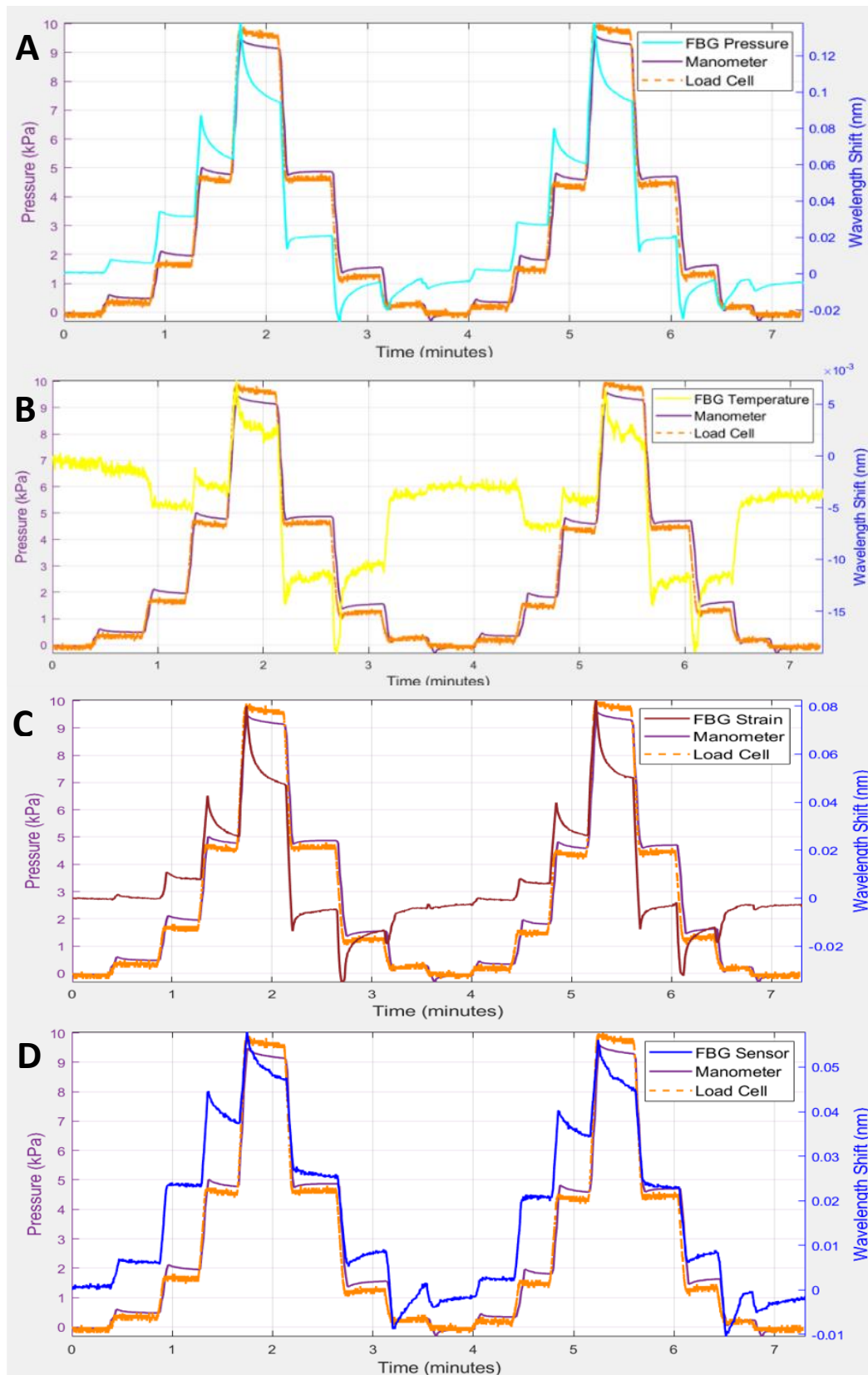


Figure 5.15: FBG response for anchored 1mm Dragon Skin in calibration tube. A: raw response of pressure FBG (located inside anchored patch). B: raw response of temperature FBG. C: raw response of strain FBG. D: response after the strain FBG was compensated for (pressure minus strain).

Like Figure 5.12b, Figure 5.15b shows that the FBG for temperature compensation experiences some strain and the response is different from the response of the strain FBG (Figure 5.15c). As both reference FBGs experience temperature and strain (like observed in Figure 5.12), to determine the contact pressure, the values of the strain FBG were subtracted from the raw pressure response (Figure 5.15a) and Figure 2.15d was created.

These observations (where the temperature FBG experiences strain) were seen throughout the repeat experiments and the magnitude of strain observed in each experiment differed. From Figure 5.15, it seems the sensor movement between cycles is less apparent. With both cycles of each FBG experiencing similar wavelength shifts for each change in pressure.

Figure 5.16 shows the anchored 1mm Dragon Skin sensor response for the second and third repeat experiments.

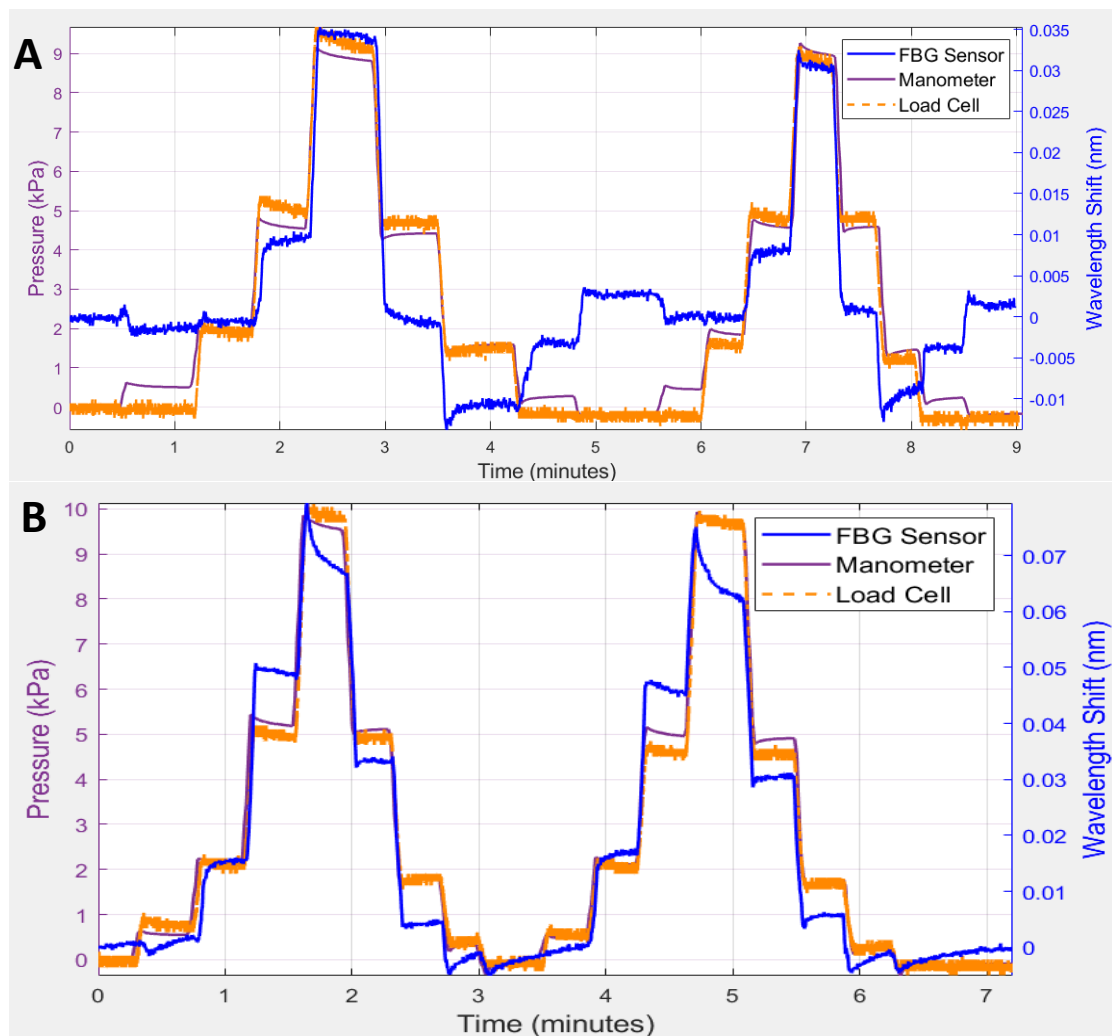


Figure 5.16: Pressure response after strain and temperature compensation. A: first anchored 1mm Dragon Skin repeat experiment, B: second anchored 1mm Dragon Skin repeat experiment.

Again, Figures 5.15d, 5.16a, and 5.16b appear different. As Figure 5.16a shows that for the FBG sensor some steps result in a negative pressure and some steps show an increase in pressure when the pressure is in fact decreasing, the sensitivity of this experiment was not calculated. It is clear from these experiments that the sensor in this environment is not repeatable and more needs to be done to improve sensor response and repeatability. Table 5.2 shows the sensitivities calculated for the other 2 experiments.

Table 5.2: sensitivities of the anchored 1mm Dragon Skin experiments as well as the sensitivity of each loading and unloading cycle.

Experiment	Sensitivity of First Cycle (pm/kPa)	Sensitivity of Second Cycle (pm/kPa)
1	<2kPa: 10.78 >2kPa: 3.83	<2kPa: 10.88 >2kPa: 3.59
3	7.33	6.86

Table 5.2 also shows different sensitivities throughout the experiments and loading and unloading cycles. Again, like with the anchored 1mm SYLGARD sensor, this sensor shows much higher sensitivities in the calibration tube than during calibration on a flat surface where the anchored 1mm Dragon Skin showed a sensitivity of around 1.62pm/kPa (Chapter 4.5.5.1).

Figure 5.17 shows the results for the first anchored 2mm Dragon Skin embedded sensor experiment.

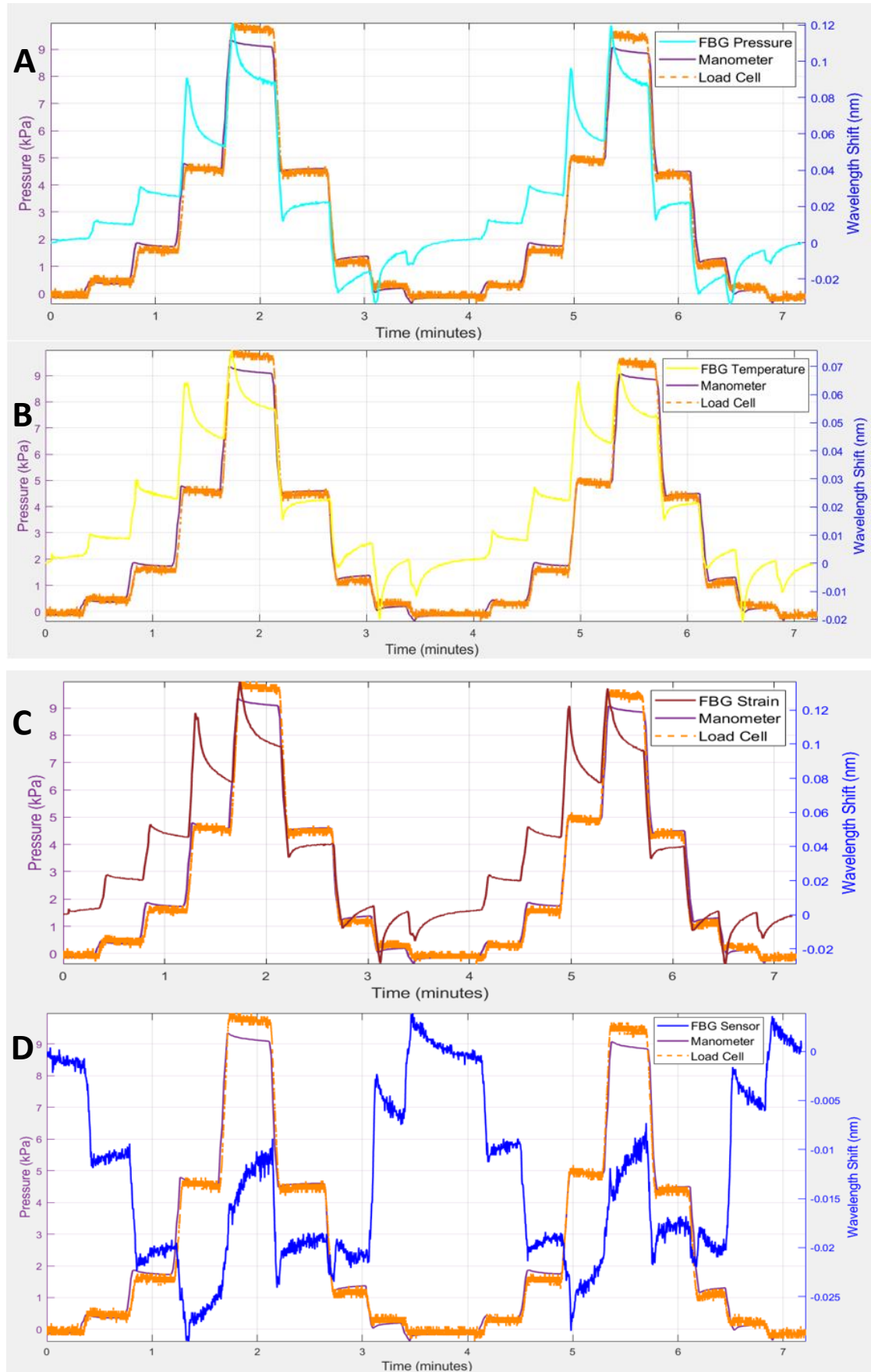


Figure 5.17: FBG response for anchored 2mm Dragon Skin in calibration tube. A: raw response of pressure FBG (located inside anchored patch). B: raw response of temperature FBG. C: raw response of strain FBG. D: response after the strain FBG was compensated for (pressure minus strain).

Figure 5.17 shows that all three FBGs exhibit similar pressure response patterns, albeit with varying magnitudes of wavelength shift. The pressure FBG (Figure 5.17a), temperature FBG (Figure 5.17b), and strain FBG (Figure 5.17c) show comparable trends in their responses to pressure changes. The strain FBG experiences a larger wavelength shift than the pressure FBG which indicates that the pressure sensor is mainly sensing strain. This is confirmed when subtracting the strain FBG response from the pressure FBG response (Figure 5.17d), the wavelength shift is in the negative direction indicating that the sensor may be bending along the curvature of the tube.

Moreover, a consistent observation across all experiments (Figures 5.12, 5.15, and 5.17) is that the strain FBG exhibits a significantly larger wavelength shift than the temperature FBG. This disparity is attributed to the positioning of the temperature FBG, which is situated 30mm away from the pressure FBG located at the centre of the ETT cuff. In contrast, the temperature FBG is positioned nearer to the end of the ETT cuff, where it experiences less tension during cuff inflation compared to the strain and pressure FBGs located closer to the ETT cuff's centre.

Figure 5.18 shows the anchored 2mm Dragon Skin sensor response for the second and third repeat experiments.

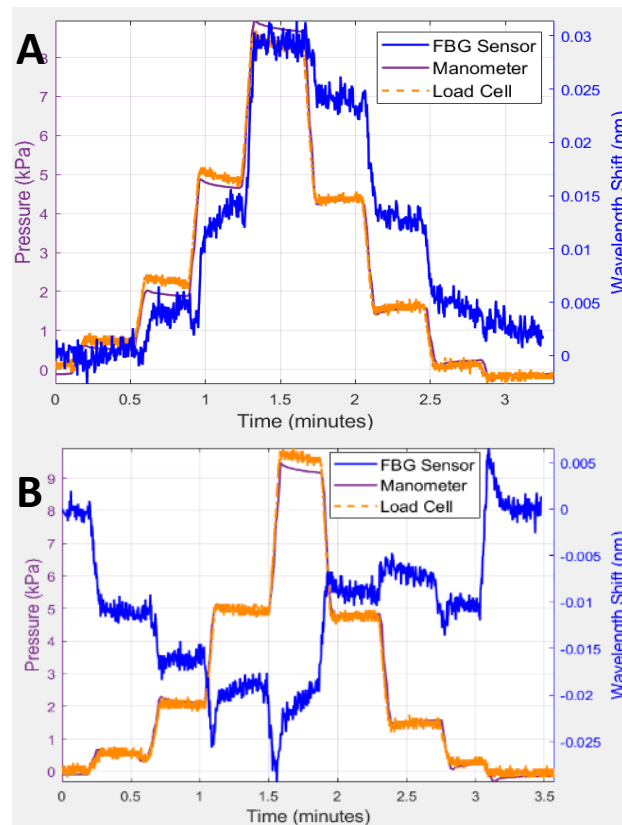


Figure 5.18: Pressure response after strain and temperature compensation. A: first anchored 2mm Dragon Skin repeat experiment, B: second anchored 2mm Dragon Skin repeat experiment.

Figures 5.17d and 5.18b show that the strain experienced on the strain FBG was greater than the contact pressure experienced on the sensor, causing a negative pressure response during these experiments. This is also seen in a few steps in Figure 5.16a. This is thought to be due to the sensor's orientation where the curvature of the calibration tube causes the sensor to bend during ETT cuff inflation. Figure 5.18a shows a positive pressure response with a sensitivity of 3.42pm/kPa. This sensitivity is similar to the sensitivity of the sensor during calibration in Chapter 4.5.5.1 (3.47pm/kPa).

The findings of these experiments are as follows:

- 1) The temperature FBG experienced strain making it difficult to distinguish the temperature contribution to the wavelength shift. In addition to this, the strain experienced on the strain FBG was different from that experienced on the temperature FBG. This may be due to the temperature FBG placement being more toward the edges of the ETT cuff rather than the centre.
- 2) The response from the anchored embedded sensors varies from experiment to experiment. For each experiment the sensitivity changes. This is most probably due to sensor orientation and the curvature of the calibration tube.
- 3) For the experiments that contains 2 cycles, each cycle contains discrepancies unlike in the scale experiments in chapter 4. This could be due to sensor movement within the double cuff in between inflation-deflation cycles.
- 4) The sensor design used in the manner of these experiments, cannot produce repeatable results and therefore cannot be used to measure wall contact pressure.

Methods for protecting the temperature FBG from strain caused by the cuff inflation should be considered. In addition to this, it may be worth attaching the sensor to the ETT cuff to prevent sensor movement during inflation-deflation cycles, however, this may affect patch expansion to pressures and could increase the strain experienced on the fibre.

While attaching the sensor to the cuff may eliminate sensor movement during inflation-deflation cycles, it may not prevent the variations in sensitivity observed due to sensor orientation and calibration tube wall curvature. It is possible that for a straight sensor placement (directly parallel with length of the calibration tube) when the sensor touches the wall it experiences minimal curvature as shown in Figure 5.19a. Whereas if the sensor moves slightly so that it is no longer parallel with tube wall and at a slight angle,

when it touches the wall, it will experience an increase of the wall's curvature. This effect would increase the more the sensor orientation moves away from the parallel direction and into a perpendicular position causing different strains on the fibre. This phenomenon is shown in Figure 5.19.

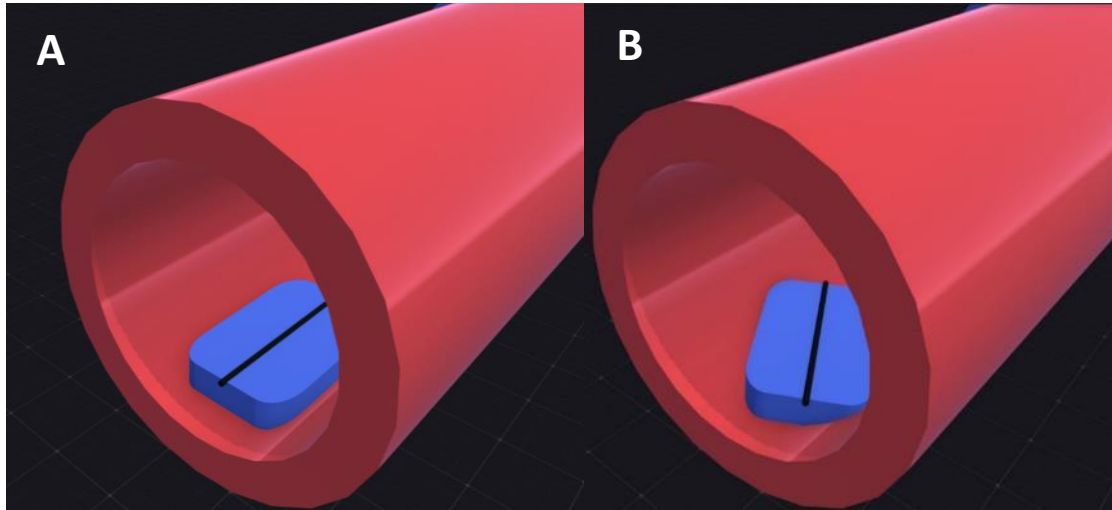


Figure 5.19: Illustration demonstrating how sensor orientation can cause bending of patch and FBG. Where the blue structure is the sensing patch, and the black line through it represents the fibre position (and FBG) through the patch. A: the sensor is placed in the calibration tube so that it is in parallel with the length of the tube. In this position the FBG is exposed to minimal curvature from the calibration tube. B: the sensor is at an angle and is no longer parallel to the length of the tube, in this orientation the sensor experiences more of calibration tube's curvature and therefore causes different strains on the fibre resulting in different sensor readings.

In theory, this phenomenon will occur no matter the shape of the sensor because the sensor will always bend when it meets the wall causing the FBG to bend too. If the sensor was instead a 4mm square or even a circle, the orientation of the sensor would still cause different strains on the fibre and therefore different responses (as shown in Figure 5.20), however the size of the differences may not be as obvious.

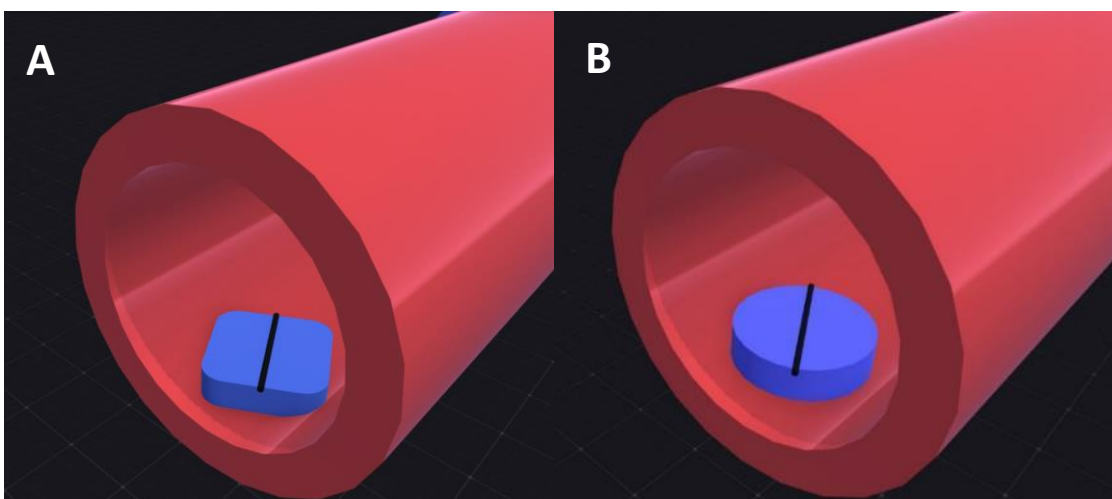


Figure 5.20: Illustration demonstrating how sensor orientation will be a challenge for any sensor shape.

Because the embedded sensors are 10mm long and 4mm wide, the curvature of the walls affect the response in a more profound way. The embedded sensor has a longer side (10mm) compared to a 4mm square, which means it provides a longer lever arm to apply force to the FBG. The longer lever arm allows for more bending under the same applied force. Therefore, the embedded sensors cause the FBG to bend to a greater extent.

The only method to avoid this phenomenon from coming into effect is to be certain that every time the embedded sensor is placed in the calibration tube, it is in the exact same orientation. This, however, is difficult. Even if the sensor is attached to the ETT cuff so that it is aligned to be placed in parallel with the calibration tube (as shown in Figure 5.19a), the ETT cuff itself could be placed in the calibration tube at an angle. This is shown in Figure 5.21.

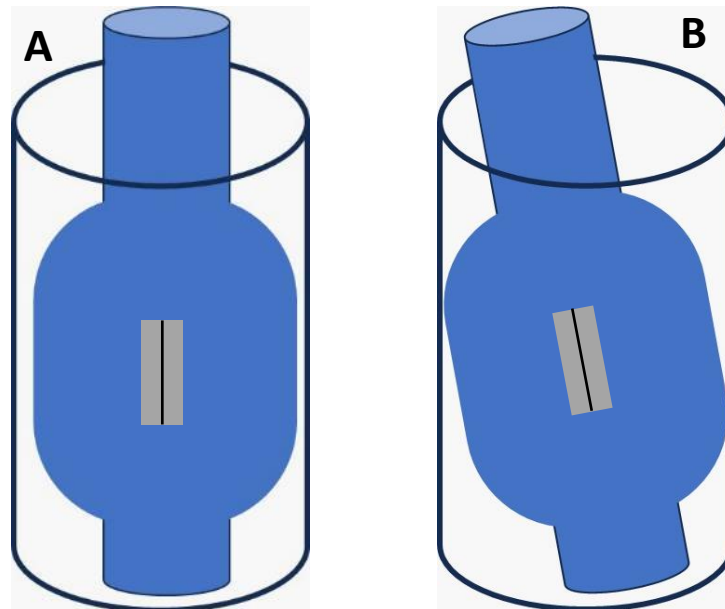


Figure 5.21: Illustration demonstrating how the ETT cuff positioning effects the sensor's orientation. The blue structure is the ETT and cuff, the grey structure is the embedded sensor, and the black line through the sensor represents the fibre (FBG) position.

In fact, positioning the ETT cuff so that it is in the same position in the in-vivo trachea every time cannot yet be achieved. The ideal sensor for in-vivo tracheal application should not be influenced by tube curvature.

It is important to note that these experiments were conducted on a smooth tube, while a trachea's wall is non-uniform. If the sensor's orientation matters in a smooth tube, it's reasonable to assume that both the sensor's orientation and location (whether the sensor is on smooth tissue or cartilage) will be relevant when testing in a trachea.

Although these experiments reduce the gap in environments (from the hard flat surface of weighing scales to the soft curved surface of the trachea), the calibration tube still differs significantly from a trachea. It is possible that these findings may be unique to the calibration tube as the trachea should expand slightly during cuff inflation. It is also possible that the curvatures of the trachea do not affect the sensor's response as much as the curvatures of the calibration tube.

Having discussed the challenges observed when the sensor is placed in the calibration tube, it is important to determine if the double cuff also has an effect on sensor response.

5.3.4 Conclusion

Anchored embedded sensors were placed in the calibration tube for testing. This would allow for the simulation of pressures applied on the tracheal wall when the ETT cuff is inflated. Results showed that strain compromised the temperature reference FBG and that both the strain and temperature FBGs experienced strain at different magnitudes. Results also showed the response of the sensor varied from experiment to experiment as well as for each loading and unloading cycle of the same experiment. This is likely due to factors such as sensor movement within the cuff, sensor orientation, and the calibration tube's curvature. Changes in sensor orientation could lead to the sensor experiencing increased curvature causing varying strain on the fibre, impacting sensor responses.

Despite these findings, it is worth noting that the calibration tube is still unlike a trachea and with the different parts of the trachea (cartilage and muscle), the sensor orientation and location are likely to be relevant factors if the embedded sensor is used in the trachea.

It is worth mentioning that although these experiments reduce the gap in environments the calibration tube is not a true trachea realistic phantom, and it is possible that these findings may be unique the embedded sensors' response in the calibration tube.

5.4 Effect of a Double Cuff on Sensor Readings During Inflation

5.4.1 Introduction

To assess the impact of the double cuff on FBG readings during tube experiments, an investigation was conducted to observe the sensor's response to double cuff inflation outside of the calibration tube. While the double cuff serves as a means of safeguarding the fibre and ensuring patient safety if the sensor were to break in the trachea, it may also inadvertently influence pressure readings.

5.4.2 Methods

For this experiment, the tube calibration setup from Figure 5.10 was utilised, except the load cell and calibration tube was removed and the ETT was held up using a clamp instead. The embedded sensor was placed in between the double cuff and was not attached to the cuff (like in the previous experiments). Figure 5.22 shows the experimental setup.

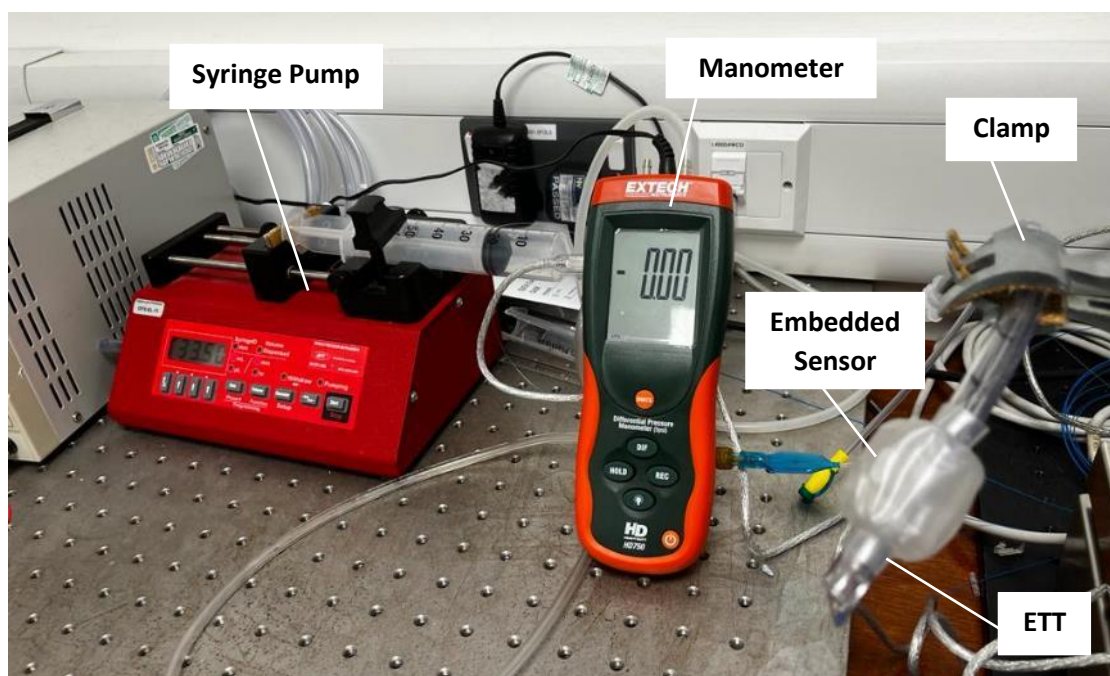


Figure 5.22: Experimental setup for ETT cuff inflation outside tube.

For the experimental procedure, the ETT cuff was inflated in the same 2ml increment until 10ml of air was placed in the cuff. This was so that it would be possible to

determine what effect this was having in the experiments. Three inflation-deflation cycles were conducted, using a manometer as a reference for the sensor response. The sensor was then repositioned to the left surface of the cuff, and the experiment was repeated. Finally, the sensor was placed on the bottom of the ETT cuff, and the experiment was repeated once more. This ensured the procedure was conducted on three different sides of the ETT cuff.

5.4.3 Results and Discussion

Figure 5.23 shows the results for the sensor on all three sides of the ETT cuff. It is worth noting that, like in the calibration tube experiments, the temperature FBG experienced strain. In Figure 5.23, the FBG response is the strain FBG response removed from the raw pressure response (like in the other calibration tube experiments).

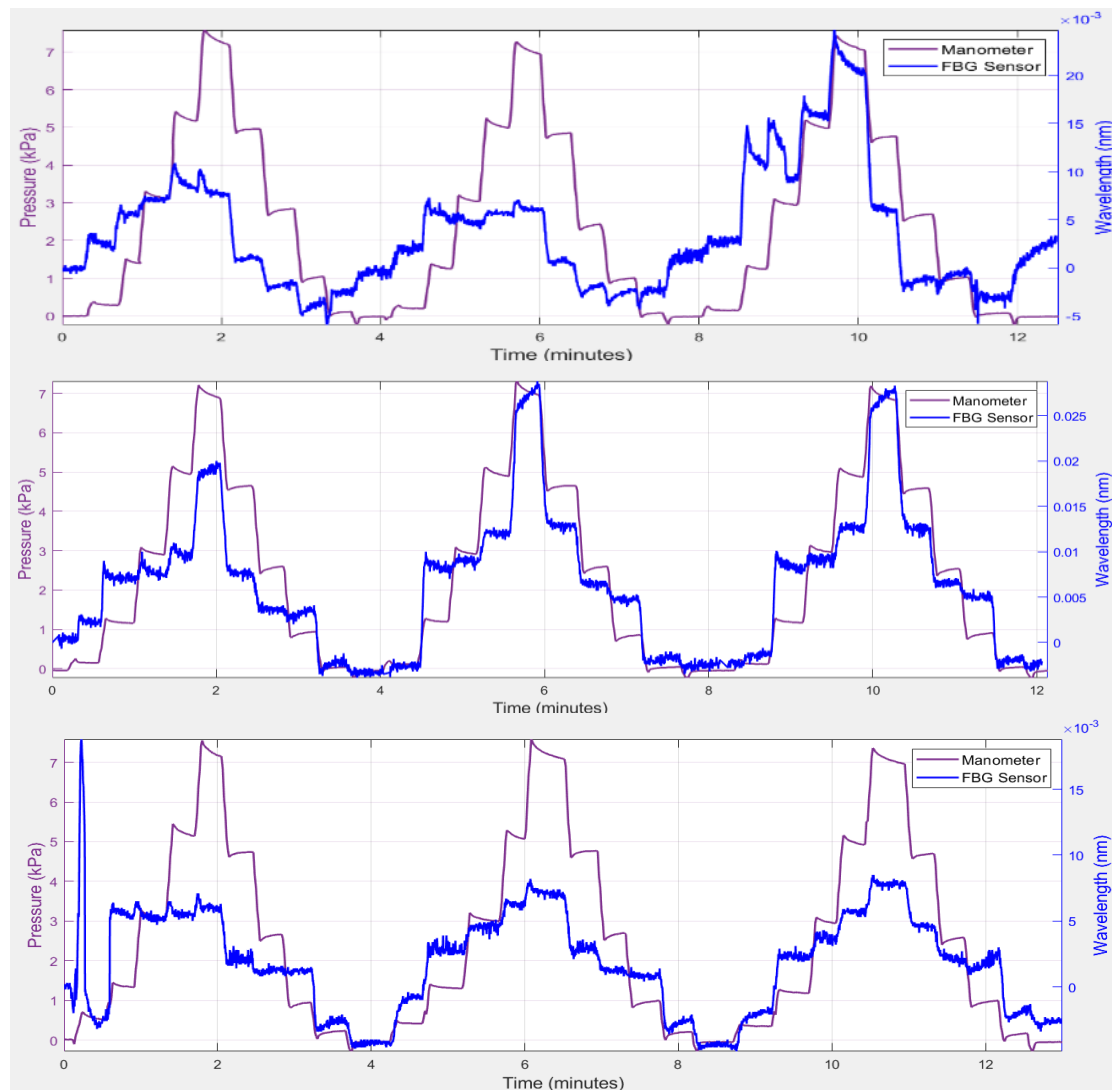


Figure 5.23: Experimental results showing the embedded sensor response to inflation and deflation. Wavelength vs time (from interrogator) and pressure vs time (from manometer). A: in the first position, on top of the ETT cuff. B: in the second position, on the side of the ETT cuff. C: in the third position, on the bottom of the ETT cuff.

Figures 5.23a, 5.23b, and 5.23c demonstrate that when the ETT cuff is inflated outside the calibration tube, the sensor detects some pressure. This is due to the inner cuff inflating and pressing against the outer cuff. Additionally, each inflation-deflation cycle produces a different response despite no changes being made to the experimental setup. Across the three experiments, the largest wavelength shift observed at approximately 7kPa is around 25pm, while the smallest is around 6pm.

These results indicate that the double cuff affects the embedded sensor's readings. The sensor detects even the very first inflation step of less than 0.5kPa. It is challenging to determine the full extent of the double cuff's impact on the calibration tube test results, as the sensor presses against the tube wall after a few kilopascals.

Furthermore, the experiments reveal that the sensor experiences different pressures on various sides of the ETT cuff, suggesting that the ETT cuff may not inflate uniformly. However, this could also be due to slight changes in the sensor's orientation as it was moved from one side of the cuff to another.

The variation observed in each cycle within the same experiment suggests that the sensor may shift during inflation-deflation cycles, even in the absence of any experimental intervention.

5.4.4 Conclusion

The embedded sensor was placed between the double cuff outside the calibration tube, and an inflation-deflation experiment was conducted. This aimed to determine if the double cuff influences the embedded sensor readings during the inflation and deflation of the ETT cuff. Results indicated that strain was still observed in the temperature FBG. Additionally, the double cuff was found to affect the embedded sensor readings, with the sensor response varying on different sides of the ETT cuff. This variation could suggest non-uniform inflation of the cuff, however this could also be caused by changes in sensor orientation when the sensor was moved to the other side of the cuff. Fluctuations within each cycle of the experiment was also observed indicating potential sensor movement between each inflation-deflation cycle.

5.5 Conclusion

Experiments were conducted to test the embedded sensors in a tube environment. The new calibration method consisted of a tube, a load cell, and an ETT. The load cell was incorporated so that it could sense wall pressure. This method aimed to determine whether the embedded sensors could be used to measure wall pressure and to provide insight into how these sensors performed in an environment where the surfaces are curved like that of the trachea.

From the experiments, the following observations were made:

- 1) The temperature FBG experienced strain making it difficult to distinguish the temperature contribution to the wavelength shift. In addition to this, the strain experienced on the strain FBG was different from that experienced on the temperature FBG. This may be due to the temperature FBG placement being more toward the edges of the ETT cuff rather than the centre.
- 2) The response from the anchored embedded sensors varied from experiment to experiment. For each experiment the sensitivity changed. This is most probably due to sensor orientation and the curvature of the calibration tube.
- 3) For the experiments that contained 2 cycles, each cycle contained discrepancies. This could be due to sensor movement within the double cuff in between inflation-deflation cycles.
- 4) When the embedded sensor was inflated on the ETT cuff outside the tube setup, results indicated that the double cuff influenced the readings of the embedded sensor, complicating the differentiation between pressure readings caused by the inner cuff pressing on the outer cuff or the cuff's contact with the wall of the calibration tube.

From these observations, the current sensor design (used in the manner of these experiments) cannot produce repeatable results and therefore does not meet the target specification for a tracheal contact pressure sensor. Despite this, there are a few changes that can be made to the sensor to improve its response however, these changes may not improve the repeatability of the sensor as the orientation of the ETT in the trachea also needs to be taken into account. Therefore, a more substantial change needs to be made,

like the consideration of a new sensor design that is not affected by the orientation of the sensor in the trachea.

When comparing these results to the literature, it is important to note differences in experimental setup. While Correia et al.'s sensor demonstrated high sensitivity on flat surfaces, it lacked a strain reference FBG (Correia et al., 2022). The results from this study highlight the importance of a strain reference FBG for tracheal contact pressure sensing. This is because, as the ETT cuff inflates, it exerts strain on the fibre. Without a strain reference FBG, it is difficult to determine how much of this strain is due to contact pressure and how much is from the ETT cuff inflation.

Additionally, Correia et al.'s sensor was calibrated on a flat surface using weighing scales. In our study, the anchored embedded sensors were also initially calibrated using weighing scales and showed promising results. However, challenges arose when these sensors were placed on a curved surface and inflated with the ETT cuff. These challenges would not have been identified without the calibration tube setup.

Previous studies in the literature did not include repeatability tests for tube sensing. In this study, repeatability tests were conducted, and results showed varying responses from each experiment. This indicates that the current sensor design is not suitable for testing in a tube-like structure and requires significant improvements and design restructuring to account for the challenges encountered.

Additionally, it is important to acknowledge that the calibration tube cannot fully replicate the complexities of a trachea, such as the presence of cartilage and muscle. Sensor orientation and location within the trachea may introduce additional variables. Although these experiments reduce the gap in environments it is possible that some of these findings may be unique to embedded sensor response in the calibration tube.

References

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Chapter 6 – Conclusions and Future Work

6.1 Conclusion

The aim of this thesis has been to optimise an FBG based pressure sensor for better sensitivity to low pressures (like those experienced on the tracheal wall during ETT cuff inflation) and to develop a new calibration technique to calibrate the sensors more accurately for measuring tracheal wall pressure. Accurate and repeatable sensors for monitoring tracheal wall pressure are highly beneficial for effective airway management, however, the current calibration technique used involves calibrating the sensor on a flat rigid surface with a loading arm. This may be considered a stark difference to the environment where the sensor needs to perform (inside a trachea with ETT cuff inflation as a means of applying pressure).

The first part of this study focused on the optimisation of the FBG based pressure sensor using materials with lower Young's modulus and higher Poisson's Ratio like rubbers and silicones as the embedding material rather than epoxy. Material selection played a crucial role in this research, with SYLGARD and Dragon Skin emerging as promising choices for contact pressure sensing. A finite element analysis (FEA) ABAQUS model was created and tested using experimental pressure values to determine if it could later be used to provide valuable insight into how to optimise the sensor further.

The second part of the thesis dived into creating and calibrating sensors on a loading arm and scale setup. SYLGARD and Dragon Skin sensors were created using a new fabrication method from BlueFrog Design that optimised the sensors surfaces to pressure measurement. These sensors had a thickness of 1mm. The ABAQUS model offered valuable insights into sensor behaviour, suggesting that sensor thickness and material properties significantly impact sensitivity. New sensors were created with a 2mm thickness and results show that the increase in thickness increases sensitivity. Additionally, adding epoxy anchors on either end of the sensors also caused an increase in sensitivity. Repeatability tests demonstrated that the optimised SYLGARD and Dragon Skin sensors exhibit consistent behaviour throughout cyclic loading, showing low hysteresis and high resolution, indicating their reliability for applications involving varying pressures. With the insights from the FEA model, the sensors were optimised for tracheal contact pressure sensing with the following target specification:

- be made of biocompatible materials

The optimised embedded sensors were made of biocompatible materials.

- response time: immediate (minimum 4 Hz)

The optimised embedded sensors showed an immediate response to loading and unloading.

- stabilisation time: immediate (< 1 second)

The stabilisation time of the optimised sensors varied, occasionally taking significantly longer than 1 second for the sensor to stabilise, indicating a need for improvement in this area. This is thought to be due to the material taking time to return to its original shape. However, it is difficult to determine the exact cause of this as the weighing scales also did not stabilise during experiments.

- be able to read pressures up to 10kPa or higher (to account for overinflation)

The optimised sensors demonstrated sensitivity to pressures up to 15kPa

- exhibit minimal/no hysteresis

The optimised sensors exhibited low hysteresis error ($0.52 \pm 0.49\mu\text{m}$ for the anchored 2mm SYLGARD sensor and to $1.4 \pm 1.3\mu\text{m}$ for the anchored 2mm Dragon Skin sensor).

- show good repeatability (coefficient of variation $< 5\%$)

The sensors showed excellent repeatability with CVs below 5% (1.13% for the anchored 2mm SYLGARD sensor, and 1.75% for the anchored 2mm Dragon Skin sensor).

- be small in size to fit on ETT cuff (ideally smaller than 1cm x 1cm and a thickness no greater than 2.5mm).

The sensors were small in size, all measuring less than 1cm x 1cm and up to 2mm thick.

- have a resolution in the range of 0.1kPa - 0.01kPa as 0.1kPa is a significant change in pressure that the sensor should detect and pressures below 0.01kPa are not significant. A sensor with a resolution below 0.01kPa are at risk of picking up minor disturbances that may not be related to the actual pressure being measured, like vibrations.

The anchored 2mm SYLGARD sensor had a resolution of 0.027kPa and the anchored 2mm Dragon Skin sensor had a resolution of 0.029kPa.

- have an ideal sensitivity of 4 $\mu\text{m}/\text{kPa}$

The optimised anchored 2mm SLYGARD sensor displayed a sensitivity of 3.67 ± 0.042 pm/kPa, while the anchored 2mm Dragon Skin sensor showed a sensitivity of 3.47 ± 0.061 pm/kPa, both approaching the ideal sensitivity of 4pm/kPa.

Overall, the results of the optimised sensors' performance indicated that they were ready for tracheal phantom testing.

The final part of this study looked at testing the optimised sensors in a simple tracheal phantom. This phantom was created using a cylinder with a diameter of 18mm which is a diameter similar to that of an adult trachea. On the side of the tube was a load cell button that allowed for wall contact pressure measurement. Sensors were placed on an ETT cuff with another cuff placed on top for protection. Results showed both the strain and temperature reference FBGs experienced strain. Furthermore, the strain experienced from both FBGs were different indicating the possibility that strain is different on different parts of the cuff. In addition to these findings, variability in sensor sensitivities was observed, probably influenced by factors like sensor movement between inflation-deflation cycles, the use of a double cuff, sensor orientation within the cuff, and the curvature of the tube.

This study highlights the importance of a strain reference FBG during ETT cuff inflation, the necessity of repeatability tests, and the need for sensor calibration in environments that closely resemble the sensors application conditions. Although literature shows similar sensors being tested in tracheal environments, they lack these experimental considerations, which could significantly influence their findings.

There are a few limitations and potential sources of bias that should be considered when interpreting the results of this study. The sample size of repeated experiments was relatively small, and the curing time for materials to reach a full cure could affect sensitivity results.

Despite these limitations, the findings of this thesis contribute to the growing body of knowledge on the use of FBG based pressure sensors for tracheal wall pressure measurement. The challenges faced underline the necessity for further research and refinement of calibration methods and sensor design to ensure the reliability and accuracy of optical sensors for tracheal applications.

6.2 Future Work

Given the demonstrated effectiveness of these embedded sensors on rigid flat surfaces, exploring new applications for this sensor is a promising avenue for further investigation. It will also be worth looking into how to apply anchors to the sensor so that they are uniform and carry out experiments to see if the size of the anchors influence sensitivity. It would also be interesting to see if more anchoring points along the fibre, possibly within the patch, would yield a higher sensitivity.

Furthermore, the expansion of experimental trials to increase the sample size is recommended, as this would contribute to more robust and accurate coefficient of variation (CV) values. This step is crucial for ensuring the reliability of the sensor's measurements.

For tracheal applications, the next step is to focus on innovative sensor designs and methods to protect the temperature reference FBG from strain. An embedded sensor that accounts for the curvature of the tube, such as a spherical patch, may improve calibration tube test results and repeatability, as this shape will not bend to the curvature of the tube, maintaining uniform deformation on both flat and curved surfaces. Attaching the sensor to the ETT cuff could eliminate sensor movement within the double cuff during inflation and deflation. Alternatively, sandwiching the embedded sensor between two rigid flat surfaces to prevent bending could address issues caused by the curvature of the calibration tube and trachea, as well as challenges associated with tracheal cartilage and smooth muscles.

Regarding the calibration techniques tailored for tracheal applications, while the calibration tube utilised an ETT cuff for inflation and a tube with a diameter like that of an adult trachea, it remains quite distant from the intricate anatomy of an actual trachea. More realistic tracheal phantoms are under development for this purpose. Experiments with this tracheal phantom may reveal new challenges for this sensor design.

Finally, an FEA model of the calibration tube setup or the trachea could be created to simulate the sensor response to pressures applied in the trachea. It could provide valuable insights into how the different sensors will perform in these applications and environments.

Appendix A

Figure 7.1 shows the mechanical properties of NOA65, and Figure 7.2 shows the mechanical properties of Vitralit 1655.

Typical Properties of NOA 65	
Solids	100%
Viscosity at 25° C	1200 cps
Refractive Index of Cured Polymer	1.524
Elongation at Failure	80%
Modulus of Elasticity (psi)	20,000
Tensile Strength (psi)	1,500
Hardness - Shore D	50
Temperature Range	-15 to 60° C

Figure 7.1: Properties of NOA65 (Norland Optical Adhesive 65, Norland Products Inc.)

Cured Material	
Hardness shore A <i>PE-Norm 006</i>	70 – 80
Temperature resistance [°C] <i>PE-Norm 059</i>	-50 – 150
Shrinkage [%] <i>PE-Norm 031</i>	<2
Water absorption [%] <i>PE-Norm 016</i>	<1
Glass transition temperature - DSC [°C] <i>PE-Norm 009</i>	30 – 40
Coefficient of thermal expansion [ppm/K] below Tg <i>PE-Norm 017</i>	<100
Coefficient of thermal expansion [ppm/K] above Tg <i>PE-Norm 017</i>	180 – 280
Surface resistance [Ohm] <i>GB/T 1410-2006</i>	5E+9 – 6E+9
Volume resistance [Ohm] <i>GB/T 1410-2006</i>	2E+9 – 3E+9
Young's modulus – Tensile test [MPa] <i>60mW/cm², 30sec, Fe spectrum + 105°C, 30min</i> <i>PE-Norm 056</i>	42 – 46
Tensile strength [MPa] <i>60mW/cm², 30sec, Fe spectrum + 105°C, 30min</i> <i>PE-Norm 014</i>	15 – 18
Elongation at break [%] <i>60mW/cm², 30sec, Fe spectrum + 105°C, 30min</i> <i>PE-Norm 014</i>	13 – 15
Lap shear strength (PET/PET) [MPa] <i>60mW/cm², 30sec, Fe spectrum + 105°C, 30min</i> <i>PE-Norm 013</i>	1 – 3

Figure 7.2: Properties of Vitralit 1655 (Technical Datasheet Vitralit 1655, Panacol Adhesives)

Appendix B

The 3 samples from AMF made from the same PDMS material as Sample 2 (UV Curable PDMS) were tested. These samples are Sample 2a, 2b, and 2c. These were cut into the same 10mm x 8mm rectangles as the original Sample 2. Figure 8.1 shows the results for Sample 2a.

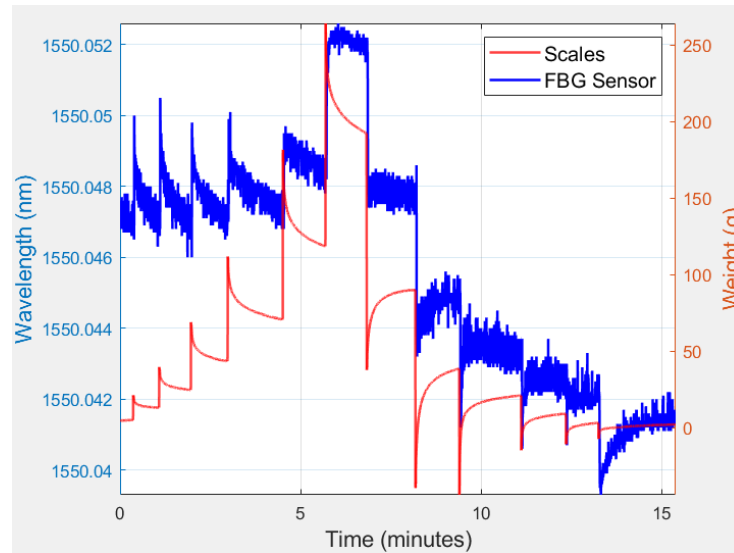


Figure 8.1: Testing results of Sample 2a.

Figure 8.1 results show a poor FBG response to loading. The baseline at the start and end do not match, and there is barely any response to loading. The test for Sample 2a should be redone.

Figure 8.2 shows the results for Sample 2b.

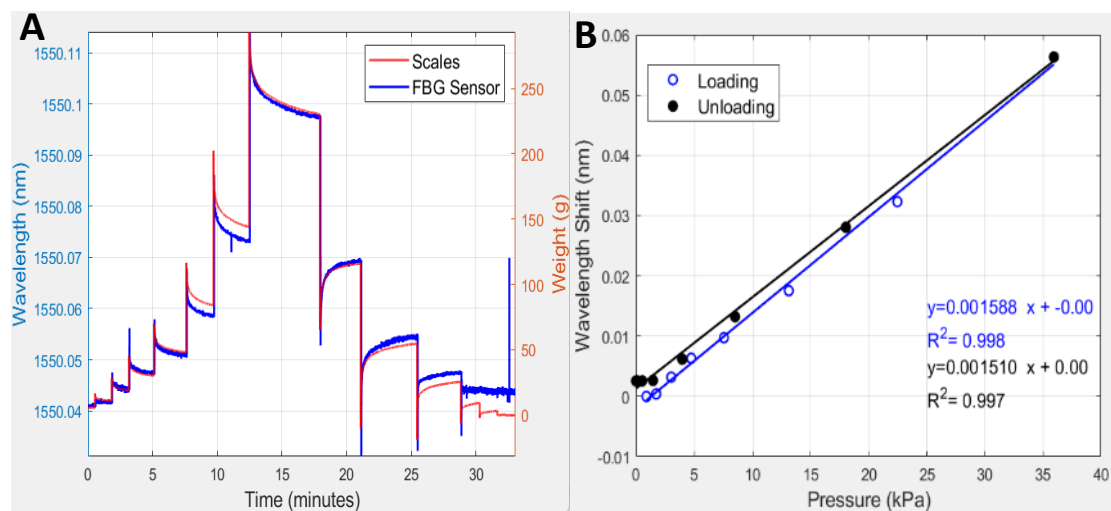


Figure 8.2: Testing results of Sample 2b. A: wavelength vs time (from interrogator) and weight vs time (from scale), B: calibration curve.

Figure 8.2a shows a better response. This was more expected as it does look similar to that generated in Figure 3.16a (Chapter 3.3.3.1). This sample produced a measurable wavelength shift after 1g was applied. Again, like in Figure 3.16a the FBG wavelength baseline doesn't go back to where it was in the beginning. The loading and unloading doesn't look smooth with both the scales and FBG taking lots of time to stabilise and in some cases may not have completely stabilised. The last 3 loads during unloading are not sensed at all by the FBG. Figure 8.2b shows a linear response with the sensitivity of this sample is 1.5pm/kPa with a hysteresis error of 0.0026nm. This is much smaller than the value from Sample 2 (UV Curable PDMS) which was 6.6pm/kPa for pressures above 2.5kPa and 3.2pm/kPa for pressures below that. This may be due to the increased number of crosslinks in the material.

Figure 8.3 shows the results for Sample 2c.

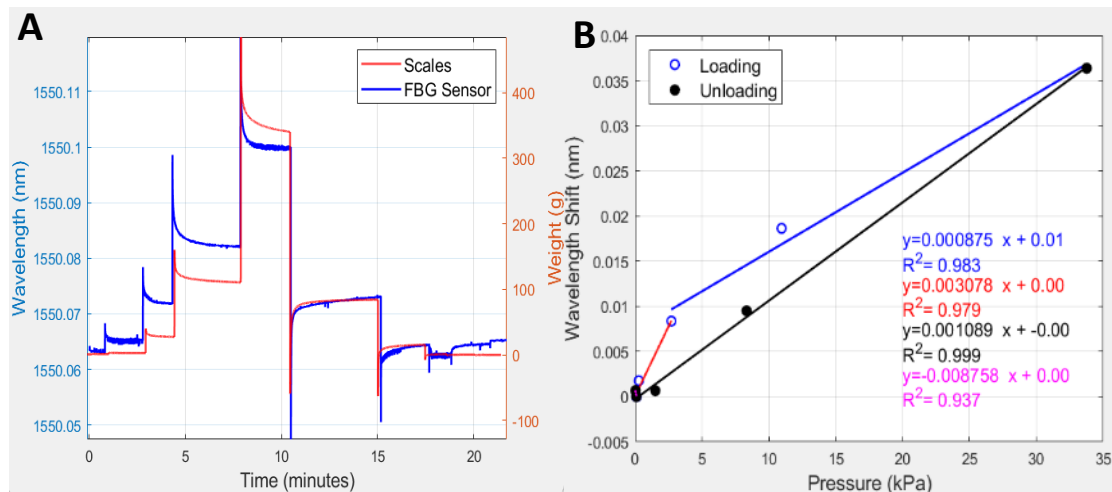


Figure 8.3: Testing results of Sample 2c. A: wavelength vs time (from interrogator) and weight vs time (from scale), B: calibration curve.

Figure 8.3a shows the same trend, not a smooth transition as each load is applied and removed. The first load removal results in a very large drop in wavelength. The baseline at the end is also different from the start. This sample also produced a measurable wavelength shift after 1g was applied. Figure 8.3b shows the sensitivity of this sample is 2.08pm/kPa with a hysteresis error of 0.0054nm, and 0.98pm/kPa for pressures above 2.5kPa with a hysteresis error of 0.0094nm. This sensitivity is even less than that of Sample 2b.

Overall, these results are not as promising as those seen in Sample 1 (SE1700), SYLGARD, and Dragon Skin. Due to this conclusion, Sample 2a wasn't tested again as this UV cured material is not desirable for the patch material.