Pressure mapping during total knee joint replacement surgery using a fibre optic pressure sensor

Mohanty, Lipi

2006

Mohanty, L. (2006). Pressure mapping during total knee joint replacement surgery using a fibre optic pressure sensor. Doctoral thesis, Nanyang Technological University, Singapore.

<https://hdl.handle.net/10356/4715>

<https://doi.org/10.32657/10356/4715>

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PRESSURE MAPPING DURING TOTAL KNEE JOINT REPLACEMENT SURGERY USING A FIBRE OPTIC PRESSURE SENSOR

LIPI MOHANTY

SCHOOL OF ELECTRICAL & ELECTRONIC ENGINEERING
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2006

**Pressure mapping during total knee joint
replacement surgery using a fibre optic
pressure sensor**

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A thesis submitted to the Nanyang Technological University
in fulfilment of the requirement for the degree of
Doctor of Philosophy

2006

Acknowledgements

I would like to thank my supervisor Associate Professor Swee Chuan Tjin for giving me this opportunity to work on this project, with his encouragement, patience and support. From core concepts to trivial solutions, many aspects of this thesis owe their present form to his advice. I am also grateful to him for giving me the flexibility to explore and implement my ideas in this research project.

The cadaveric tests were possible because of many participants from Singapore General Hospital. My gratitude goes to Dr. Silvino Panganiban, Dr. Denny Lie, Dr. Pierce Chow, Mr. Robert Ng and also the laboratory staff at Department of Experimental Surgery who helped me willingly.

I have learnt the art of moulding the tibial spacer samples from Ms. Habsah Salleh of National Dental Centre, Singapore. I thank her for teaching me with patience.

I am grateful to Assistant Professor John Ngo for spending valuable time trying to improve my scientific writing skills for my journal papers, giving feedback and help. I would also like to thank the technical staff of Photonics Research Centre who have helped whenever needed: Yong, Yvonne, Seet, Debbie, Poh Chee, Desmond, Clarence and Mr. Chow. Also, I would like to thank Thomas and Meng Kiow of IC Design III for their assistance.

My friends and colleagues have given me help and support throughout, especially Eddie Tan, Benjamin Tay and Rupali Suresh. Their cooperation has surpassed team spirit.

The encouragement and support that my family-my parents, brother and sister-have given me is beyond the scope of any acknowledgement page. Also, my in-laws, especially my mother-in-law, have helped me in every way. My daughter, who dances every time I go home, and my husband who has been there with me all this time, have inspired me to do well and do better.

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Summary

Total knee joint replacement surgery is the only solution for thousands of patients each year who are severely afflicted by osteoarthritis. The knee joint, consisting of the femoral head, the tibial condyles and the patella, may be replaced by two or three prosthetic implants during a typical surgical procedure. A pressure-mapping sensor, that can check the alignment of the prostheses, can reduce the chances of an early revision surgery.

To achieve this objective, a pressure-mapping sensor was developed with fibre Bragg gratings. Fibre Bragg gratings have been found to be useful in communications and sensing. Besides having all the advantages of an optical fibre, the fibre grating can give absolute measurement of various parameters that induce thermal or mechanical strain. The technology is immune to intensity fluctuations caused by the source or other losses such as bending.

For a pressure-mapping sensor, a specific fibre Bragg grating was designed by sampling a chirped grating. The aim was to be able to determine the magnitude of the load and the position simultaneously. The sampled chirped grating was embedded in fibre-reinforced composite, which has been a chosen material for prostheses as well. Tibial spacer samples were moulded with guidance from National Dental Centre, Singapore. The embedded grating was built into the tibial spacer. The tibial spacer sensor has an array of sensing points in the two condyles. The response of the gratings was tested individually and as an array.

The pressure-mapping tibial sensor can give the distribution of the load on it. The wavelength shifts of the individual sub-gratings, in each sampled chirped grating, were monitored. The values were used to generate a color-coded pressure chart. The sensor shows the effect of the femoral implant on the tibial spacer. It can detect varus-valgus and femoral rotation, which are the two main forms of malalignment. This sensor can also be used to detect the long-term behaviour of prostheses and help in improving their designs. A simple analytical model of the sensor has been developed to predict the response of the tibial spacer sensor and to explain the contribution of the various materials.

Simulated laboratory tests have shown the sensor to be successful in pressure-mapping across the knee joint. Cadaveric tests with the sensor, at Singapore General Hospital (SGH), have shown that the sensor can detect various forms of malalignment including varus-valgus and femoral rotation. During the total knee surgery carried out on cadaveric knees at SGH, the sensor has been tested for feasibility of use in the operating theatre. The tibial spacer sensor shows sufficient promise as a pressure-mapping sensor that can be used during the process of total knee joint replacement surgery.

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Chapter 1

Introduction

1.1 Motivation

Total knee joint replacement (TKR) surgery is a procedure carried out, usually, on patients severely afflicted with Osteoarthritis and occasionally, on accident victims. It involves the replacement of the articulating surfaces of the knee joint with metallic and polymer prostheses.

The success of a TKR procedure is defined by the number of years a patient can enjoy moderate levels of mobility. Doctors do not promise that a person can be as active as with a healthy knee! Two major factors determine this success of a TKR; firstly, the skill and experience of the surgeon carrying out the procedure, and secondly, the quality and design of the implants.

Several companies produce a variety of implants. Even more research groups are involved in the study of materials for these implants. However, nothing comes close to the natural knee. The polymer parts wear out slowly due to the constant grinding against the metal parts. The cementing materials lose their grip and eventually, a revision surgery is required to replace the implants.

The capability of the surgeon varies from one to another and is assumed to be competent so that the best alignment of the joint may be achieved. Currently, surgeons use visual observation and experience to align the prosthetic knee. This project aims to provide a pressure-mapping sensor to the surgeons to assist the surgeons in aligning the knee implants. A pressure-mapping sensor can provide the spread of contact loads across the knee joint interface and thus help the surgeons in deciding on the best possible alignment for the specific case. Proper alignment can help in alleviating the trauma and cost of an early revision surgery caused due to malalignment. Besides being used during the TKR procedure, prostheses manufacturers can also use the sensor to study the response and quality of their designs.

1.2 Objectives

The main objective of this project is to design and develop a pressure-mapping sensor that can be used in-vivo, during total knee joint replacement surgery. The entire project covers the following steps.

1. The fundamental sensor is based on optical fibres and the sensing medium is a fibre Bragg grating. The optical fibre has sensing capability and the advantages of small size, flexibility and multiplexing capacity. The fibre property that is modulated is the wavelength of light passing through it. Using the wavelength modulation property of fibre gratings ensures immunity to various types of loss in intensity due to bending, coupling and back scattering. A specific type of fibre grating was designed to obtain the load and position information simultaneously.

2. The fibre grating was embedded in a suitable fibre-reinforced composite material to enhance the mechanical strength, define its directional properties and reduce the effects of birefringence. This embedding influences the properties and response of the sensor. The behaviour of the embedded fibre grating was monitored individually and in an array.
3. The samples of the sensor were made in the shape of tibial spacers moulded in the laboratory. The embedded array of fibre gratings was built into the tibial spacer sample to give the final design of the sensor. The tibial spacer sensor was calibrated in the laboratory.
4. The sensor data, obtained from the laboratory testing, was supported by analytical understanding of how the properties of the fibre vary under different load conditions. The theoretical model incorporates the effect of the various parameters influencing the sensor response.
5. The culmination of the project was in cadaveric testing of the sensor in a simulated environment of total knee joint replacement surgery. The cadaveric model was used to test the viability of the sensor for in-vivo use during surgery. It provides the background to propose similar sensors for the shoulder and hip joint replacement procedures.

1.3 Major contribution of research

Research in pressure mapping has been pursued and is ongoing for applications in ergonomics, robotics, pressure mats for the automotive industry and healthcare (to prevent bed sores). Pressure mapping sensors exploring the joints and crevices in

orthopedics have also existed for many years. The aim of this thesis has been to develop a pressure mapping sensor that can guide surgeons *during* the total knee joint replacement procedure.

The first part of this thesis presents a novel fibre-optic sensor that can detect the load and position simultaneously: the embedded sampled chirped grating. The advantage of this sensor lies in its simplicity of concept and design.

The preliminary idea of a single sampled chirped grating has also been extended into an array. This pressure-mapping array can have a wide range of applications as it can detect the movement of point loads on the sensor, when connected to a dynamic interrogation system.

The sensor was built into the polymer prosthesis used for the tibia at the knee joint. This sensor design proposed in the thesis can be fundamental in developing similar sensors for the glenoid component in the shoulder, the acetabular socket of the hip, or other orthopedic interfaces. It can be modified for any design of the polymer implants.

Cadaveric tests have been conducted with the sensor and the results provide sufficient evidence that the fibre-optic pressure-mapping sensor can be used during the process of surgery. The sensor provides pressure maps that can help the surgeons to correct the alignment of the prostheses and guide the surgeons.

1.4 Organization of the thesis

This thesis describes the various stages of research work undertaken. This chapter introduces the motivation, objectives and major contributions. The following seven chapters will present the details involved in developing and testing the sensor.

In Chapter 2, the background on fibre optics and in particular the fibre Bragg grating sensor is presented. It begins with a simple understanding of fibre Bragg gratings with some background of the previous work done, that has guided this research.

Chapter 3 explains the complexities of structure involved at the knee joint with the bones and muscles. An overview of the total knee joint replacement surgery is given. The problems of pressure mapping at the tibio-femoral joint are discussed and prevalent methods of study are introduced.

The entire experimental procedures involved in the design and development of the sensor are given in Chapter 4. A few guiding tests are also presented that have influenced the choices made in the laboratory.

In Chapter 5, all the tests done in the laboratory are presented. This chapter includes the investigation of several properties such as sensitivity and array response. It also mentions the generation of the pressure map display with basic software programming.

The analytical background for the sensor is given in Chapter 6. The calculations and theory influencing the results are covered in this chapter.

Chapter 7 presents the methods, results and discussion for the cadaveric tests carried out. Malalignment and angles of flexion are studied. In this chapter, various contact conditions of the knee joint are simulated to test the response of the sensor.

The conclusions of this research and recommendations for future work are discussed in Chapter 8. This chapter is followed by a list of references and a few appendices that provide supplementary information.

Chapter 2

Fibre Bragg Gratings

The objective of this research work is to develop a pressure-mapping sensor for knee joint surgery. As the knee joint is a complex structure of various curved surfaces, a sensor that can adapt to complex contours is required. Due to their small size, flexibility, sensing capability, and other advantages, optical fibres were chosen as the sensing medium. This chapter will introduce the technology of fibre optics that forms the core of this project and is the basic principle of the sensor.

2.1 Fibre optic sensors

Fibre optics has made major inroads into the field of sensors [1] besides overhauling communications [2]. The optical fibre, based on the simple principle of total internal reflection from a rarer medium [3], has become a powerful tool for sensing in harsh and restricted environments [4]. The optical fibre is made of glass (silica), which makes it immune to many chemicals, electromagnetic effects and nuclear hazards [5]. This gives fibre optic sensors added durability in specific environments. A single strand of fibre has a diameter of 125-250 μm . The size allows an optical fibre to slide into crevices, holes, and tiny apertures or in between surfaces. The fibres can be embedded in "smart" materials and structures. Fibre sensors also have a longer life, without experiencing the fatigue common with conventional piezoelectric or silicon gauge sensors.

For these and many other reasons, the optical fibre is finding many applications in various fields of sensing. These sensors have been used for a variety of applications, including:

- Strain sensors that monitor the curing of composite materials [6].
- Sensors that measure blood flow in cardiology [7].
- Temperature sensors for use in oil pumping [8].
- The monitoring of strain in concrete, composite and aluminum bridges [9].

Fibre optic sensors are of different types depending on the principle of operation and sensing. A simple classification of the fibre optic sensors would be as follows.

1. Intensity based sensors
2. Phase (birefringence) based sensors
3. Wavelength based sensors

As the names imply, the basis of their operation is the modulation of intensity, phase and wavelength, respectively.

Intensity based sensors are simple and only need a detector to monitor the response. However, it has two major drawbacks as it is subject to source power fluctuations and various losses like bending loss, coupling loss and back end scattering [10].

Phase based sensors are highly sensitive, which may be an advantage for some applications. However, this property could lead to cross sensitivity between different parameters and difficulty in isolating the effects. Also, it needs a very stable arrangement to give results with high accuracy – a simple twist of the fibre can distort results. In many cases, a lot of optical instruments are also involved [11, 12]. These factors influence the absence of practical application in most sensing environments.

Another category of sensors are wavelength based sensors such as fibre grating sensors. Fibre grating sensors have a grating built into the core of the optical fibre. The period of this grating changes with applied perturbations and is encoded in the reflected or transmitted wavelength. This makes fibre grating sensors immune to source power fluctuations and greatly reduces the effects of bending/coupling losses. The wavelength encoding gives the added advantage of multiplexing by which a single strand of fibre can bear multiple sensors of different wavelength along the length.

Fibre gratings are of many types such as the uniform fibre Bragg grating, the chirped fibre grating, sampled gratings, superstructure gratings and long period gratings. Whereas all of them are guided mode gratings, the last one is a cladding mode grating. However, the basic concept of these gratings is the same and can be understood from the principle of the uniform fibre Bragg grating (FBG).

2.2 The fibre Bragg grating:

The fibre Bragg grating is a structure comprised of a periodic change in the refractive index along the core of an optical fibre [13]. It is based on the principle that, when light is coupled into the fibre, the wavelength satisfying the Bragg condition (equation 2.1) is reflected whereas all other wavelengths are transmitted without much attenuation (Appendix 1).

Figure 2.1. Basic principle of wavelength shift in fibre Bragg gratings. The normal grating reflects a certain wavelength of light (a) whereas the strained grating (b) reflects a different wavelength of light depending on the change in the grating period.

Figure 2.1 shows an optical fibre with a grating (transverse planes of varying refractive index) created in the core. When light is allowed to pass through the fibre, some wavelengths (with central wavelength λ_B) are reflected while other wavelengths are transmitted. However, when the grating experiences a strain, the period of this grating changes resulting in the reflection of different wavelengths (with central wavelength $\lambda_B + \Delta\lambda$).

Thus, any change experienced by the grating is observed as a wavelength shift ($\Delta\lambda$) or bandwidth change in the reflected spectrum. In an FBG, the grating in the fibre core couples the forward going LP_{01} guided mode to the backward-going LP_{01} mode at a specific Bragg wavelength, causing the reflection.

The Bragg wavelength reflected by the FBG is given by

$$\lambda_B = 2n_{eff}\Lambda \quad (2.1)$$

where n_{eff} : Effective refractive index of the core at the Bragg wavelength

Λ : Period of the FBG

The phenomenon that led to the development of FBGs was the discovery by Kenneth Hill in 1978 [14]. He reported that certain optical fibres were photosensitive to ultraviolet light (UV), which could be used to create a refractive index modulation in the core. Though the phenomenon is yet to be thoroughly understood, the applications have brought a major change in the field of communications [15,16,17] and are being explored

for sensing [18,19,20]. At this stage, there is a consensus among researchers that this photosensitivity of the optical fibre to UV radiation is due to the formation and annihilation of certain defects in the silica-germanium complex. It is believed that photosensitivity is a consequence of colour centres [21].

Photosensitivity is generally observed in optical fibres with a high concentration of germanium (3%) or in fibres co-doped with germanium and boron. It may be noted that ordinary, telecommunication fibre has about 0.3% germanium and is not photosensitive. Lemaire and co-workers were first to report a simple but effective method to make germanium-doped fibre sensitive to ultra violet light [22]. In that process, optical fibres are immersed in a chamber of hydrogen gas maintained at temperatures of 25-75°C and at pressure ranging from 2.02 MPa to 75.75 MPa. This loading of hydrogen causes diffusion of hydrogen atoms into the fibre core. This method results in very high photosensitivity and permanent refractive index changes up to 0.01. No single model can explain all the experimental results, as there are many different microscopic mechanisms involved. The influence of the laser inscription wavelength, laser power, fibre types and process of hydrogen loading lead to different reaction mechanisms. However, this technique of hydrogenation to make fibre photosensitive has made it easier for researchers to study FBGs, as telecommunication fibre is at least 10 to 20 times cheaper than specially doped fibre.

Another major discovery that advanced FBG research was the report of the phasemask technique [23] that can be used to inscribe fibre gratings. A phase mask is a

relief grating etched in a silica plate (usually). It is a diffractive element that can be used to form an interference pattern laterally. The main features of the mask are the grooves etched into a UV transmitting silica mask plate, with a carefully controlled mark-space ratio as well as etch depth. The grating period obtained after inscription is half of the phase mask period. The design depth d of the mask is influenced by the UV operation wavelength [24]. The period of the phasemask determines the Bragg wavelength and the effective index of the mode.

Silica is used as a material for phasemask because it transmits mid-UV light. The phase mask is used to split the incoming beam into two diffraction orders, +1 and -1, with an equal power level. One advantage of the phase mask is that it is not sensitive to environmental conditions, such as vibrations. It requires the coherence of the UV laser source to imprint the interference pattern. Another major advantage is that the results are very consistent, and do not depend on the setup configurations. This allows the user to generate gratings with high repeatability. The sole disadvantage is that different phase masks are required if the resonant wavelengths of the fibre Bragg gratings are changed.

Though researchers use alternative methods such as the holographic method [25], the combination of hydrogen loading followed by the phasemask technique are the most popular approaches. This technique can be used to generate uniform fibre Bragg gratings, chirped, sampled and superstructure gratings with a variation of the steps or tools (phasemask). In this work, chirped and chirped sampled gratings have been used to develop a pressure mapping sensor.

2.3 Chirped fibre grating

The chirped fibre grating (CFG) is similar to a uniform fibre Bragg grating (see Appendix 1). It is made of a continuum of Bragg reflectors of increasing wavelength. The simplest case is that of a linearly chirped grating in which the grating period is a linear function of the position. A chirped fibre grating has a wide reflected spectrum bandwidth because of the chirp in the grating period. The Bragg wavelength in a chirped fibre grating is a function (linear or non-linear) of the position z ,

$$\lambda_B = 2n_{eff}\Lambda(z) \quad (2.2)$$

where $\Lambda(z)$ is the period of the FBG that is a function of the position z .

$$\Lambda(z) = \Lambda_0(1 + c_p z), \quad (2.3)$$

c_p is the chirp in the period of the grating.

The grating reflects a continuous range of wavelengths i.e. the shortest period reflects the shortest wavelength while the longer periods reflect longer wavelengths. In other words, each point on the reflected spectral bandwidth is an image of a corresponding point along the length of the grating. This factor can be used as an advantage in pressure mapping. Generally, the chirped grating spectrum differentiates between the location of the applied force (see Figure 2.2) unlike the uniform FBG.

Figure 2.2 shows the reflected spectra of a uniform FBG and a CFG. Both gratings are of the same length (for example, 1 cm). As a point stress is applied along different parts of the 1 cm length of both gratings, the chirped grating spectrum will show a change at different points whereas the FBG spectrum shows peak shift.

Figure 2.2. Stress applied at different points on a grating and the corresponding affected point of the reflected spectrum. The black traces denote the reference spectrum. (a) When stress is applied at point X (orange trace) or point Y (green trace) along the uniform grating, the effect will be observed as a shift of the entire spectrum. (b) When stress is applied at point X or point Y along the chirped grating, the effect will be observed as localized changes at two different points in the spectrum. The spectra have been displaced on the intensity scale for illustration.

2.4 Sampled chirped fibre grating

The sampled chirped grating combines some advantages of the chirped and uniform fibre gratings. A sampled grating is a conventional grating with grating elements

removed in a periodic fashion [26]. A sampled chirped fibre grating (SCFG) is a sampling made of a chirped grating. Effectively, it is a periodic modulation of the reflected/transmitted intensity where the periodic modulation leads to reflection spectra with periodic maxima. The modulation can be done as a change in the grating structure during the inscription of the grating. Mathematically, it may be explained as the multiplication of a rectangle function with the reflected amplitude [27].

Figure 2.3 shows the principle of removing parts of a wide spectrum to give the sampled spectrum. A single, continuous wide-band peak can be inscribed selectively, in portions, to generate a sampling of narrower peaks.

Figure 2.3. The concept of sampling a wide-band spectral peak into narrower peaks.

(a) A selected portion of the grating spectrum is removed. A single, continuous grating is split into two sub-gratings by blocking a part of the phasemask during inscription. This results in two separate peaks in the reflected spectrum (b).

2.5 Principle for sensing

The basic principle for sensing in the case of uniform FBG, CFG and SCFG is the same. The perturbation affects the period of the grating and the effective refractive index consequently changing the reflected wavelength. In most cases, the change in the effective refractive index is negligible as compared to the change in the grating period. It is further explained as follows.

The wavelength shift experienced by a fibre grating is influenced by two main parameters, which are strain and temperature. Any perturbation that can induce a strain or alter the temperature can be detected by this simple principle of wavelength shift. This is one of the major advantages of a fibre Bragg grating i.e. the measurand information is wavelength encoded. Any interrogation system that can monitor the change in central wavelength can be used to detect the response of a fibre grating sensor.

The following equations can be obtained by differentiating equation 2.1 and demonstrate the effect of strain and temperature on the shift in the Bragg wavelength $\Delta\lambda_B$. The effective refractive index and the grating period are differentiated with respect to the variables of length and temperature.

$$\Delta\lambda_B = 2n_{eff} \left(\frac{\partial\Lambda}{\partial l} \Delta l + \frac{\partial\Lambda}{\partial T} \Delta T \right) + 2 \left(\frac{\partial n_{eff}}{\partial l} \Delta l + \frac{\partial n_{eff}}{\partial T} \Delta T \right) \Lambda \quad (2.4)$$

Multiplying and dividing the terms in the right parenthesis with n_{eff} and the left parenthesis with Λ , respectively

$$\Delta\lambda_B = 2n_{eff}\Lambda\left(\frac{1}{\Lambda}\frac{\partial\Lambda}{\partial l}\Delta l + \frac{1}{\Lambda}\frac{\partial\Lambda}{\partial T}\Delta T\right) + 2n_{eff}\left(\frac{1}{n_{eff}}\frac{\partial n_{eff}}{\partial l}\Delta l + \frac{1}{n_{eff}}\frac{\partial n_{eff}}{\partial T}\Delta T\right)\Lambda \quad (2.5)$$

$$\Delta\lambda_B = \lambda_B\left(\frac{1}{\Lambda}\frac{\partial\Lambda}{\partial l}\Delta l + \frac{1}{\Lambda}\frac{\partial\Lambda}{\partial T}\Delta T\right) + \lambda_B\left(\frac{1}{n_{eff}}\frac{\partial n_{eff}}{\partial l}\Delta l + \frac{1}{n_{eff}}\frac{\partial n_{eff}}{\partial T}\Delta T\right) \quad (2.6)$$

Substituting equation 2.1 in equation 2.5 and reorganizing the terms gives the following.

$$\frac{\Delta\lambda_B}{\lambda_B} = \left(\frac{1}{n_{eff}}\frac{\partial n_{eff}}{\partial T}\Delta T + \frac{1}{\Lambda}\frac{\partial\Lambda}{\partial T}\Delta T\right) + \left(\frac{1}{n_{eff}}\frac{\partial n_{eff}}{\partial l}\Delta l + \frac{1}{\Lambda}\frac{\partial\Lambda}{\partial l}\Delta l\right) \quad (2.7)$$

The Bragg wavelength is influenced by the strain-optic (change in n_{eff} due to Δl) and the thermo-optic effects (change in n_{eff} due to ΔT) that lead to a change in the material properties and a consequent change in the effective refractive index [28]. Similarly, the period of the grating (Λ) is affected by a change in length of the fibre caused by strain (compression/tension) or thermal expansion (or contraction). Thus, any wavelength shift observed in the reflected/transmitted spectrum is the sum of both temperature and strain effects. In many sensing applications for the measurement of strain, where severe temperature fluctuations (tens of degrees) are expected, the temperature effect has to be compensated [29]. The thermal effect can be neglected only when all measurements are taken at the same temperature. For example, when the effect of pressure is measured at 22°C, the central wavelength at that temperature can be used as the reference to determine the wavelength shifts.

2.6 Lifetime of fibre Bragg gratings

The response of a fibre grating sensor is influenced by extrinsic factors such as stress and temperature. However, some intrinsic properties such as the stability (of reflected intensity) also influence its application in sensing. When fibre gratings are used for sensing, it is important to know that whatever treatment (such as hydrogen loading) is given to the fibre before exposure to UV, there is always some thermal decay of reflectivity even at room temperature. To arrest the thermal decay at room temperature and any consequent error in the response of the sensor, the fibre gratings are subjected to accelerated ageing or annealing before the grating is used for any measurements. This ensures that most of the thermal decay occurs before the grating sensor is implemented and the reflectivity is stabilized. In this process, the fibre grating is subjected to a high temperature (30-50 degrees higher than room temperature) treatment for a few hours. However, there is no temperature that is universally applicable for all types of fibre.

The mechanism of this annealing is understood as follows. The UV-induced defects are not thermodynamically stable. The change in refractive index is due to defects or sites, which have an associated energy and are reversible in nature. This associated energy can be explained as an activation energy barrier that has to be overcome to cause the refractive index decay. The defects that have lower energy barrier decay faster than those with a higher energy barrier. It has been observed that initially there is a rapid decay followed by a slowing but non-zero decay. This thermal decay is much greater in the case of hydrogen-loaded fibres as compared to other photosensitive fibres [30]. Thus,

to ensure stability of the gratings, annealing is undertaken as a part of the sensor fabrication process.

2.7 Fibre grating pressure sensors

Besides the intrinsic properties of fibre gratings, the actual design or configuration of the sensor will also be determined by the parameter to be measured. In this case, the configuration of the fibre grating sensor has to map contact pressure in the knee joint requiring the quantification of the load and the location/distribution of the load. Pressure is defined as the force per unit area and can be measured for a solid, liquid or gas. The choice and use of a pressure sensor depends on the particular application. There are mainly two classes of pressure measuring instruments: one that requires electrical power to function (electromechanical) and the other, which does not need electrical power. The former are also known as pressure transducers. In the latter category, we have bellows, diaphragms, Bourdon tubes and manometers. In this report, we will consider the pressure experienced by one solid body when another solid body acts on it. When a solid is subjected to a pressure, it will undergo deformation in length and also volume. This factor can be used to design a pressure sensor that responds due to the electromechanical effect or the photoelastic effect [31].

There are many high quality electromechanical pressure sensors that are commercially available [32, 33]. In an electromechanical sensor, the sensing element can be a diaphragm. Some examples are capacitive pressure sensors, semiconductor (silicon) strain gauges and piezoresistive IC. All these sensors are monitored by the modulation of

the electrical signal by the transducer. The main disadvantage is susceptibility to electromagnetic interference (EMI). Also, most of the sensors are unable to determine the point of application of the pressure. Pressure sensors that are entirely free from EMI and that rely on the photoelastic effect are, generally, based on optical fibres.

In the last decade, fibre Bragg gratings have found many applications in sensing parameters such as strain and temperature. Bragg gratings also solve the problem of reproducibility in production, which is a drawback of mass-produced, miniaturized, and low-cost disposable sensors [34].

The range of pressure sensitivity of a bare fibre Bragg grating inscribed in telecommunication fibre is not very large [35]. The usual procedure is to “amplify” the effect by attaching the grating to a transducing element, which translates the pressure into strain in the fibre. The actual design of the transducing element depends on the required shape, size and sensitivity specific to the application.

The sensitivity of a fibre Bragg grating has also been expressed as follows [28].

$$\frac{\Delta\lambda_B}{\lambda_B} = \varepsilon \left[1 - 0.5 \times n_{eff}^2 \{ p_{12} - \nu(p_{11} + p_{12}) \} \right] + \xi \Delta T \quad (2.8)$$

where ε is the axial strain, p_{ij} are the strain optic coefficients, ν is the Poissons ratio and ξ is the thermo-optic coefficient.

Xu et al. reported the effect of hydraulic pressure on FBG [36]. They have used a bare FBG and reported the sensitivity to be -2.02×10^{-6} /MPa. Later, they improved the

sensitivity by one order of magnitude by using a glass bubble housing. Liu et al. have enhanced the sensitivity to -6.28×10^{-5} /MPa by coating the fibre grating with a polymer [37]. Zhang et al. have improved the sensitivity by encasing the FBG with polymer in an aluminum cylinder. They have reported a sensitivity of -3.41×10^{-3} /Mpa [38]. All these designs work on the principle of strain on the FBG, which results in wavelength shift of the grating. However, none of the above designs are suitable for use in pressure mapping and do not offer the mechanical strength and flexibility required for use across the knee joint.

Some groups have also reported the effect of direct application of pressure as in the case of transverse loading [39]. Transverse loading also introduces birefringence effects in the spectrum of the grating, usually evident in the form of spectral splitting.

The pressure sensor reported by Tjin et al. offers high mechanical strength, low sensitivity to temperature and flexibility for the measurement of contact pressure [40]. The sensor is made of an FBG embedded in fibre-reinforced composite. The sensor measures direct pressure, as the applied pressure is translated into axial strain in the fibre by the composite layers; but it can only give the excess pressure experienced by the entire grating. The requirement for total knee joint replacement surgery is to quantify the pressure at multiple points, which is the major objective of this research work.

Many reports in the literature can be found with the uniform grating as the sensing element as in all the cases mentioned above. Though the uniform grating has its

advantages, the chirped fibre grating also has potential for applications in distributed sensing due to its wide bandwidth that can provide spatial resolution of the location of strain. Some groups have recently demonstrated the effect of local pressure on the chirped fibre grating with experimental [41] and simulation [42] results.

Combining the chirped grating properties with the sensor concept reported by Tjin et al. [40] can give a suitable sensor to measure direct pressure under different loads at multiple loading points along the sensor. This optimization can be used to develop a fibre grating sensor for the knee joint that can map the distribution of pressure.

2.8 Scope of fibre grating sensors in biomedical applications

The fibre grating sensor has to be used across the knee joint and it is necessary to know that the optical fibre sensor can be suitable for biomedical applications. In medicine, one of the first applications of optical fibres has been in endoscopy [43]. Optical fibre-based sensors have also raised interest in the measurement of various physiological parameters and potential biomedical applications are still being discovered. The intrinsic properties of electrical isolation and immunity to electromagnetic interference are useful in avoiding the risk of electrical shock when high voltages are applied to a patient, such as for resuscitation through defibrillation after sudden cardiac arrest. The small size of optical catheter probes is ideally suited for the in-vivo measurement of organ functions through minimally invasive procedures [44]. There is minimal tissue trauma due to the dielectric properties of the optical fibre and the small sensor size. Sensor material (glass fibre) is also inert that prevents any undesirable

immunity response from the human defense system. Such a reaction can lead to the aggregation of plaque, inflammation and coagulation. In this perspective, it is considered advantageous if the sensor can be intrinsic to the optical fibre as in the case of fibre gratings. During certain procedures [45], high electric or ultrasonic pressure fields can result in temperature gradients of tens of degrees Celsius per centimetre that need to be accurately mapped and offer scope for FBG sensors. As minimally invasive sensors, they have a lot of potential and can be inserted into catheters and hypodermic needles. The use of fibre grating sensors, for medical applications, has already started [46] but in very small steps.

The properties of optical fibre establish the fact that it is a suitable medium for biomedical sensing and also in this research. Size, flexibility and inert nature suggest that optical fibres can be used for the knee joint. Also, wavelength encoded response from a chirped fibre grating can be used for simultaneous measurement of pressure and position. The development of the sensor is influenced by the anatomy of the knee joint and the joint replacement surgery, which is discussed in the next chapter.

Chapter 3

Knee joint surgery

The properties of fibre grating sensors show the potential to probe the knee joint. The healthy knee involves the coordinated function of bones, cartilage, ligaments, muscle, fluid and other tissue. The fibre sensor has to probe the interaction between the tibia (shin bone) and the femur (thigh bone) during total knee joint surgery. For this purpose, it is crucial to have a basic understanding of the anatomy of the knee joint, the knee joint replacement procedure and the actual requirement of the sensor. This chapter presents the target area for the fibre optic sensor, which is the knee joint.

3.1 Knee joint

The knee is a large synovial joint (see Appendix 2). It consists of three components of bone: the upper end of the tibia, the lower end of the femur and the patella (knee cap). The surfaces of these three bones that come into contact with each other are covered with articular cartilage. The weight bearing joints are the two condylar articulations of the tibio-femoral joint (between the two femoral condyles and the two tibial condylar grooves), with the third articulation being the patello-femoral joint [47] (Figure 3.1). The structure of the knee allows it to bear tremendous loads as well as provide the mobility required for locomotor activities. A typical example is that the knee carries about five times the body weight during normal walking [48]. So a 50 kg person loads his knees with 250 kg. The magnitude of this load increases to seven times the body

weight during activities such as stair climbing. These factors contribute to tibiofemoral pressure, which is the result of the tibiofemoral condylar articulation, in the case of a healthy or a prosthetic knee.

Figure 3.1 illustrates the basic structure of the knee when the knee is in extension (as in the case of standing) and in flexion (when the knee is bent). In a healthy knee, the condylar and patellar articulations are controlled by the ligaments and the tendons. Osteoarthritis causes severe damage to the smooth articular cartilage. Figure 3.2 shows a close-up view of the knee joint with the patella removed from the front view. The mechanisms of the supporting ligaments and the cushioning effect of the menisci are lost during most TKR procedures.

Figure 3.1. Illustration of the knee joint in flexion and extension.

(Source: www.arthroscopy.com)

Figure 3.2. Illustration of the knee joint articulating surfaces.

(Source: www.arthroscopy.com)

The tibiofemoral joint is the main load-bearing interface in the knee. The tibiofemoral joint is loaded both in compression and shear during daily activities. It is impractical to isolate the forces of compression, shear or friction at this interface. Weight bearing and tension in the muscles, ligaments and tendons crossing the knee, contribute to these forces. Compression dominates when the knee is fully extended. The medial and lateral condyles of the tibia and the femur articulate to form two side-by-side condyloid joints [49] (Figure 3.2) that function as a modified hinge joint because of the ligaments, with some lateral and rotational movement allowed. The condyles of the tibia, known as tibial plateaus (or grooves), form slight depressions separated by a region known as the intercondylar eminence. The menisci that border the condylar grooves help in force

absorption and in the distribution of loads at the tibiofemoral joint over a broader area, thus reducing the magnitude of joint stress.

Due to the interdependent and complex mechanisms of the ligaments, tendons and bones, it has been said that the knee joint is a “nightmare” from the engineering perspective [50]. The stress distribution at the tibiofemoral interface during standing, sitting or walking, is the combined effect of these mechanisms.

3.2 Osteoarthritis

The root cause of knee joint replacement lies in osteoarthritis. About 20% of people over 65 years are affected by osteoarthritis, the most common form of disability among older men and women [51]. It is also seen, nowadays, in an increasing number of patients who are much younger. Arthritis of the knee is, by definition, a wearing away of cartilage. Articular cartilage is a firm rubbery protein material covering the end of the tibia and the femur that acts as a cushion or shock absorber between the bones. The knee is a synovial joint (Figure 3.3) and this cartilage across the tibiofemoral joint is lubricated and cushioned by the synovial fluid. The drying up of this fluid begins the degeneration of the cartilage. Wearing away of this cartilage leads to the bones facing the impact of the body weight in all activities and soon the bones grind against each other. Such a condition can result in degeneration of the knee joint, causing pain. The principal symptoms of osteoarthritis are pain and restricted joint movement. A painful knee can severely affect a person’s ability to lead a full active life with some patients experiencing pain in the knees even at rest.

Figure 3.3. Schematic diagram of the knee joint area showing the joint interface surrounded by the synovial membrane. Osteoarthritis affects the articular cartilage.

(Source: www.arthroscopy.com)

Figure 3.3 shows the schematic of the knee joint with the cartilage and synovial fluid. Osteoarthritis wears away the synovial fluids and the articular cartilage causing the bones to grind together and lose their surface structure at the joint.

Some statistics of Americans opting for total knee joint replacement obtained from the website: <http://www.aaos.org/wordhtml/research/stats/arthropl.htm>, are shown in Table 3.1. Note that surgery may be a result of severe, debilitating accidents as well.

Table 3.1. TKR surgeries carried out on Americans in different years.

Year	1991	1992	1993	1994	1995	1996	1997	1998	1999	2000
Total Knee Replacement	160,000	167,000	179,000	209,000	216,000	245,000	259,000	266,000	267,000	299,000

3.3 Knee joint prostheses

Many patients choose to (perhaps are compelled to) undergo surgery that can ensure many pain-free years. A surgical procedure to alleviate pain may be carried out on a part of the knee or on the whole knee. The latter is known as total knee joint surgery and includes a replacement with implants/prostheses in most of the cases. Total knee joint replacement involves an operation in which the bones at the knee joint are re-shaped and artificial implants are cemented into place [52, 53]. The implants consist of three components: the femoral component (made of a highly polished strong metal), the tibial component (made of a durable plastic spacer fitted into a metal tray), and the patellar component (plastic). The materials and designs of the implants may vary between hospitals depending on the decision of the surgeons.

Up to three bone surfaces may be replaced during surgery: the lower ends of the femur, the top surface of the tibia and the back surface of the patella. Components are designed so that metal always articulates against plastic, which provides smooth movement and results in minimal wear.

1. Femoral component: The metal femoral component curves around the end of the thighbone and has an interior groove so the kneecap can move up and down smoothly against the bone as the knee bends and straightens. Usually, one large piece is used to resurface the end of the bone. If only one side of the thighbone is damaged, a smaller or half piece is used (unicompartmental knee replacement) to resurface that part of the bone.
2. Tibial component: The tibial component consists of a flat metal platform with a polyethylene cushion. The cushion may be part of the platform (fixed) or separate (mobile) with either a flat surface or a raised, sloping surface. This cushion is also known as the tibial spacer. The long-term success of a knee implant procedure is determined by the properties of this spacer as it wears out eventually.
3. Patellar component: The patellar component is a dome-shaped piece of polyethylene that resembles the shape of the kneecap. This polyethylene piece is attached to a flat metal plate. This component is used only when the patella has suffered damage.

Several manufacturers make knee implants [54, 55] and there are more than 150 knee replacement designs available in the market. The brand and design used by the doctor or hospital depends on many factors, including the patient (age, weight, activity level and health), the surgeon's experience and familiarity with the device, and the cost and performance record of the implant. The implants chosen for this project (Zimmer, Insall Burstein II, see Figure) are same as those used at Singapore General Hospital and are shown in Figure 3.4.

(a)

(b)

Figure 3.4. Pictures of the metallic femoral implant (a) and the polyethylene tibial spacer fitted onto a metallic tibial tray (b). The curved, top surface of the femoral implant rolls over the depressions in the tibial spacer.

The implants:

The metal parts of the implant are made of titanium-based (or cobalt/chromium-based) alloys. The plastic parts are made of ultra high molecular weight polyethylene.

The materials are chosen to meet certain criteria such as:

1. They must be biocompatible; that is, they can function in the body without creating either a local or a systemic rejection response.
2. Their mechanical properties must be able to duplicate the structures they are intended to replace; for example, they are strong enough to take weight-bearing loads, flexible enough to bear stress without breaking and able to move smoothly against each other as required.

3. They must be able to retain their strength and shape for a long time with minimal wear and tear.

So far, man-made joints have not solved the problem of wear and tear of the prostheses. When bone rubs against bone, or metal rubs against plastic, the friction creates microscopic particulate debris [56]. Just as wear in the natural joint contributed to the need for a replacement joint, wear in the prostheses may eventually require a second (revision) surgery. Knowing the distribution of contact pressure across this metal plastic interface can be crucial designing better implants and determining their survivability.

3.4 Knee joint surgery procedure:

The solution for severe osteoarthritis is to undergo total knee joint surgery replacing the affected bone surfaces with the metal and polymer implants. During surgery, the knee is kept in a bent position so that all the surfaces to be replaced can be exposed [57]. Usually, a lengthwise cut is made through the front of the knee that is 15 cm to 30 cm long. The muscles and the kneecap are carefully moved to the side to reveal the bone surfaces, without damaging the supporting tendons. After taking several measurements to ensure that the new implant will fit properly, the surgeon saws off and smoothes the rough edges of the bones. Usually, the surgeons begin with the femur and follow with the tibia. Special jigs provided by the implant manufacturer are used to accurately trim the damaged surfaces at the end of the femur. Several devices help to shape the end of the femur so it conforms to the inside of the prosthesis (femoral implant). The tibia is cut flat across the bone and a hole is drilled in the centre of the bone

(see Figure 3.5). The surgeon removes just enough of the bone so that when the prosthesis is inserted, it recreates the joint line at the same level as before surgery. This helps to maintain the length of the leg.

Figure 3.5. Shaved ends of femur and tibia.

(Source: www.kneesociety.org/totalKnee.cfm)

After the bones are shaped, the prostheses are inserted, tested and balanced. A sensor implant can help to make an informed judgment on the alignment at this stage of testing and balancing. The proposed sensor (moulded into the tibial implant) can be used to check the alignment of the femoral implant and can be removed before cementing in the tibial implant. The detailed procedure is described in Flow chart 3.1 where the use of the sensor is a part of the usual surgical steps.

Flow chart 3.1 (sensor requirement step for alignment during knee joint replacement surgery)

The diagram in Figure 3.6 shows the knee joint after a total knee joint replacement procedure. The patella button is also replaced if necessary. The stemmed tibial plate goes into the hole drilled in the tibia. The tibial spacer (polyethylene) is fitted onto the plate. The pressure mapping sensor developed in this research is built into the tibial spacer. The femoral component is cemented onto the femur before the knee is sutured.

Figure 3.6. The illustration shows the knee joint with the implants in place.

(Source: www.aaos.org)

3.5 Failure of a TKR procedure

The reasons for failure of total knee joint replacement can be as varied as implant loosening and soft tissue (muscles, ligaments, tendons) balancing [58]. One study of 279 revision surgeries [59] shows that 27% failed due to instability. A major contribution to instability is from implant malalignment. Another 8% failed due to patella-femoral problems. Patella-femoral problems arise after total knee arthroplasty mainly due to patella mal-tracking or the lack of smooth sliding of the patella button over the femoral implant [60] that can be caused by tibio-femoral malalignment. It is understood that proper alignment of the implants can also reduce the chances of complications such as varus-valgus that result in knock-knees or bow-legs [61].

Figure 3.7. Three tibial spacers with three sets of pressure patterns. (a) Uniform loading on both sides and evenly spread stress distribution in the centre. (b) The load is unequal on both sides and the contact regions are towards the edges. (c) The axis of loading is rotated due to which both grooves have contact regions at opposite ends.

To understand the effect of malalignment in the prosthetic knee, it is necessary to visualize alignment. Figure 3.7 shows three tibial spacers with pressure patterns (in blue

and pink), generated by femoral contact, showing situations that may be said to be alignment (a) and malalignment (b, c). The pattern in (a) shows similar stress distributions at the centre of the condylar grooves, (b) shows loading at the edges and the pattern (c) shows rotational malalignment. These situations are for illustration only. The actual picture can be different for every patient and figure (a) may not always be achievable in practice.

3.6 Factors influencing the longevity of the implants

Besides attributing failure to malalignment, there can be a combination of other factors such as contact stresses and polyethylene wear. Component positioning, restoration of leg alignment and polyethylene damage, all lead to early revision surgery of the knee. A better understanding of the factors contributing to early loosening and failure of TKR is needed to improve the longevity of the procedure. Many studies have concluded that the degradation of the polyethylene component is a major reason for failure. In one study, the prosthetic femoral and tibial components of eleven patients were retrieved during revision surgery after 6-26 months [62]. The reasons for revision included loosening and malpositioning and the prosthetic components were replaced during the revision surgery. It was observed that the damage to the tibial component was concentrated in two distinct regions on both the medial and lateral plateaus.

Total knee joint replacement surgery removes the damaged and painful areas of the femur and tibia. These areas are then replaced with specially designed metal and plastic parts. The tibia, with its plastic part and the femur with its metallic part form the

new knee joint [63]. The study of the contact pressure between the metallic femoral implant and the polyethylene tibial spacer is crucial to the design of the prostheses, which influence the long-term survival of the implants. Gomez et al. have reported the results of a clinical study of 193 total knee replacements (Insall-Burstein) over a range of 1-9 years. The analysis of survival after long term monitoring showed a 96.95% survival rate and complications due to loosening in 3.10% of cases [64]. Besides tibiofemoral contact pressure, the influence of contact alignment of the tibiofemoral joint has also been investigated [65]. In in-vitro biomechanical testing, the total contact areas were measured and the malalignment situations were simulated, on the application of a compressive load. Three different types of knee prostheses were used (Omnifit, Genesis and AMK implant models) in the study. The study shows the importance of contact alignment of the tibiofemoral joint of the prosthesis in in-vitro biomechanical testing. For conforming designs, even small malalignment (anterior-posterior, femoral rotation) can make the femoral component contact on the edge of the tibial spacer and induce stress concentrations.

The contact surface of the tibial spacer produces a local deformation when testing is performed [66]. After repeated testing, the geometrical change will increase the conformity of the tibiofemoral joint at the contact region. This is the reason why malalignment will result in larger total contact areas than normal contact alignment. It has been recommended that the loading condition should be selected properly to prevent plastic deformation of the polyethylene. {The compressive yield strength of UHMWPE (ultra high molecular weight polyethylene) is about 20 MPa.} The effect of alignment

during unicompartmental (half side of the knee) surgery has been reported [67]. It has been emphasized that tibiofemoral alignment is crucial to avoid post-operative complications.

The contact area and contact stress for both intact and prosthetic knees are essential for the evaluation of the joint's performance [68]. Excessive joint contact stress has been identified as an important factor in the pathogenesis of degenerative joint disease [69]. In a prosthetic knee, excessive contact stress causes surface damage and high wear rate of the UHMWPE leading to the failure of the knee implants. This would mean that a revision surgical procedure is required sooner than is normal.

Another factor affecting the stresses in tibiofemoral contact is the conformity of the contacting surfaces [70]. Three contemporary tibial configurations (flat-on-flat, curved-on-flat and curved-on-curved) have been analyzed in one study. Curved-on-flat identifies a pair of prostheses where the femoral surface is curved and the tibial spacer is flat, thus generating curved-on-flat contact. It has been observed that the peak contact stress, von Mises stress, and von Mises strain were lowest for the curved-on-curved model. The von Mises effect is a generalized response to stress observed in ductile materials. Von Mises stress reduces the six dimensional strain tensor to a single number (a scalar) for calculating yield criteria. Due to this effect, a load applied on the top surface of the spacer (or similar material) concentrates into higher stresses just beneath the surface of the spacer. It has been reported that a curved indenter produces a maximum effective stress (e.g. von Mises stress) 1-2 mm beneath the centre of contact [70]. Also, it

is understood that for curved profiles, the contact is initiated at a point and the contact region expands as the compressive load is increased.

These are some of the major factors that influence the degradation of the tibial spacer leading to revision surgery of the knee joint. Proper loading and alignment of implants can delay the requirement of revision surgery.

3.7 Pressure mapping systems for the knee joint

Most of the above conclusions have been reported using available pressure mapping systems at the tibiofemoral interface. Pressure mapping is the simultaneous measurement of pressure and position. In other words, a pressure map gives the distribution of pressure across a certain area. It measures the magnitude of pressure and the location of pressure. Various instrumentation such as the Tekscan sensor [71], Fuji pressure sensitive film [72], dye injection [73] and silicone rubber [74] have been used to study the contact area and the contact pressure in the tibiofemoral joint and the prosthetic components. The knowledge of these parameters helps in the design of better prosthetics and proper alignment of implant during surgery to make the total joint replacement procedure last longer. The two popular methods of pressure mapping are the Fuji film and the Tekscan sensors.

Fuji pressure sensitive film (Prescale) is a popular method for the measurement of contact area and pressure. The working principle of the pressure film is shown in Figure 3.8. However, this method involves the insertion of a sensor in between the two

contacting surfaces, which will change the original contact behavior [75]. Also, it is unable to detect any plastic deformation occurring in the UHMWPE due to repeated measurements [76]. Finite element analysis, along with complementary experiments, has been used to conclude that there is an error in the estimation of the contact area, and hence the pressure, of up to 77%. However, this error in estimation of area is dependent on the load applied and the loading condition [76].

Figure 3.8. Working principle of the Fuji pressure sensitive film. In case of the two-sheet type pressure film (a), the colour forming layer is sandwiched inside whereas for the mono-sheet type film (b), the colour forming layer is exposed.

Red patches appear on Fuji Prescale when pressure is applied. The colour density changes according to pressure level. Five types of Prescale are available to measure a pressure range of 0.2 – 130 MPa. (Source: <http://www.prescale.com/E/E.htm>)

The film is thin and sensitive. However, the inherent drawbacks are that a different grade film is required for a change in the range of pressure, and the measurement is static. Any minor change in position would lead to errors in the estimation of pressure and contact region.

The K-Scan (Tekscan) sensor is another system that is frequently used at the knee joint to measure the contact pressure [77]. The sensor uses piezoresistive strips to measure pressure distribution over a grid of small elements. The pressure on each element is assumed to be constant and equal to the pressure measured at the centre where the piezoresistive strips cross. It is difficult to predict the error made by this assumption.

Figure 3.9. Picture of a K-Scan sensor (a) and method of insertion at the tibiofemoral interface (b).

The K-Scan System by Tekscan is a pressure measurement system using thin-film pressure sensing technology. Figure 3.9 shows a system used to measure tibiofemoral pressure. The grey regions bear the sensing elements and are introduced on top of the condylar grooves of the tibial spacer. The two sensing regions are for the two condylar grooves. It can measure pressures as low as 1 KPa and as high as 175 MPa. Its capability lies in the density of sensing elements with 60-100 sensors/cm².

(Source: http://www.tekscan.com/medical/specs_kscan.html).

The K-scan system is thin, but due to its finite thickness, it may alter the contact area and pressure measurements by altering the topography. It is affected by temperature and needs to be calibrated at the temperature used during measurements. It is solid, and though it flexes very well about a single axis, it may crimp around dual axes with very high radii of curvature [78].

Hochmann *et al.* have carried out a comparative study of various pressure mapping systems [79]. Their study concludes that the various systems have different parameters influencing the measurements and so the systems cannot be standardized to give a direct comparison. However, all of the pressure mapping systems for the knee joint would still require insertion between the tibio-femoral interfaces and affect the actual distribution of pressure. Fuji pressure film is the thinnest; but the thinnest version is meant for the lowest magnitude of pressure. Recently, another sensor has been reported that can be placed under the tibial tray and can estimate forces [80]. It works on the

principle of telemetry and may be useful for measuring absolute loads. It cannot give the distribution of the load across the condyles.

3.8 Motivation for an alternative sensor

Besides individual advantages and disadvantages of the pressure mapping sensors currently used for the study of tibiofemoral pressure distribution, the following two critical points are the motivation for designing this sensor. These two reasons are not defined by the author. It has been raised in several published journal papers [75, 78].

1. The insertion of a sensor in between the two contacting surfaces will change the original contact behavior (Fuji film and Tekscan).
2. The sensors do not take into account the response of the polymer surface (Fuji film and Tekscan).

These points may be understood from the following diagram. Figure 3.10 demonstrates the effect of inserting a sensor between the articulating surfaces of the femur and tibia. One of them (a) does not conform completely to the curvatures whereas the other (b) does conform. In both cases, the presence of the intermediate layer (sensor) of finite thickness changes the contact characteristics. However, a sensor that lies beneath the articulating surface of the tibia (c) can give the correct contact behaviour. Also, the sensors placed on top of the tibial spacer will give the contact response between the femoral implant and the sensor itself. The proposed sensor can give the response of the polymer surface in contact with the femoral implant. This information is useful in

predicting wear patterns because the impact of a load and the actual stress distribution can be different.

Figure 3.10. Pictorial representation of inserting a sensor between the articulating surfaces of the femur and tibia. (a) The sensor does not conform completely and may create gaps at the interface. (b) A surface-mounted, conforming sensor of finite thickness may wrongly estimate the region of contact. (c) A sensor lying beneath the contact surface will not change the contact distribution.

The proposed sensor can predict the actual distribution of pressure as it lies beneath the contacting surface. Also, it estimates the response of the polymer surface,

which is the actual cause of the degradation of the tibial implant. In this study we have assumed that PMMA is a suitable substitute for UHMWPE, which is the implant material in many cases; but the intrinsic sensor can be inserted into any similar material.

3.9 Singular advantage

The procedure used to align the prostheses relies on visual confirmation by the surgeon. The arbitrary nature of the procedure may produce slightly different anatomical orientations for each case [81] (later proved in Chapter 7). For example, the vertical axes of the leg can tilt towards the bodyline or away from it. These deviations in the loading configuration have an effect on the freedom-of-motion region. This is a reason why quantitative differences in the stress patterns are possible. For example [66], when the maximum von Mises stress at the knee joint is 16 MPa in neutral position, a 5° varus tilt can result in maximum von Mises stress of 28 MPa and quicken the wear of the tibial spacer.

Currently, there is no sensor that can guide a surgeon during the alignment of the implants. The proposed sensor can detect malalignment and incorrect orientation before the implants are cemented and the knee sutured. This can be done by referring to pressure maps displayed by the sensor on a computer. The best judge of a medical procedure is the surgeon; however, the sensor can help to reduce the chances of human error.

Chapter 4

Design and development of the sensor

The fundamentals of a fibre optic sensor and the measurement of tibiofemoral pressure have been discussed in the previous chapters. This chapter will present the design and development of the sensor. The development of the sensor involves choice of materials, making the optical sensor, packaging it and modifying it for this specific application. It also includes all preliminary tests carried out as proof-of-principle to finalize the actual design of the sensor. The sensor would comprise of fibre gratings embedded in fibre-reinforced composite and moulded into the tibial spacer. The actual design finalized during experimentation will be discussed in subsequent parts of this chapter. The guiding principle for the sensor development is the fact that it should be easy to use for the surgeons without involving much deviation from the actual surgical procedure of total knee replacement.

4.1 Biomaterials

The prospective use of this sensor is in in-vivo pressure-mapping during total knee joint replacement surgery. This is the reason for an appropriate selection of “biomaterials” for designing the sensor.

Optical fibres, made of silicon dioxide and other dopants, have been used reliably in various forms of endoscopy. They have been found to be safe for use inside the human

body due to their inert nature. This property, combined with its sensing capability, makes the optical fibre a suitable choice as the sensing medium.

Fibre-reinforced composites are being considered for prosthetics [82]. They have high mechanical strength and are available in a variety of configurations and material combinations. Here we choose glass/epoxy as it is biocompatible.

Silicone rubber (or silicone) is used for implants in the body and it forms the main component in many forms of life-supporting medical equipment such as pipes and tubes. The material provides flexibility and can be used for the sensor.

PMMA is commonly used by dentists and orthopaedics as a cementing material. It is to be noted that though most tibial spacers are made of UHMWPE, PMMA is a suitable alternative that can be moulded and set under normal conditions.

These materials form different parts of the sensor and each one has a significant role in influencing the properties of the sensor. The following sections give the details of the methods of incorporating these materials.

4.2 Grating inscription

The first stage of the development of the sensor begins with the inscription of the fibre grating. The chirped fibre gratings (and all subsequent gratings) were written on hydrogen loaded fibre. Ordinary telecommunication fibre was loaded with hydrogen in a

chamber at 10.3 MPa for 3 days at 70 °C. Gratings were inscribed with a frequency-doubled Argon-ion laser (Coherent) at 244 nm with the phase mask technique. The phase masks (Ibsen, Stocker Yale) used have chirp rates of 2.25 nm/cm and 2.18 nm/cm, with lengths of 6 cm and 3 cm respectively. The laser power used for the inscription was 60-70 mW. The spot size of the beam was 2 mm. The laser was allowed to translate over the mask at speeds of 0.01-0.03 mm/second. The gratings were then pre-annealed at 90 °C for 5 hours to arrest the decay of reflectivity with time.

4.3 The chirped grating sensor

Figure 4.1. Principle of reflection of wavelengths by a chirped grating. Different regions along the grating reflect different wavelengths.

Figure 4.1 shows the principle of a chirped fibre grating. The grating reflects a continuous range of wavelengths i.e. the shortest period reflects the shortest wavelength while the longer periods reflect longer wavelengths. In other words, each point on the reflected spectral bandwidth is an image of each point in the grating. This concept can be

used for distributed or multipoint sensing. The change experienced by a particular point along the grating affects the periodicity at that point and hence it brings a change in the spectrum at the corresponding wavelength.

Initially, in this project, bare fibre gratings were tested to study their response to pressure. The bare fibre, after exposure to UV radiation, is very brittle. It is also sensitive to fluctuations of temperature [83]. Touching the fibre can change the reflection spectra due to body temperature. To overcome these factors, the grating has to be protected in a suitable material for use as a pressure sensor. Embedding it in a stack of glass-epoxy prepregs (defined below) in longitudinal and transverse directions has been chosen as a suitable improvement for the tibial sensor.

4.4 Glass/epoxy composite

The glass/epoxy composite is available as sheets of unidirectional prepreg. Prepreg is the term assigned to a single sheet of fibre-reinforced composite as each sheet is made from bundles of fibres held together (or “pre-impregnated”) by epoxy. The prepreg can be cut into the desired shape and stacked. The layering of the prepregs determines the overall flexibility of a stack. In other words, the angles at which the layers are oriented with respect to each other determine the flexural modulus of the stack/laminate. The prepreg can be moulded in any shape for curing. This property can be used to obtain difficult shapes that may be impossible without sophisticated and expensive 3-D machining. The fibre-reinforced composite also imparts high mechanical strength to the optical fibre sensor. Also, it has been reported that longitudinal embedding

in unidirectional fibre-reinforced composite reduces birefringence effects of the fibre grating [84]. This composite used for the sensor was chosen for the above reasons. Besides, the two other practical reasons were that it was easily available in the laboratory and previous tests had shown promising results with other similar prepreg [85].

4.5 Prepreg design with a chirped grating

This section presents the first design of the sensor, which is only a fibre grating embedded in fibre-reinforced composite at this stage. Due to the reasons given below, the prepreg was stacked as shown in Figure 4.2 (a). A picture of the sensor can be seen in Figure 4.2 (b). The two cross or transverse layers at the top were used to enhance the spatial resolution of the sensor. It is essential to have spatial resolution of the stress to obtain an accurate pressure map. For example, a 3 MPa region of stress and a 3.2 MPa region lying 5 mm apart on the sensor should give different responses at different wavelengths in the reflected spectrum. The cross layers exploit the linear chirp of the grating for determining the location of the load. As explained earlier, the reflected wavelength is a linear function of the position. The longitudinal layer (parallel orientation) is not easily deformed along the direction of the fibre reinforcements. When the deforming force is lifted, the longitudinal layers relax to their original dimension and the strain experienced by the fibre grating is removed. This causes the sensor to lose the effect of the force quickly.

(b)

Figure 4.2. (a) Stacking design of prepreg layers. There are two cross layers on top and two parallel layers below. The optical fibre lies between the third and fourth longitudinal layers. (b) Picture of the embedded fibre grating sensor.

Also, it may be noted that the optical fibre (between the third and fourth layers) is away from the neutral layer of the stack. This factor is consistent with all subsequent designs as well. This translates a compressive stress (load) applied from the top into an axial tensile strain in the optical fibre, thus lengthening the grating periodicity. However, when the sensor is flipped and a compressive stress is applied to the one-layer side, the

optical fibre experiences a compressive strain and hence shortening the grating periodicity.

The physical considerations for the design are logical. However, the mathematical aspect can be understood from a definite case study presented in Chapter 6. It may be said that the Young's modulus of a stack, along longitudinal and transverse axes, is different from that of the single layer of prepreg (see Appendix 3).

The stack of prepreg with the optical fibre was sandwiched between two metal plates and cured in a vacuum oven at 120 °C for 1 hour 20 minutes. One of the metal plates was wrapped in non-porous Teflon cloth and the other plate was wrapped in three layers of non-porous Teflon, cotton wool gauge and porous Teflon. This method of curing ensures that no impurities or air gaps remain in the stack.

4.6 Experimental setup

The basic experimental setup for testing this sensor and all subsequent sensors is the same. A tunable laser source (Ando TLS), an optical spectrum analyzer (Ando OSA), a circulator (C) and a digital force gauge (Chatillon DFS) were used as shown in the configuration below (Figure 4.3). The load applied to the sensor is varied with the vertical force gauge.

Figure 4.3. Schematic diagram of the experimental setup used for the interrogation of an FBG sensor. Optical power from a tunable laser source (TLS) is coupled through a circulator (C) into the fibre sensor. The reflected power passes through the circulator and is measured by the optical spectrum analyzer (OSA). The digital force gauge (DFS) is used to apply the load on the sensor.

1. Tunable Laser Source:

The tunable laser source was used as input for the fibre Bragg gratings. The tunability allows the selection of the wavelength of operation (1520 nm – 1580 nm) and the input power, which was set at 0 dBm or 1 mW for all experiments.

2. Optical Spectrum analyzer:

The spectrum analyzer was synchronized with the tunable laser source for all measurements and for monitoring FBGs during inscription. The span, centre

wavelength and reference level and other parameters can be changed on the OSA for the measurements. For monitoring a single peak, a 3 dB threshold was used to identify the central wavelength, whereas the wavelength division multiplexing (WDM) analysis or peak search was used for simultaneous monitoring of several peaks. An Anritsu OSA was used during the cadaveric tests due to portability of the system.

3. Three port circulator:

The unidirectional circulator (C) was used to route the input optical signal from the source to the FBG and then the reflected signal from the FBG to the spectrum analyzer. The circulator is a coupler that routes the power in a single direction and lowers the loss of power due to reflection into the input channel.

4. Digital Force Gauge:

The force gauge (DFS) has a force sensitivity of 0.5 N and a range up to 500 N. The gauge was a part of a setup that can be used to mount the sensor and move the gauge up and down, to increase or decrease the applied stress.

4.7 Response of the chirped grating sensor

The preliminary tests of this sensor included two steps. An increasing force was applied at one point along the length of the embedded sensor shown in the picture (Figure 4.2 (b)) and the change in the spectrum was monitored, in the first part. The response is shown in Figure 4.4. In the second part, a force of fixed magnitude was applied at various points along the length of the grating. The response is shown in Figure 4.5.

The effect of pressure on a chirped fibre grating can be observed from the change in spectral shape. The reflected spectrum shows peaks and dips as the reflectivity of the grating changes at a certain wavelength.

Figure 4.4 shows the spectra of the sensor when the force is increased at a single point on the grating. The peak-to-dip reflectivity change increases as the force applied increases from 0 to 12 N.

Figure 4.4. Reflected spectra of a chirped grating with increasing loads applied at a specific point along the sensor. The loads identifying the spectra are shown on the right side scale. The numbers (00, 01, 02, 03) are various traces recorded at different intervals of time.

Figure 4.5 shows the spectra of the sensor when a 10 N force was applied at different points on the grating. As the point of application is shifted (3 mm), the position of the dip also shifts linearly with wavelength.

Figure 4.5. Reflected spectra of the chirped grating with a load of constant magnitude applied at different points along the sensor. The numbers (00, 01, 02, 03, 04) are various traces recorded at different intervals of time.

Figure 4.6 shows the plot of point of application of the force along the grating and the wavelength where the dip is observed. The tests were repeated with forces of 10 N and 15 N. The wavelength change is linear with the position for both values of the applied force. The dip in the 15 N data may be due to some air pocket in the prepreg layers at that position resulting in the inaccuracy.

Figure 4.6. Variation in dip position as the load is moved along the length of the sensor.

The two sets of data were measured for two different loads.

Figure 4.7 shows the plot of the force applied and the peak-to-dip change observed. The two plots represent the effect at two different points on the grating (5 mm apart). The reflectivity change was observed to increase monotonically with applied force.

Figure 4.7. Change in peak-to-dip height in reflectivity with increasing load, measured at two different points along the sensor. The effect was observed at two different points of loading. Error bars of ± 0.5 dBm has been used due to the resolution of the scale used during measurement.

The reasons for observing these changes are as follows. The chirped grating can be thought of as a continuous chain of smaller sub-gratings, each having a different period and hence reflecting a particular wavelength. When a force is applied at one point of the chirped grating, the period of the smaller grating experiencing the force changes, which means that the small grating will reflect the next higher wavelength. So there will be a dip (corresponding to the original period) and a peak (corresponding to the strained period) in the reflected spectra.

The above tests were repeated with different combinations of embedding for confirmation. The results show that the chirped fibre grating has potential for use as a distributed sensor. Embedding the bare fibre grating with transverse layers of the

composite on either side of the optical fibre leaves a large ripple in the spectrum after curing. This is because, after curing, the cross ply glass fibres create a permanent strain (impression) on the optical fibre, which is seen as a ripple in the reflection spectra. Also, the force creeps into the design, slowly, as the lower layer supports the deformation caused by the force on the upper layers. This creep is observed as the dip and peak remaining in the spectrum for a longer time after each repetitive loading.

This result of observing peaks and dips can be corroborated using the commercial software IFO Gratings from Optiwave. The software allows the construction of a chirped grating of a definite length and chirp rate. A specific magnitude and profile of strain can also be applied. A simple simulation shows that when a chirped grating is subject to a point load, there will be peaks and dips. This is shown in Figure 4.8. In this case, a grating length of 30000 μm was assigned and Gaussian stress of peak 500 μstrain and width 0.05 was used to generate the figure.

Figure 4.8. Simulated reflected spectrum of a chirped grating showing a peak and a dip when a load is applied.

The drawbacks of this sensor are as follows.

1. The sensor range saturates above 15 N, which is not desirable for this application.
This was observed as there was no increase in the peak-to dip height when the load was increased from 20 N to 30 N.
2. The edges of the spectrum show wavelength shift instead of peaks and dips, which would lead to errors in quantifying the effect of loads on the basis of peak-to-dip heights.

This approach of quantifying the load in terms of reflectivity change is also subject to the absolute optical power guided in the fibre. It has been mentioned above that the underlying explanation for the peaks and dips is a shift in wavelength caused by a change in the grating period. To exploit this fact, an alternative technology was tried that would be based on the principle of wavelength shift and have the advantage of chirp or position information as well. A high-density multiplex of gratings on a single fibre can fulfill such a requirement.

4.8 The sampled chirped grating

A simple method to obtain a multiplex of fibre gratings in a small region (a couple of centimetres) is to sample a chirped grating. Inscribing a sampled chirped fibre grating onto the optical fibre is the first step in developing the sampled chirped grating sensor. The sampled chirped grating provides the advantage of position information, as the reflected wavelength is a linear function of the chirp. The magnitude of the load is

obtained from the wavelength shift. The sampled chirped grating is preferred as the quantification of the load is obtained from the wavelength shift as compared to reflectivity change in the case of chirped grating.

Multiplexing many fibre Bragg gratings (FBGs) to have many point sensors in a small area (say, a few centimetres) of a fibre segment is difficult using the phase mask technique of inscription. This is because of the need to inscribe many small gratings a few millimetres long, which may compromise the strength of the reflectivity. Besides, the separation of adjacent gratings has to be precisely defined on a transparent bare fibre to prevent them from being overwritten. In addition, different phase masks would be required for inscription of each grating, which makes it costly. However, the use of a sampled grating will allow multiplexing of many point sensors in a small region. It should be noted that a uniform sampled grating or superstructure grating would require many parameters to be precisely controlled to obtain many spectral peaks of equal reflectivity [86].

Figure 4.9 shows the reflected spectrum of a single sampled chirped grating. There are ten sub-gratings with wavelength spacing of approximately 2 nm. The reflectivities of all the peaks are similar and lie within 3 dB of each other. The individual reflectivity for all sub-gratings was higher than 99.9%. The bandwidths of individual sub-gratings are around 0.4 nm.

Figure 4.9. Reflected spectrum of a sampled chirped grating with ten sub-gratings. Each peak in the spectrum corresponds to one sub-grating.

The fabrication of many point sensors within a short grating length of several centimetres using a single chirped phase mask has the advantages of providing many spectral peaks of equal reflectivity with repeatability, desired spatial resolution and convenience in inscription.

To fabricate the sampled chirped grating sensor, a specific design was inscribed with sub-gratings that were 2 mm in length, and gaps of 4 mm were introduced between adjacent sub-gratings. The laser beam was allowed to translate selectively along the phase mask. This configuration was chosen so that the available phase mask could be

used to generate maximum density of sub-gratings without any overlap in the reflected spectrum.

Figure 4.10. Principle of selective inscription of a sampled chirped grating from a linearly chirped phase mask. Each section is identified by a specific spatial position and has a different grating period.

Figure 4.10 shows the principle of a sampled chirped grating. Several short regions of the optical fibre are inscribed with different periodicities of gratings. These gratings reflect different wavelengths giving rise to separate peaks in the spectrum.

4.9 The embedded sampled chirped grating

The sampled chirped grating was embedded in a stack of prepreg in the configuration 90/0/0/F/0. This representation denotes the position of the optical fibre in the stack and the relative orientations of the four layers of prepreg with respect to the direction of the optical fibre. The top layer is at 90 degrees whereas all other layers are at zero degrees or parallel to the optical fibre. Figure 4.11 illustrates the design. This specific design (the parallel layers) supports axial deformation of the fibre grating. In the case of a parallel array of fibres, the single cross layer at the top spreads the force

partially in the transverse direction as well. The chapter on numerical analysis illustrates the effect of the cross layer; due to the cross layer, the tensile modulus of the stack along the transverse direction is increased, as compared to an entirely longitudinal stack.

Figure 4.11. Stacking of the prepreg layers for a sampled chirped grating array. The top layer is transverse to the optical fibre. The three lower layers are parallel to the optical fibre.

4.10 Curing the curved array

The above sensors were flat and were used to verify a few fundamental properties of the sensor. However, the condylar groove of the tibial spacer is a complicated structure with the radii of curvature varying in all directions. It is impossible to obtain a simplified mould to cure a congruent sensor array. A simple alternative was to design a sample of plaster of Paris to complement the tibial spacer. The picture in Figure 4.12 shows the plaster of Paris sample and a tibial spacer sample, layered with Teflon cloth, that were used to sandwich the prepreg stack. The cured sensor conforms to the curves of the tibial spacer (Figure 4.13).

Figure 4.12. The tibial spacer sample (top) and the plaster of Paris complement mould (bottom) covered with Teflon layers to cure the fibre-reinforced composite.

Figure 4.13. Schematic figure of the curved, embedded fibre grating array.

The layout of the fibre gratings in the sensor is shown in Figure 4.14. Due to limitations in the range of phase masks, 4-5 sub-gratings were inscribed on one fibre. Both edges of each of the condylar grooves had sampled gratings of 2 cm length with four sub-gratings. The dimensions of the tibial spacer governed this design as the arc length at the centre is about 3 mm whereas the arc length at the edges is about 2.6 mm. This resulted in a total of 46 sub-gratings in the two grooves of the tibial spacer.

Figure 4.14. Layout of the sampled fibre gratings in the two condylar grooves of the tibial spacer (Figure not to scale). Each dot denotes the position of one sub-grating.

4.11 Moulding the tibial spacer

The tibial spacer for this sensor was moulded in the laboratory with PMMA (commercial name: Orthoresin). The moulding of the tibial spacer was carried out with the expertise and help of a technician at the National Dental Centre, Singapore. A rubber mould was made from a commercially available sample of tibial spacer (Figure 4.15). The mould was then used to cast a number of samples. PMMA powder and clear liquid PMMA were mixed in the ratio of 2:1 by volume. The mixture was immediately spooned into the rubber cast and allowed to set at room temperature.

(a)

(b)

Figure 4.15. The two halves of the rubber mould used to cast the samples of tibial spacer. The PMMA mixture is poured into one half (a) and covered with the other half (b).

This process of producing tibial spacer samples in the laboratory provides the flexibility of having many samples for testing and is also cheaper than procuring the commercial ones. At one stage, grooves were built into the spacer, during moulding, to explore a different layout of the gratings.

4.12 The tibial spacer sensor

The final design of the sensor was the culmination of many considerations. The pressure-mapping sensor was designed to cover the entire area of the condylar groove, on both sides. The moulded PMMA spacer groove was covered with a thin layer of silicone rubber (in the grooves). The rubber has a lower Young's modulus as compared to PMMA and allows for greater flexion of the composite sensor. This layer provides greater sensitivity without saturating the linear range of the sensor. The embedded fibre-

reinforced composite sensor was stuck onto the silicone rubber, before the top layer of the rubber dried. The sensor was loaded and allowed to cure overnight. Then a mixture of PMMA was prepared with equal parts of powder and liquid Orthoresin to obtain a highly viscous fluid. This fluid was then poured over the sensor to coat the composite sensor and the edges (forms the top layer as shown in Figure 4.16). This process ensures that the composite with the optical fibre and the rubber are all encased in PMMA (Figure 4.16). This factor takes care of the in-vivo issues of biomedical safety to a large extent.

Figure 4.16. Vertical cross section of the tibial spacer sensor showing the various layers in the design (Figure not to scale).

The optical fibres that remained protruding out of the sensor were inserted into standard heat shrink tubing (see Figure 4.17). Fibre optic connectors were spliced on to the ends of these fibres to allow for interrogation with a spectrum analyzer and then the tibial spacer sensor was ready for laboratory testing.

Figure 4.17. The photograph shows one of the sensors in the process of being moulded. The two grooves display the embedded gratings.

This chapter follows the development of the tibial spacer sensor through the various stages of experimental design, starting with the inscription of the sampled chirped grating; the final sensor is an optimization of the properties of the various materials to obtain a pressure mapping sensor for the knee joint. The influence of the materials and their configuration can also be understood in the next two chapters on experimental results and theoretical modeling.

Chapter 5

Results of laboratory testing

A certain sequence of testing was followed before the sensor could be approved as suitable for in-vivo testing. The first step involved laboratory calibration to determine sensitivity, hysteresis, effect of temperature and array behaviour. The next step was cadaveric testing to determine response of the pressure map for various loads, angles of flexion and simulated conditions of malalignment. The first step will be discussed in this chapter while the second step will be the subject of a subsequent chapter (Chapter 7). In this chapter, the results are presented for the flat composite embedded sensors and then for the final design - the tibial spacer sensor.

5.1 Preliminary tests

Before carrying out systematic and rigorous testing of the sensor, some simple tests were done for proof of concept. A pressure mapping sensor should have the following properties.

1. The wavelength shift of each sub-grating should increase with the force applied on it to allow the quantification of force.
2. The location of the sub-gratings should be a linear function of the length so that the location of the load can be determined. It would also simplify the creation of an array and subsequent mapping. This property should be a direct consequence of using a

linear phase mask. However, any changes during grating inscription or embedding should be tested.

3. To ensure that different forces applied to the sensor are resolved spatially, it has to be tested that when two forces, a few millimetres apart, are applied along the length of the grating, two different regions of the spectrum register wavelength shift.

These tests help to build the background for a pressure-mapping sensor that can give the distribution of the pressure along with the position information.

Figures 5.1 and 5.2 show the effect of a 4 N force on a single (fourth) sub-grating and the effect when the force is increased to 8 N. The wavelength shift, of the affected sub-grating, increases as the force increases. The double traces in both plots are the reference (red) and shifted (green) spectra, which provide easy comparison.

Figure 5.1. Reflected spectra showing the effect of a 4 N force on one sub-grating. The trace of the shifted spectrum is in green colour and the reference spectrum is in red colour. It can be seen that the central wavelength of the fourth sub-grating shifts.

Figure 5.2. Reflected spectra showing the effect of an 8 N force on the sub-grating. The trace of the shifted spectrum is in green colour and the reference spectrum is in red colour. The fourth sub-grating shows greater wavelength shift as compared to the shift observed in Figure 5.1.

A simple simulation with IFO Gratings (Optiwave) software also endorses this proof-of-concept experiment. The software allows the design of a sampled chirped grating with optional sub-grating length and sub-grating central wavelengths. Strain of various magnitudes can be applied directly to the sub-gratings. The simulation in Figure 5.3 shows a reference spectrum of a sampled grating and the shifted spectrum of a strained grating. The wavelength shift is obvious in the case of the fourth sub-grating.

Though the software allows for a simulation using bare fibre gratings (no embedding), it verifies the principle of wavelength shift caused by strain, for a sampled

chirped grating. In this case, grating lengths of $2000\text{ }\mu\text{m}$ were used with phase shifts of $4000\text{ }\mu\text{m}$ in between the sub-gratings. A uniform stress of $200\text{ }\mu\text{strain}$ was applied to the fourth sub-grating to simulate the wavelength shift. This simulation gives an estimate of the strain that causes a wavelength shift of $\sim 200\text{ pm}$ and can be used to predict the strain in Figure 5.1 ($\sim 200\text{ }\mu\text{strain}$) and 5.2 ($\sim 450\text{ }\mu\text{strain}$). However, an accurate comparison is not possible as the software does not include the effect of embedding.

Figure 5.3. Simulated reflected spectra of a sampled chirped grating showing the unstrained (a) and strained (b) gratings. The fourth sub-grating shows central wavelength shift.

Figure 5.4 shows the wavelength shift with the force applied with a “ ω ” shaped tool. In this case the femoral implant has been used to load the flat sensor. The spectra can be seen to shift at two different points as the tool makes contact at two points along the sensor. These points are subjected to pressure and it is evident from the response of

the sensor. Also, it can be seen from the unsymmetrical spectra that the force is not distributed uniformly in both regions. As the two halves make contact with the sensor at different regions, the affected regions show wavelength shifting whereas the unaffected region in the centre between the contact points and beyond the contact points show no wavelength shift.

Figure 5.4. Reflected spectra showing the effect of simultaneous loading at two different points along the grating. The reference spectrum is in red colour and the shifted spectrum is in blue colour. It can be seen that the second, third and eighth sub-gratings show shift in their central wavelength. The diagram in the inset (right) shows the loading configuration with the “w” shaped tool on the flat sensor.

5.2 Response of a single sampled grating

A single sampled chirped grating was embedded in fibre-reinforced composite (glass/epoxy) in the configuration 0/0/F/0. The grating contains 10 sub-gratings, each of them 2 mm long and spaced 4 mm apart. The position of the sub-gratings in the embedded sensor can be easily determined simply by pressing different points along the sensor. As the finger is moved down the optical fibre, the sub-grating coming into contact shows wavelength shift. This shift is easily monitored with an optical spectrum analyzer. A systematic study was done using a force gauge tool of known area with the sensor placed on a solid metallic rack. Increasing force was applied to individual gratings and the change in wavelength shift was observed. The sub-gratings show similar response with the plots closely following each other.

The minor variation in sensitivities is due to the location of the point of contact of the force gauge tool. It will be seen in subsequent parts of this chapter that the distance, of the point of application of the force from the sub-grating influences the wavelength shift caused by the force. Figure 5.5 shows the response of five different sub-gratings. All the plots show monotonically increasing but non-linear trends. The wavelength shifts saturate at very low pressures.

Figure 5.5. Variation of wavelength shift of five sub-gratings with increasing pressure. Each sub-grating was loaded separately, one at a time. The legend identifies the reference central wavelengths of the sub-gratings.

However, in Figure 5.6, a few (4) layers of masking tape were placed below the sensor and a single sub-grating was monitored. A significant improvement in the linearity is observed although it is combined with a fall in sensitivity. The reason for this improvement may be that the softer material below the sensor allows the sensor to flex easily and prevents the wavelength shift from saturating at low loads. The straight line is a linear least-squares fit to the data and lies with the range of the error bars. Expected peak pressures in total knee joint surgery would be around 10 MPa. So, the conclusion was drawn that a layer of soft material below the embedded grating is necessary to ensure good linearity between the force applied and the wavelength shift. The error bars used are 0.5 N (0.25 MPa) for the x-axis and 15 pm for the y-axis. It may be noted that 0.5 N is the resolution (least count) of the digital force gauge and 15 pm is the accuracy of the spectrum analyzer.

Figure 5.6. Variation of wavelength shift of a single sub-grating with applied pressure.

The data is plotted with a linear fit passing through (0,0).

5.3 Determination of the location of the force

The response of a pressure mapping sensor that is embedded in fibre-reinforced composite can be determined from the bending experienced by the sensor when pressure is applied. A point pressure will flex the sensor, which results in compression or tension experienced by the fibre grating. Depending on the magnitude of the applied pressure and the design of the tool used, the adjacent gratings may or may not be affected. The number of sub-gratings showing a wavelength shift and their magnitude can be used to determine the spread of the force. Besides, when the point of application of the force is not directly on any sub-grating but in between two sub-gratings, the magnitude of wavelength shift experienced by either sub-grating will depend on its proximity to the point of application of the force. The distance of the sub-grating from the point of application of the force is

inversely proportional to the wavelength shift. This concept is valid assuming that the wavelength shift is linearly dependent on the applied force (Figure 5.6) and that the chirp of the grating is linear. The following results support the above concept.

Figure 5.7 illustrates the experimental procedure used to determine the location of the pressure when it is not acting directly on any sub-grating. The black regions denote four consecutive sub-gratings and the arrow denotes the load. P1 and P2 are two different loading points between adjacent sub-gratings. The first loading point P1 is at a distance x from the left sub-grating and at a distance y from the right sub-grating.

Figure 5.7. Schematic diagram explaining the method used to determine the point of application of a load. The two adjacent sub-gratings are denoted by 1 and 2. P1 and P2 are two different points of loading between sub-gratings 1 and 2.

It is observed that the wavelength shift resulting from the force is inversely proportional to the distance from the location of the force. In this case, the sub-grating on the left of P1 will show higher wavelength shift than the sub-grating on the right as x is

smaller than y . Applying the ratio of inverse proportions, the point of application can be obtained from the simple calculation given below.

Let $\Delta\lambda_{\max 1}$ be the highest wavelength shift observed by a sub-grating 1 and $\lambda_{\max 1}$ be the corresponding central wavelength of that sub-grating, when loaded at a certain point P. Let $\Delta\lambda_{\max 2}$ and $\Delta\lambda_{\max 3}$ be the wavelength shifts of sub-gratings (2 and 3) on either side of sub-grating 1, such that $\Delta\lambda_{\max 3} < \Delta\lambda_{\max 2}$. Also, the central wavelength of sub-grating 2 is denoted by $\lambda_{\max 2}$. The location of the applied force will be between sub-gratings 1 and 2 and can be determined as follows.

$$Location = \lambda_{\max 1} \pm \left[\frac{|\lambda_{\max 1} - \lambda_{\max 2}|}{\Delta\lambda_{\max 1} + \Delta\lambda_{\max 2}} \times \Delta\lambda_{\max 2} \right] \quad (5.1)$$

This equation is based on the fact that the central wavelength of the sub-gratings in the spectrum has a one-to-one correspondence with the spatial position along the optical fibre and $\lambda_{\max 1} - \lambda_{\max 2}$ is equivalent to $x+y$ of Figure 5.7. In the above algorithm, a negative sign applies to the case of $\lambda_{\max 2} < \lambda_{\max 1}$ and the sign is positive when $\lambda_{\max 2} > \lambda_{\max 1}$.

However, if $\Delta\lambda_{\max 2} = \Delta\lambda_{\max 3}$, the location is $\lambda_{\max 1}$. So, when the magnitudes of the wavelength shifts, and correspondingly the force, are uniformly distributed on either side of a sub-grating, then the force is acting directly upon it.

Figure 5.8 shows the wavelength shift experienced by four consecutive sub-gratings, with increasing force, when loaded with a hemispherical shaped point tool. Four sub-gratings in the sensor are affected due to the shape of the tool in the force range

of 5 N to 15 N. The point of application of force is an arbitrary point p1 (analogous to P2 of Figure 5.7) for the first three series. A comparison of the relative wavelength shift can provide an estimate of the location of the force. Another location of the force p2 (analogous to P1 of Figure 5.7) gives a set of complimentary spectra.

Figure 5.8. Wavelength shift experienced by four consecutive sub-gratings with increasing force. The p1 series denote one point of application and the p2 series denotes another. The curves connecting the data points are for illustration.

Using equation 5.1, the location of p1 in this plot can be determined as follows (Appendix 3). Consider the series p1-15 N. The actual values measured are tabulated as follows.

Table 5.1. Experimental data used to determine the location of the load between adjacent sub-gratings.

Wavelength of sub-grating (nm)	Wavelength shift (nm)
1538.32	0
1540.22 ($\lambda_{\max 2}$)	0.32 ($\Delta\lambda_{\max 2}$)
1542.22 ($\lambda_{\max 1}$)	0.66 ($\Delta\lambda_{\max 1}$)
1544.1	0.08

As $\lambda_{\max 2} < \lambda_{\max 1}$, equation 5.1 reduces to the following.

$$Location = 1542.22 - \left[\frac{|1542.22 - 1540.22|}{0.66 + 0.32} \times 0.32 \right]$$

$Location = 1541.567$ nm (in the reflected spectrum that corresponds to a specific point on the sampled chirped grating).

Similarly it can also be determined that location for the p2 series is at the 1540.98 nm point. Thus the separation between p1 and p2 is about 0.587 nm on the wavelength scale. The same wavelengths of the sub-gratings, as given in the table, can be used to generate a plot between the physical distance (measured with uncertainty of ± 0.1 mm) on the optical fibre and the spectral separation in wavelength (measured with uncertainty of ± 0.015 nm), as shown in Figure 5.9. The straight line is expected as a phase mask with a linear chirp was used to generate the sampled sub-gratings.

Figure 5.9. Plot of wavelength separation in the optical spectrum and the distance on the optical fibre used to determine the location of force. The data points are plotted with a linear fit through (0,0) that gives the slope to be 3.1064.

A linear fit of the data gives a slope of 3.1 mm/nm. From the slope of this plot (Figure 5.9), the wavelength separation of 0.587 nm can be converted to a distance of 1.8 mm (0.587×3.1) along the sensor. This experiment clarifies the concept of using the sampled chirped grating as a pressure-mapping sensor as the load and location can be extrapolated from the measured data even when the point of loading is not directly on any sub-grating. An extension from a single fibre to an array of gratings on multiple fibres can generate a two-dimensional pressure map.

This calculation can be extended to determine the magnitude of the pressure at the given point. Taking the located point as 1541.56 nm, x ($1542.22-1541.56$) is evaluated as 0.65 nm (~ 2 mm) and y is 1.345 nm (~ 4 mm). A simple linear plot of distance and wavelength shift {from point (0.65, 0.66) to point (1.345, 0.32)} can give the intercept at zero distance and is shown in Figure 5.10. In this case, the wavelength shift at 1541.56 (0 mm) would be 0.98 nm. The equivalent load can be deduced from this value.

Figure 5.10. The plot (solid line) of wavelength shift (nm) with respect to distance from sub-grating (mm). The dashed line can be used to extrapolate the plot to obtain the y-intercept.

5.4 Array of sampled chirped gratings

The above results show that a sampled chirped grating can be used as an array for pressure/load sensing. Two aspects of an array will be discussed in this section.

1. The effect of a distributed load applied simultaneously to two gratings in parallel.
2. The effect of a point load as it moves transversely from one grating to the other.

Figure 5.11 shows the gratings in a dual array loaded simultaneously by a semi-cylindrical weight. The spectra from both the gratings were recorded under similar conditions of loading. Knowing the layout of the gratings and the sub-gratings in the sensor helps in determining the position of the tool from the reflected spectrum. The relative shifts of the sub-gratings give the amount of force applied. By using the calibration graph of wavelength shift as a function of force, an unknown force can be quantified and located.

Figure 5.11. Embedded sensor with two sampled gratings loaded with semi-cylindrical load. The gratings have been shown for illustrative purpose only.

The selected sub-gratings, in the two spectra (Figures 5.12 and 5.13), show similar wavelength shift indicating that the loading conditions are almost identical for both gratings. The magnitude of the force on both gratings is equal. However, the

wavelengths affected are different depending on the position of the sub-gratings inside the embedded sensor. The embedding has been done such that 1537 nm sub-grating of one is located next to 1556 nm sub-grating of the other. This is also illustrated in the setup shown in Figure 5.11. It is also evident from the spectra that the measurand information, that is the magnitude of the load/pressure, is reflected in the magnitude of the wavelength shift. The spectral shape is almost unchanged.

Figure 5.12. Shifted spectrum of one grating in the dual array, when loaded as shown in Figure 5.11. The shifted peaks are highlighted with a rectangle.

Figure 5.13. Shifted spectrum of second grating in the dual array, when loaded as shown in Figure 5.11. The shifted peaks are highlighted with a rectangle. The response complements the spectra in Figure 5.12.

Figure 5.14 shows the setup used for testing the dual array of gratings. As the point of application of force was moved (along the dotted line) on the plane of the sensor, from fibre 1 (A) to fibre 2 (B), the shift experienced by a sub-grating (at A) in fibre 1, decreases linearly with distance. Simultaneously, the shift experienced by a sub-grating (at B) in fibre 2 increases monotonically.

Figure 5.14. Schematic diagram of the experimental setup showing the point of application of the force between the two gratings in the dual array.

Figure 5.15 shows the plot of the measurements taken. It is evident that as the point of application moves away from the grating, the wavelength shift experienced by the sub-grating decreases. This result verifies that the array can be used to detect the pressure in between the two fibre gratings, hence avoiding dead zones.

Figure 5.15. Wavelength shift as a function of distance between the two gratings in the dual array. The data points are connected for illustration.

5.5 Testing the tibial spacer sensor

All the experiments described so far have been done with loads applied onto flat, composite embedded sensors i.e. the optical fibre sampled chirped grating embedded in planar fibre-reinforced composites. It is crucial to study the response of the sub-gratings after the tibial spacer sensor has been designed. It has been mentioned in the previous chapter that the composite was cured in a curved shape and then built into the tibial spacer sensor. In this section, the response of a single sub-grating is presented, after the whole tibial spacer sensor was prepared and had been washed with the PMMA mixture as the top layer. This sub-grating lies encased in prepreg and PMMA, with the layer of silicone rubber in the condylar groove, as in the final design. The femoral implant was loaded using a force gauge and the setup was used to test the sub-grating in the tibial spacer sensor. The loading configuration is shown in Figure 5.16.

Figure 5.16. The loading configuration of the sensor. The force is applied on top of the femoral implant to simulate the knee joint.

This sub-grating is a part of the curved, composite embedded sensor (as shown in Figure 4.13) lying beneath the curved, articulating surface of the tibial spacer and a curved surface (femoral implant) comes into contact with it. This configuration of contact

forms the basis of all loading for the tibial spacer sensor. However, as the laboratory tests rely on a vertical load, it is difficult to test all the sub-gratings with the available setup. Ideally, the 46 sensing points (sub-gratings) would need 46 mounting configurations to be tested under a vertical load. Figure 5.17 demonstrates the application of vertical loading to individual sub-gratings. The arrows denote the direction of the load that would be required. For simplicity, only three sub-gratings are shown. A simpler way to test the sensor is by using distributed loads as in cadaveric testing, which will be discussed in Chapter 7. However, only one sub-grating, lying at the central pit of the tibial spacer groove, may be tested reliably with a vertical load.

Figure 5.17. Vertical loading configurations (arrows) for three sub-gratings (black squares) in one condylar groove of the tibial spacer sensor.

The response of one such sub-grating, when subjected to increasing loads, is presented in this section. Figure 5.18 shows two plots denoting the wavelength shifts of one sub-grating when subject to increasing loads at two different points. The straight lines denote the linear fits to the data. This experiment was carried out with the femoral

implant where the contact area cannot be estimated without theoretical calculation. The higher slope was obtained when the loading point was on the sub-grating (Point 1). The lower slope, for the same sub-grating, was obtained when the loading point was offset by 2-3 mm (Point 2). As in the case of a flat sensor the point of application influences the measured wavelength shift. This is essential in determining the stress distribution in a pressure map.

Figure 5.18. Wavelength shift of a sub-grating in the tibial spacer sensor when increasing loads were applied. The two plots are for two different points of application for the same sub-grating. The data points are plotted with linear fits passing through (0,0). Point 2 is closer to the sub-grating as compared to point 1 by a distance of 1 mm.

5.6 Hysteresis

Negligible hysteresis is a pre-requirement for dynamic use of the sensor. The area enclosed by a continuous cycle of loading and unloading determines the hysteresis of a sensor. A repeated loading/unloading test is presented in this section to demonstrate the hysteresis behaviour of the sensor. The data are compared in the next chapter with a

simplified numerical model. The same data shows more linearity when plotted against pressure (Chapter 6) instead of force (Figure 5.19). This difference in linearity may be because the increasing force applied to the surface of the tibial spacer does not transfer completely onto the sub-grating; the stress is partially distributed in a larger contact area between the femoral implant and the tibial spacer. The change in area of contact due to changing applied load is a consequence of a concave solid surface of one material being in contact with a convex solid surface of another material. In Chapter 6, the effect of changing area of stress distribution has been eliminated as the value of pressure is obtained from the ratio of the force and the calculated area of contact.

Figure 5.19. Loading and unloading plots of wavelength shift and applied force of a single sub-grating in the tibial spacer sensor.

The plots in Figure 5.19 have error bars of ± 15 pm, which is the accuracy of the OSA. Considering the uncertainty in the data, the figure shows negligible hysteresis with no definite area enclosed between the curves. Assuming that the full scale wavelength shift is 880 pm, and the hysteresis is calculated as the difference between the data points, Table 5.2 can give an estimate of the percentage of hysteresis between the curves.

Table 5.2. Percentage of hysteresis.

	Data point 1- Force 5 N	Data point 2- Force 25 N	Data point 3- Force 50 N
Load1-Unload2	4.5%	4.5%	
Load1-Unload2	4.5%	4.5%	
Unload2 - Load3		4.5%	4.5%
Unload2 - Load3		4.5%	4.5%

This test shows that the deviation in the data is much lower than the absolute values measured, as the loading and unloading curves follow each other closely. Low hysteresis of the sensor will allow the sensor to be used for dynamic measurements without the loss of accuracy and without waiting for the sensor to recover.

5.7 Effect of temperature

Increasing temperature leads to thermal expansion of the optical fibre and an increase in the fibre grating periodicity, which could cause a wavelength shift in the reflected spectrum. This sensor is meant for use at room temperatures (15-35°C). In-vivo

use would mean that the sensor will be subject to body temperature, which is expected to be steady and the ambient temperature in the operation theatre is also constant. In-vitro use will also be at room temperature in the laboratory. Due to the combined thermal conductivity of rubber, composite and PMMA, all the sub-gratings will show identical wavelength shifts when there is any temperature fluctuation. The new central wavelengths at an elevated temperature can be used as the reference for quantifying the pressure.

It has been observed in the case of fibre Bragg gratings that there is a wavelength shift of about $13 \text{ pm}/^{\circ}\text{C}$. It is understood that wavelength shift due to temperature changes will also occur in the case of sampled gratings. However, the wavelength dependence of the temperature coefficient is small [35]. Due to this, the effect of temperature on a grating at 1530 nm is almost the same as that of a grating at 1550 nm . For a sampled chirped grating, it is important that all the sub-gratings have the same gradient with temperature. This will ensure that if the ambient temperature changes, all the sub-gratings will shift by the same amount and the relative shifts will be independent of temperature.

The sensor was placed in a Feutron climate control chamber and the temperature was raised from 20°C to 40°C to observe the effect of temperature on sub-gratings in the tibial spacer sensor. The central wavelengths of five sub-gratings were recorded at intervals of five degrees. The slope of the plots lie between $14 \text{ pm}/^{\circ}\text{C}$ and $19 \text{ pm}/^{\circ}\text{C}$ (Figure 5.20). The values are recorded in Table 5.3.

Figure 5.20. Temperature response of five sub-gratings measured simultaneously.

The values in the legend identify the reference wavelengths of the different sub-gratings.

The data points are connected for visual clarity.

Table 5.3. Temperature sensitivity of various sub-gratings

Reference wavelength (nm)	Temperature sensitivity (pm/°C)
1534.84	17
1536.84	19
1538.84	18
1540.8	15
1542.64	14

The results may be interpreted as follows. When there is a temperature change of 5°C, the sub-gratings will register wavelength shifts of 70 ± 15 pm (or 14×5) and 95 ± 15 pm (or 19×5). For a few degrees rise in the ambient temperature, it can be said that all the sub-gratings have the same temperature gradient. Thus, an increase in temperature will

not affect the relative wavelength shifts that will give the magnitude of the pressure. However, for greater fluctuations in temperature, the relative thermal shifts of the subgratings have to be taken into account. In this project, all tests have been carried out at constant room temperature because TKR significant temperature changes are not expected during TKR.

5.8 Effect of water

During any in-vivo measurement, the sensor will come in contact with blood. This test was conducted to ensure that the sensor response does not fluctuate due to humidity or due to the effect of neutral fluids. To investigate this, the sensor was placed in a climate control chamber (Feutron) to measure the effect of humidity. The temperature was kept constant at 20°C. The relative humidity was changed from 75% to 45%. The effect of the change in humidity was observed by monitoring the reflected spectrum. There was no change in the spectrum implying that the sensor is not affected by changes in atmospheric humidity. A teaspoon of water was then poured onto one of the grooves of the tibial sensor. The spectrum was monitored for one hour. No change of the spectra was observed. This test suggests that the sensor is not affected by the humidity or wetness of the surrounding region and hence can be used in-vivo accurately. However, fluids of varying pH (acidic or basic) may affect the sensor response as they will be corrosive on PMMA.

5.9 Repetitive loading test

Accuracy under repetitive loading is one of the criteria for a sensor. The femoral implant was placed on top of the tibial sensor (as shown in Figure 5.16) and a vertical load was applied on the femoral implant with the digital force gauge. A single fixed load of 20 N was applied to the same location again and again, for five seconds each minute. The wavelength shift pattern of the sub-gratings remained the same for each loading. The magnitude of the wavelength shift also remained the same for all the gratings. The loading was repeated ten times (over a span of ten minutes) and there was no change in the response of the sensor. Besides, the hysteresis curve, this test verifies the repeatability of the sensor.

5.10 Effect of prolonged loading

It is known that when a load is applied to some materials, the effect of the force creeps in slowly into the material and there is increasing deformation with a constant load. This effect may damage the sensor. As the sensor may not have fully recovered before subsequent measurement, errors will be introduced. This sensor has been tested with a load of 45 N applied continuously for a period of 15 minutes. No change was observed in the original spectrum after 15 minutes of loading and so it may be concluded that the time span of loading will not affect the magnitude of the wavelength shift measured.

5.11 The pressure map

The pressure-mapping sensor should, ultimately, generate a pressure map. To clarify, when the tibial spacer sensor is loaded with a femoral implant, the 23 sub-gratings in each groove register wavelength shifts that give the distribution of the pressure although the contact area may be small. This response is the combined effect of the PMMA, composite and rubber as observed in the reflected spectra of the optical fibres. The data can be displayed in Matlab with the XYZ plot function “meshgrid” and “surf”. This function takes the physical positions of the sub-gratings as the intersections of an XY grid map. The wavelength shifts are plotted on the Z axis. “Surf” generates a 3 dimensional surface with the depths and peaks controlled by the z-axis. Each grid is coloured with a Gaussian average of the four corners. Adding an “interp” command to the program gives an interpolated display that is easier to understand.

Figures 5.21 (a-d) show the Matlab patterns generated from the measured wavelength shifts. The load was applied to the femoral implant placed on the tibial spacer. Figure 5.21 (a) shows the spread of wavelength shifts at 25 N of vertical loading whereas Figure 5.21 (b) shows the effect of 55 N of vertical load. The positive wavelength shift at the centre increases as does the negative wavelength shift at the edges, when compared to the 25 N load. Figure 5.21 (c) shows the distribution when the loaded (25 N) femoral implant is severely tilted towards the right edge. Figure 5.21 (d) shows the distribution when the femoral implant axis is rotated by approximately 5–10 degrees.

Figure 5.21. Maps of wavelength shift (in nm) showing the effect of various loading conditions on one condylar groove in the tibial spacer sensor: (a) 25 N of vertical loading, (b) 55 N of vertical loading, (c) femoral implant is severely tilted towards the right edge, and (d) femoral implant axis is rotated by approximately 5–10 degrees. The maps

represent the response of 23 sub-gratings. The x-axis represents the length in millimetres (20 mm) along the width of the condyle. The y-axis represents the length in millimetres (24 mm) along front-to-back axis of the condyle. The colours represent the z-axis, with the magnitude of wavelength shift in nm, as shown by the colour bars.

The results in Figure 5.21 show that the pressure map can detect the change in magnitude of the load. The 25 N and 55 N maps show that the magnitude of the wavelength shift increases with the overall pattern remaining almost the same. From the maps of the inclined and rotated loading cases, it is evident that the contact region can be determined as well.

5.12 Explanation of the results

The results displayed by the pressure map show a definite spread of compressive and tensile behaviour i.e. positive and negative frequency shifts. The negative shift at the edges of the map shows that there is a certain amount of force that is pushing the composite embedded fibre upwards. Both the PMMA and rubber can cause this. A block of PMMA or rubber will show upward displacement at the edges when compressed downwards at the centre. The magnitude of displacement will be much smaller in the case of PMMA. It is likely that this behaviour of the top layer of PMMA in the sensor combined with the rubber results in the negative shifts.

The following example will be used to illustrate the cause of positive and negative wavelength shifts. Finite elements method is used to model the effect of load on a block

of acrylate. The model is qualitative and does not represent real data. Figure 5.22 (a) shows a block of acrylate constrained at all the nodes on the bottom surface. A force applied at the edge produces a depression where expected, but there may be a very small rise as well (b).

Figure 5.22. Finite elements figures of a constrained, acrylate block (a) and the effect of strain on it (b). The bottom figure (b) shows the z-displacement only. This is a model calculation only and does not represent the experimental data quantitatively.

From practical experience, it is obvious that rubber will demonstrate similar behavior to a greater extent. The colour bar in the figure shows the z-displacement in metres. It records a depression of 13 nm and a rise of 23 pm (approximately 500 times less). However, the actual magnitudes of positive and negative displacement will differ, depending on the nodes, meshing, constraints and loads used in the analysis.

5.13 Summary

The above tests demonstrate the suitability and feasibility of the sensor in the context of its application. Success in simultaneous measurement of pressure and position is crucial in pressure mapping. Array behaviour is important to determine the distribution of stress. Negligible creep and hysteresis influence the scope for dynamic use. These results provide the background for the cadaveric tests that will eventually determine the success of the sensor. Repeatability, sensitivity, creep, hysteresis and resolution are important parameters for most sensors. This chapter presents the results for all of the above factors.

Chapter 6

Theory and numerical model of the sensor

The development of the sensor described so far has been influenced by fundamental properties of the materials used, such as the unidirectional properties of the composite, the change in periodicity of the optical fibre grating, the curvatures of the PMMA spacer and the compressive behaviour of the silicone rubber. The overall response of the sensor is governed by the mutual influence (mechanical and material) of all the components. Though some of these properties have guided the development from the beginning of this work (Chapter 2 and 4), it is appropriate that they should be considered in the context of the final tibial spacer sensor. The response of the embedded sampled chirped grating could not be presumed before systematic experiments were carried out. The requirement for the layer of silicone rubber to enhance linearity of the sensor response was also determined during experimentation. Hence the influence of all the factors can be analysed at this stage.

This chapter presents the basic theory of the composite, the contact mechanics of the curved femoral implant with the curved tibial spacer, and their effect on the optical response of the fibre grating. To retain the continuity, the basic theories are discussed in this chapter together with the inferences drawn from the experimental results. The numerical calculation is carried out to analyze the results obtained in Chapter 5.

6.1 Composite analysis

The analysis of the simple, four-layer stack used in this work has vast scope. The stresses, strains, moments and couplings involved in a stack of four unidirectional prepreg layers are many. The composite embedding of the optical fibre is a part of the entire sensor and testing. In the following section, a simple explanation of the theory is provided. The Matlab computer program implementing the model is attached in Appendix 4.

It is understood that the sequence of the stacking and the angle of orientation of the prepreg layers influence the final behaviour of the laminate or prepreg stack [87].

For isotropic materials, properties such as the Young's modulus and the Poisson's ratio are the same in all directions. However, this is not applicable to orthotropic materials such as a layer of unidirectional prepreg. All material constants are assigned a subscript to define their direction. For example, in the case of unidirectional prepreg, the modulus of elasticity E_1 (along the longitudinal direction) is not equal to E_2 (along the transverse direction).

In the Figure 6.1, the material axes X, Y coincide with the geometric axes 1,2. In the Figure 6.2, the material axes are rotated by an angle θ with respect to the geometric axes. In the following analysis, the conventional notation will be E_x (or E_1) along X and E_y (or E_2) along Y.

Figure 6.1. The orientation of geometrical (1,2) and material axes (X,Y) of a unidirectional prepreg sheet. Both sets of axes are aligned in the same direction. The material axis is shown with dotted lines.

Figure 6.2. The orientation of geometrical and material axes of a unidirectional prepreg sheet, when the sheet is cut at a certain angle. The angle θ between the two sets of axes influences the stacking properties.

For a lamina, or a sheet of the prepreg, the stiffness matrix is defined as follows [88]:

$$Q = \begin{bmatrix} Q_{11} & Q_{12} & Q_{16} \\ Q_{21} & Q_{22} & Q_{26} \\ Q_{61} & Q_{62} & Q_{66} \end{bmatrix} \quad (6.1)$$

To determine this matrix, the following parameters have to be defined.

$$\nu_{21} = \frac{\nu_{12} \times E_2}{E_1} \quad (6.2)$$

where ν_{12} and ν_{21} are the Poisson's ratios, and

$$Q_{11} = \frac{E_1}{1 - \nu_{12}\nu_{21}} \quad (6.3)$$

$$Q_{12} = \frac{\nu_{12} \times E_2}{1 - \nu_{12}\nu_{21}} \quad (6.4)$$

$$Q_{21} = Q_{12} \quad (6.5)$$

$$Q_{22} = \frac{E_2}{1 - \nu_{12}\nu_{21}} \quad (6.6)$$

$$Q_{66} = G_{12} \quad (6.7)$$

where G_{12} is the in-plane shear modulus.

$$Q = \begin{bmatrix} Q_{11} & Q_{12} & 0 \\ Q_{21} & Q_{22} & 0 \\ 0 & 0 & Q_{66} \end{bmatrix} \quad (6.8)$$

To analyze a stack or laminate having multiple laminae, it is necessary to know the stress-strain relationships and the elastic constants for the 'generally orthotropic lamina' where the angle can take any value.

The angle θ can be understood from the illustration given in Figure 6.2.

The elastic constants for these off-axis directions have to be determined in terms of the orientation angle. The transformed lamina stiffness matrix now becomes $\bar{Q}$ (or P).

The components of this transformed matrix are as follows [88]:

$$q_{11} = Q_{11} \cos^4 \theta + Q_{22} \sin^4 \theta + 2(Q_{12} + 2Q_{66})(\cos^2 \theta + \sin^2 \theta) \quad (6.9)$$

$$q_{12} = (Q_{11} + Q_{22} - 4Q_{66})(\cos^2 \theta + \sin^2 \theta) + Q_{12}(\cos^4 \theta + \sin^4 \theta) \quad (6.10)$$

$$q_{22} = Q_{11} \sin^4 \theta + Q_{22} \cos^4 \theta + 2(Q_{12} + 2Q_{66})(\sin^2 \theta + \cos^2 \theta) \quad (6.11)$$

$$q_{16} = (Q_{11} - Q_{12} - 2Q_{66})\cos^3 \theta \sin \theta - (Q_{22} - Q_{12} - 2Q_{66})\cos \theta \sin^3 \theta \quad (6.12)$$

$$q_{26} = (Q_{11} - Q_{12} - 2Q_{66})\cos \theta \sin^3 \theta - (Q_{22} - Q_{12} - 2Q_{66})\cos^3 \theta \sin \theta \quad (6.13)$$

$$q_{66} = (Q_{11} + Q_{22} - 2Q_{12} - 2Q_{66})\cos^2 \theta \sin^2 \theta + Q_{66}(\sin^4 \theta + \cos^4 \theta) \quad (6.14)$$

$$q_{21} = q_{12} \quad (6.15)$$

When the lamina is at 0 or 90 degrees, q_{16} and q_{26} are zero.

$$P = \begin{bmatrix} q_{11} & q_{12} & 0 \\ q_{21} & q_{22} & 0 \\ 0 & 0 & q_{66} \end{bmatrix} \quad (6.16)$$

For the analysis of a stack or laminate, the Classical Lamination Theory is considered [87, 88]. This theory is generalized to analyze non-symmetric laminates with the plies oriented at arbitrary angles. Such a laminate has various coupling effects

between the plies and may have combinations of extensional, flexural and torsional deformations. However, each ply is assumed to be in a state of plane-stress. It is also assumed that individual laminae are homogenously bonded together to behave as a plate.

Figure 6.3. The diagram explains the nomenclature of z with respect to a stack of prepreg.

z_0 is the distance from the middle plane to the top surface of the stack and z_N is the distance from the middle plane to the bottom surface of the stack.

For the stack of thickness t and number of layers N , Figure 6.3 depicts the distances of the various planes with respect to each other. The term z_k denotes the distance from the middle surface to outer surface of the k th lamina. In terms of z_k , the following matrices predict the behaviour of the stack.

The laminate extensional stiffnesses are,

$$A_{ij} = \sum_{k=1}^N (Q_{ij})_k (z_k - z_{k-1}) \quad (6.17)$$

The laminate coupling stiffnesses are,

$$B_{ij} = \frac{1}{2} \sum_{k=1}^N (Q_{ij})_k (z_k^2 - z_{k-1}^2) \quad (6.18)$$

The laminate bending stiffnesses are,

$$D_{ij} = \frac{1}{3} \sum_{k=1}^N (Q_{ij})_k (z_k^3 - z_{k-1}^3) \quad (6.19)$$

Now we can define certain matrices that are derived from the above stiffness matrices [88]. The following matrices are useful in determining the elastic constants of the laminate.

$$[A^*] = [A]^{-1} \quad (6.20)$$

$$[B^*] = -[A]^{-1}[B] \quad (6.21)$$

$$[C^*] = [B][A]^{-1} \quad (6.22)$$

$$[D^*] = [D] - [B][A]^{-1}[B] \quad (6.23)$$

$$[A'] = [A^*] - [B^*][D^*]^{-1}[C^*] \quad (6.24)$$

$$[D'] = [D^*]^{-1} \quad (6.25)$$

The effective longitudinal Young's modulus of the laminate E_x , that determines the response of the laminate under a single axial load per unit length, is given by

$$E_x = \frac{1}{tA'(1,1)} \approx 33 \text{ GPa} \quad (6.26)$$

The effective transverse Young's modulus E_y is similarly determined as

$$E_y = \frac{1}{tA'(2,2)} \approx 19 \text{ GPa} \quad (6.27)$$

For a rectangular lamina, the flexural modulus is given by

$$E_{fx} = \frac{12}{t^3 D'(1,1)} \approx 28 \text{ GPa} \quad (6.28)$$

The values of E_x , E_y and E_{fx} given here are the calculated values obtained for this specific sensor design using the Matlab program given in Appendix 4.

The flexural modulus determined from here will be used later as part of the numerical model. The value obtained here has been solved for the specific case of 90/0/0/0 stacking as used for the sensor. At this stage, the effect of the angle of orientation, θ in this case, may also be pointed out. It changes E_x and E_y . To improve the array behavior, the angle is chosen such that E_y is also enhanced (Table 6.1).

Table 6.2 shows the effect of the number of transverse layers on the flexural modulus. It may be noted that the flexural modulus is highest when all the layers are longitudinal. However, array behavior (determined by the magnitude of E_y) and flexion (determined by the magnitude of E_{fx}) have to be optimized, because more cross layers will increase E_y but will reduce E_{fx} and vice versa.

Table 6.1 Variation in elastic moduli with angle of orientation. (All values are in GPa)

Angle of top layer	E_x	E_y	E_{fx}
0	42	15	41.95
30	34.86	14.27	27.49
60	30.89	14.79	24.85
90	32.72	18.85	27.95

Table 6.2 Variation in elastic moduli with laminate configuration. (All values are in GPa)

Placement of transverse layers	E_x	E_y	E_{fx}
90/0/0/F/0	32.72	18.85	27.95
90/0/0/F/90	35.36	21.82	18.39
90/90/0/F/90	34.83	21.6	16.43

From Table 6.1, it can be seen that E_y is highest for top layer at 90 degrees. More layers at 90 degrees can increase E_y (Table 6.2) but significantly reduce E_{fx} as well. The

two tables above are guidelines for the selection of the design. The values are calculated from the equations given above. However, many more combinations can also be made.

6.2 Area of contact between the femoral implant and the tibial spacer sensor

The contact between the femoral implant used for the loading and the groove of the tibial sensor is treated as two circular bodies in contact [89, 90]. Though neither of the surfaces is circular, it is assumed that over the region affecting a single sub-grating (2-5 mm) the curvatures have single radii, respectively. There is another set of radii along the perpendicular direction.

The knee structure cannot be simplified by a couple of radii of curvatures. The structures may be modeled with finite elements method. However, current research reports reveal that when the size of the elements is reduced and the number of elements increased, the finite elements model is more accurate. With this perspective, if the knee joint is divided into hundreds of sections, with each section or element having different radius of curvature, the accuracy of the model will be higher. Thus, it is unrealistic to assign a couple of radii to the femur or tibia. In this analysis, the radii will be assigned only to a specific area of a few millimetres.

The area of contact is based on formulae proposed by a Hertzian mechanics [90] model of a convex surface of radius R_1 ($D_1/2$) touching a concave surface of radius R_2 ($D_2/2$) (Figure 6.4).

Figure 6.4. Illustration of a curved-on- curved contact. The radius of the contact area is a .

The equivalent diameter K_d and the equivalent elastic modulus C_E of the two body system are given by

$$K_d = \frac{D_1 \times D_2}{D_1 - D_2} \quad (6.29)$$

$$C_E = \frac{1 - \nu_1^2}{E_1} + \frac{1 - \nu_2^2}{E_2} \quad (6.30)$$

For an applied load F , the radius of the contact area will be given by “ a ” that is related to F as follows.

$$a^3 = F \frac{K_d}{2} \times \frac{1}{4/3C_E} \quad (6.31)$$

$$a = \sqrt[3]{\frac{3}{4} \times F \times \frac{K_d}{2} \times C_E} \quad (6.32)$$

$$a = 0.721 \times \sqrt[3]{FK_d C_E} \quad (6.33)$$

The contact area is approximated as an ellipse of changing semi-major axis “a” and constant semi-minor axis “b” (Figure 6.5). The value of “b” is equal to “a” when the force F is tending to zero. The direction of “a” is along the length of the sub-grating in contact. It is obvious that the area of contact changes with the applied force and is dependent on the materials of the bodies 1 and 2.

Figure 6.5. The changing contact area between a concave surface and a convex surface, of different radii, due to the effect of applied force.

The force F is the load applied to the femoral implant. The pressure experienced in the contact region is given as,

$$P = \frac{F}{\pi ab} \quad (6.34)$$

In case of the tibial spacer sensor coming into contact with the femoral implant, the materials that are involved are Titanium alloy and the PMMA that forms the top

surface of the spacer sensor. The major and minor axes of the contact ellipse are calculated using the properties of these two materials (Appendix 5).

6.3 The mechanics and optics for a single sub-grating

From the theory of a fibre Bragg grating [91] it is known that the wavelength shift is a linear function of the axial strain, at constant temperature. (Also shown in equation 2.7.) The strain is a result of the applied stress (or pressure) as above. In this section, a simple approximation is presented to relate the mechanical effect and the optical shift. The resulting plot will be compared with the previous, experimental hysteresis results.

The expression for wavelength shift at constant temperature that reduces to,

$$\Delta\lambda = \lambda(1 - p_e)\varepsilon \quad (6.35)$$

where ε is the axial strain. The above equation can be approximated to the following, assuming that the strain–optic effects are kept constant [91],

$$\Delta\lambda = \lambda \times 0.78 \times \varepsilon \quad (6.36)$$

The variable in this equation is the strain as λ will be constant for a fixed sub-grating. Also, the influence of λ is negligible within a range of a few nanometres and the major contribution comes from the strain effect. The following points have to be considered before making any assumptions for the model.

1. A vertical transverse section of the sensor will show three materials: the top and bottom PMMA layers, the composite layer with the grating and the layer of rubber beneath it.
2. The wavelength shift is caused by the axial strain in the fibre that is a result of the flexion of the composite.
3. The layers in contact with the composite will influence it most.
4. The flexion of the composite layer will be mainly determined by the flexural modulus of the composite (as calculated above) and the compressive modulus of the rubber.
5. The bottom layer of PMMA may be neglected.
6. In this model, the influence of the top layer of PMMA will also be neglected. This is because it has been reported [92] that for materials such as PMMA that are elastic-plastic, the peak stress occurs 1-2 mm below the surface (von Mises effect). The thickness of the top layer in the sensor is about 1 mm.

This set of considerations brings us to the two layers of rubber and composite. For a beam of two materials that experience equal strain for a given stress, it can be said [93] that the force is proportionally distributed between the two materials depending on their area of cross-section.

In this case, if “b” determines the breadth of contact, the heights of the layers will complete the cross-section (Figure 6.6).

Figure 6.6. Vertical cross-section of the tibial sensor layers.

Let the total normal stress (P) be available as axial stress. The force will be distributed across the two materials.

Total stress = stress across composite (with the optical fibre) + stress across rubber

$$P = A_f E_f \varepsilon + A_r E_r \varepsilon \quad (6.37)$$

where A and E denote the surface area and the elastic modulus of the respective material.

$$P = b h_f E_f \varepsilon + b h_r E_r \varepsilon \quad (6.38)$$

The proportion of the stress across the composite will be as follows.

$$P_f = \frac{b h_f E_f \varepsilon}{b h_f E_f \varepsilon + b h_r E_r \varepsilon} \times P = \frac{h_f E_f}{h_f E_f + h_r E_r} \times P \quad (6.39)$$

For the grating, the strain is determined by P_f/E_f . So the wavelength shift can be written as

$$\Delta\lambda = \lambda \times 0.78 \times \frac{P_f}{E_f} \quad (6.40)$$

Substituting equation. 6.39 in equation. 6.40, we obtain

$$\Delta\lambda = \lambda \times 0.78 \times \left\{ \frac{1}{E_f} \times \left(\frac{h_f E_f}{h_f E_f + h_r E_r} \times P \right) \right\} \quad (6.41)$$

So the final expression for the wavelength shift in terms of the applied load, the elliptical area of contact and the elastic modulus of the composite can be given as follows.

$$\Delta\lambda = \lambda \times 0.78 \times \frac{F}{\pi b} \times \frac{h_f}{(E_r h_r + E_f h_f)} \quad (6.42)$$

In this final numerical representation of wavelength shift, a , b and E_{fx} are calculated values using standard equations as mentioned above (in sections 6.1 and 6.2 of this chapter) for this specific design of sensor.

However, it may be noted that the equation will only match the behaviour of a sub-grating when the point of contact is located on the sub-grating. When the point of loading is not directly on the sub-grating, the experimental data will have a slope lower than the theoretical curve. For a comparison between the theoretical and experimental values, a plot is generated (Appendix 5) and is shown in Figure 6.7. The experimental values are tabulated from the spectral wavelength shift of the sub-grating. The force applied is converted to pressure by using the calculated contact area. This eliminates any mismatch due to errors in contact area estimation. Thus, the comparison of data is only between the measured shift and the calculated (from optics and mechanics principles)

wavelength shift. All the plots pass through (0,0) as the wavelength shift is a linear function of the applied pressure.

Figure 6.7. The experimental and theoretical plots of wavelength shift as a function of the applied pressure.

The simplified model presented above gives the intrinsic response of the sub-gratings when they are subjected to pressure. In principle, the model can be extended to all the 46 sub-gratings, provided that each sub-grating can be loaded in an identical manner. The radii of curvatures would be different in each case.

This analysis shows that the sensor response is as expected for a model based on fundamental mechanical and material properties of the constituent components. The next chapter will establish the practical aspect of the tibial spacer sensor in cadaveric testing.

Chapter 7

Cadaveric Testing

The cadaveric tests were intended to verify the suitability of the sensor on two grounds.

1. The sensor can be used for alignment during the surgical process of joint replacement. This was the basis for testing the sensor in various configurations of malalignment and angles of flexion.
2. The sensor can be used by manufacturers of implants for total knee arthroplasty. It can be used to improve the quality, design or material parameters depending on the response in pressure mapping. This was the basis for testing the sensor with multiple loads in extension.

Figure 7.1. Picture of a cadaveric knee cut open showing the femur cartilage affected by arthritis (outlined in black).

7.1 Implanting the sensor

The tibial spacer sensor is meant to assist surgeons with prostheses alignment. The best way to test the sensor in a cadaveric knee is to fit it where the polymer implant should actually be positioned. An entire procedure of total knee joint replacement surgery was performed for this. A longitudinal cut was made along the length of the knee and the muscles and tissues were cleared from the joint to expose the bones (Figure 7.2). The femoral end of the cadaveric knee was sawed off in a series of steps, with many measurements taken in between sawing, with the appropriate jigs. Then the femoral implant was hammered onto the shaved end of the femur. A layer of bone was shaved off the tibia. A hole was drilled into the tibial bone and the tibial tray was inserted into the hole. The tibial spacer sensor was inserted and the femur was rolled over the tibial spacer to check the alignment. *This is exactly where the sensor finds its use.* During an actual surgical procedure, the tibial tray and the femoral prosthesis can be correctly positioned and balanced by observing the pressure maps from the sensor. The bones can also be restructured by repeated monitoring with the sensor. After satisfactory alignment, the tibial tray and femoral prosthesis can be cemented in place. Then the sensor can be removed and the actual prosthetic spacer fitted in. However, for the experimental procedure, the sensor was allowed to remain on the tibial tray in the cadaveric knee joint. The patella was flipped into position and the cut was sutured. All the tests presented in this chapter were carried out with the sensor sutured inside the knee joint.

Figure 7.2. Total knee joint replacement being performed on a cadaveric knee joint- femoral implant has been placed and tibial end is intact.

Figure 7.2 shows one of the cadaveric legs during the process of total knee surgery. The femoral end has been sawed away and the femoral implant hammered into place. A set of metallic measurement tools (seen in the picture) and pins were used to determine the axis and length of the tibia that needed to be cut. Figure 7.3 shows the tibial spacer sensor inserted onto the tibial base plate, where it is in position to measure the alignment. The black tubes in the background encase the fibres, including the splices. The patella has been flipped to one side to allow for the TKR. The patella tendon has been stringed to control the angles of flexion.

Figure 7.3. A cadaveric knee joint with the total knee procedure completed and the sensor inserted for measurements.

Figure 7.4 displays a pressure map within an approximate contour of the condylar groove and an estimated layout of the sensors. The black lines with the heavy dots show the fibres with the sub-grating positions. The actual pressure map obtained after measurement of the wavelength shifts forms the background of the figure. As the layout of the sub-gratings in the tibial spacer sensor is known, the pressure distribution can be estimated directly from the tabulation of the wavelength shifts (Appendix 7).

Figure 7.4. Pressure map within an approximate contour of the condylar groove and an estimated layout of the sensors. The dots represent the sub-gratings.

7.2 Vertical loading test

A vertical loading test was carried out with high static loads. A person weighing 50 kg is equivalent to a load of 500 N. However, during the process of surgery, the person lies down and the expected weights are much lower. The sensor was loaded vertically in extension to ensure that the sensor can withstand forces used by surgeons to lock the tibia and femur and check the alignment. At a certain load, the wavelength shifts of the sub-gratings were recorded and the pressure distribution was mapped out.

The picture in Figure 7.5 shows the cadaveric knee, with the sensor sutured in, mounted on a loading machine (Lloyd). The knee is in extension, as would be the case in standing upright.

Figure 7.5. A cadaveric knee, with the sensor sutured in, mounted in extension for vertical loading tests.

The wavelength shift map in Figure 7.6 is of one condyle of cadaver 1 at 45 N load. The compressive load converts to contact pressure at the interface, which is measured by the sensor. The cadaver had obvious varus-valgus malalignment at the end of the total knee procedure. The map shows that the femoral implant is making contact at the edges of the condylar groove. The severe force at the edge also contributes to the large area of negative displacement. It is also possible that the femoral axis may be rotated with respect to the tibia.

Figure 7.6. The wavelength shift map (in nm) under a vertical load. The x-axis is 20 mm and the y-axis is 24 mm in length.

Figure 7.7 shows the possible mismatch in the axis of the femoral implant and the tibial condylar groove, during the loading, which is the cause of generating a pressure map as shown in Figure 7.6. Such a condition of alignment can lead to two areas of contact depicted by the regions of high wavelength shift. It may be noted that a higher portion of load is on the anterior side of the condylar groove.

Figure 7.7. Illustration of loading condition as inferred from pressure map. The femur is shown in wire frame and the contact regions between the femur and the tibial spacer are shown as red and blue patches.

Figure 7.8 shows the effect of a vertical load of 45 N for another cadaver. The wavelength shift maps give the response of both the condylar grooves. The dotted ellipse at the top denotes the patella. The map on the right shows a higher magnitude of load and as this was the right leg, it shows a tendency towards varus. Severe varus or the axis of the leg inclined towards the body can result in knock-knees. Also, the loading is towards the posterior edges of the tibial spacer. This may arise due to tightness at the joint.

Figure 7.8. Wavelength shift maps of both the condylar grooves for a vertical load of 45 N for a cadaver. The dotted pink line shows the possible contact axis that can generate such a pressure pattern. The map of the right condylar groove shows a peak wavelength shift of 0.8 nm at the lower edge.

From Figure 7.8, it may also be observed that the femoral axis is rotated with respect to the condylar groove. This conclusion is drawn from the location of the contact region (or peak pressure) on the two sides. The two parallel dotted pink lines are the likely axes of contact between the femoral implant and the tibial spacer sensor.

7.3 Angles of flexion and malalignment

On the surgical table, the load is not very high because the knee lies supported on a surface. However, rotation and misalignment of the femoral implant contribute to complications. The tibial spacer sensor can be used to check the correct positioning of the implants. To verify the suitability and sensitivity of the sensor, it was tested at various angles of flexion. The procedure is outlined below.

The femoral bone of the cadaveric knee was clamped and the tibial end was left free. The patella tendon was stringed to a pulley on a loading/displacement machine (Instron). As the pulley moved up, it pulled the patella tendon changing the angle of flexion of the knee. With this method, the sensor was used to study various angles of flexion ranging from 30 to 70 degrees and the consequent pressure maps generated are given in this section.

The picture in Figure 7.9 shows the setup used for testing various angles of flexion. The femur was clamped and the tibia was allowed to hang freely. The patella tendon was stringed and allowed to run over two pulleys. The pulleys were aligned in one vertical line. As the top pulley moved upwards, it tugged at the string and consequently, at the patella tendon. The tendon pulled up the tibia, thus changing the angle of flexion of the knee. This method simulates the natural mechanics of the knee. The pictorial representation is shown in Figure 7.10.

Figure 7.9. A cadaveric knee, with the sensor sutured in, clamped and stringed to study the angles of flexion.

Figure 7.10. Illustration of the effect of the patella tendon: (a) when the knee is in extension and (b) when the knee is in flexion. The tendon pulls the tibia with it.

For a quick understanding of the location of the pressure map, the subsequent figures will bear an icon similar to the figure given below (Figure 7.11). The two vertical ellipses denote the condylar grooves of the tibial spacer. The coloured rectangle denotes the location of the pressure map given. The ellipse at the top depicts the position of the patella.

Figure 7.11. Icon showing pressure map location in the tibial spacer.

Results are given below for three different cadaver knees.

Cadaver 1 - Angles of flexion:

The articulation surfaces of the tibial spacer and the femoral implant are not congruent. They have varying radii of curvatures in all directions. When there is a slight twist in their axes of orientation, the conformity is reduced. The contact area between the surfaces is reduced. Also, there may be contact at more than one point, away from the centre. As the patella tendon was pulled to change the angle of flexion of the knee, the force distribution also changed. The plots (Figure 7.12) show that the contact moves to the edge. The peak pressures also change. These measurements were taken with a

malaligned knee, which was an actual consequence of the surgical procedure. The specimen had a severe varus tilt as the tibial end was cut off on a slant.

Figure 7.12. Maps showing the wavelength shifts (in nm) for various angles of flexion: top row, from left, 60 degrees, 50 degrees; bottom row, from left, 40 degrees, 30 degrees. Note the similarity in pattern and difference in magnitude for 60 degrees and 30 degrees.

Each map covers an area of 20 mm × 24 mm.

Fig 7.13. The top view of the cadaveric leg that may generate a loading pattern as in the case of the pressure maps for cadaver 1 shown in Figure 7.12. The bones are in contact on one side whereas they are apart on the other side.

Cadaver 2 - Angles of flexion:

As mentioned earlier, the relative alignment of the tibial and femoral components influences the pressure maps. In practice, the axial orientations are overlooked, based on the viewpoint that the tibia and femur adjust themselves to lock into each other. However, the loss of alignment is transferred to the hip, which can be the beginning of an entirely different joint problem.

For cadaver 2, a smaller tibial spacer was used than what was actually needed. According to the size of the cadaveric knee, the 59 mm I/B-II (Zimmer Insall/Burstein II) was one size smaller and hence a misfit. In this case, there was no obvious varus-valgus tilt or femoral rotation. The maps in Figure 7.14 show that the loading is off-centre, but there is an improvement as compared to cadaver 1. The axis of contact remains congruent in this case.

Figure 7.14. Maps showing the wavelength shifts for various angles of flexion: top row, from left, 70 degrees, 60 degrees; bottom row, from left, 50 degrees, 40 degrees. The dotted pink line on the 70 degrees map shows the possible contact alignment of the femoral implant. Note the similarity in the patterns of 70 degrees and 40 degrees. Each map covers an area of 20 mm $\times$ 24 mm.

As the angles of flexion change, the pressure maps change as well. The contact area reduces and increases. At 40 degrees flexion, the free-hanging tibia was twisted away from the vertical axis of the leg to two different positions (Figs. 7.15 and 7.16). The pressure maps show slight changes in the distribution of the pressure with the axis of

contact remaining almost same in all the three cases. This set of data demonstrates the sensitivity and repeatability of the pressure mapping sensor.

Figure 7. 15. Various twists of the tibia, at 40 degrees flexion.

Figure 7.16. Possible contact alignment for cadaver 2 that generates the pressure maps given in Figures 7.14 and 7.15. The femoral and tibial implant fit into each other giving well-defined stress regions.

Cadaver 3 - Angles of flexion:

Figure 7.17 shows the wavelength shift maps for cadaver 3 at 40 degrees of flexion. The axis of contact remains in the centre but is rotated slightly as observed in vertical loading. The map on the right again experiences the higher magnitude of pressure. Figure 7.18 shows the wavelength shift maps of the same setup at 60 degrees of flexion. The contact axis remains same with contact stresses and patterns changing. Figure 7.19 illustrates the tilted_contact generated at the interface of the tibia and femur at 60 degrees of flexion. The femur is represented in wire frame for easy visualization.

Figure 7.17. Wavelength shift maps for cadaver 3 at 40 degrees of flexion. The axis of contact remains in the centre but is rotated slightly as observed in vertical loading.

Figure 7.18. Wavelength shift maps of the same setup of cadaver 3 at 60 degrees of flexion.

Figure 7.19. Illustration of the tilted contact generated at the interface of the tibia and femur at 60 degrees of flexion.

7.4 The *pressure map*

Though it may be appropriate to understand the sensor response completely, the only part relevant to the surgeon is the pressure distribution. Figure 7.20 shows the pressure maps for cadaver 3 at 40 degrees flexion in terms of megapascal (MPa). A sensitivity of ~ 121 pm/MPa is used to generate the maps. Also, all the regions of negative wavelength shift have been set to zero. The colour bar intensity is in terms of MPa. The overall pattern remains the same because of the linear response of the sensor (linear scaling between wavelength shift and pressure).

Figure 7.20. Pressure maps for cadaver 3 at 40 degrees flexion in terms of MPa.

7.5 Discussion

The pressure maps show the behavior of the knee in various modes of extension, flexion and rotation. These results obtained from the three cadavers follow some established trends [94, 95]. The angles of flexion repeat a cycle of contact every 30 degrees. The exact contact area and loads are however, different. Also the axis of contact remains the same for all the angles of flexion for one cadaveric knee. This axis is determined at the point of fitting the femoral implant. The tilt in the axis of contact is a consequence of femoral rotation that is relative to the position of the tibial spacer.

A comparison with published results using finite element methods [96, 97] shows similarity with the results obtained in this project and is discussed in more detail below. Other statistical analyses [98] and pictures of damaged tibial spacers [99, 100] also corroborate the data. It may be possible that the combination of finite negative displacements (depression) and miniscule positive displacements causes the tibial spacer to chip off in layers.

The results obtained with this sensor during cadaveric testing are consistent with published results. The following discussion illustrates this.

Figure 7.21 shows a picture of wear patterns of a tibial spacer obtained during revision surgery from a patient [100]. This tibial spacer (implant) was removed because it was worn out and had to be replaced with a new one. The worn out areas are marked with a boundary. It can be seen that, although the entire condylar region is affected by wear,

higher loads can cause more damage to specific areas that may not lie in the centre of the condylar regions. Many factors such as the gait of a person, unbalanced loading or malalignment can lead to such unequal wear patterns. In case of cadaver 1, the peak pressures were mapped at the edges of the condylar region and such loading can lead to wear patterns that are off the centre.

Figure 7.21. Picture of retrieved tibial spacer. From Harman et al [100].

In Figure 7.22 the results calculated from another study using finite element analysis (FEA) are shown [97]. This study shows very specific FEA data that agrees well with the results obtained in this research. These FEA results show that implant mismatch could be the cause of obtaining pressure patterns towards the edges as in the case of cadaver 2. Also, the maximum stress occurs just below the tibial spacer surface. It also shows the expected stress pattern caused by a rotated femoral implant.

Figure 7.22.1

Figure 7.22.2

Figure 7. 22. Finite element analysis results of tibiofemoral contact. Source: Research showcase, Aeromechanics department, University of Sydney, Australia [97]. Figure 7.22.1 shows the stress distribution for two sets of implants in alignment. Figure 7.22.2 shows the stress distribution when malalignment is introduced.

Table 7.1. Comparison of FEA results of Figure 7.22 [97] with cadaveric test results.

	Cadaveric testing	FEA [97]
1.	Figure 7.14: Stress distribution aligned near the centre of the condylar region.	Figure 7.22.1 (a): Stress distribution aligned near the centre of the condylar region.
2.	Figure 7.12: Stress distribution near the edges of the condylar region. Possible cause: Varus cut of the tibia	Figure 7.22.1 (c): Stress distribution near the edges of the condylar region. Cause: Implant mismatch
3.	Figure 7.20: Asymmetric stress patterns similar to Figure 7.22.2 (a), displaced vertically. Possible cause: Rotational malalignment.	Figure 7.22.2 (a): Asymmetric stress patterns caused by simulated rotational malalignment.
4.	Figure 7.12 (a) shows higher value of peak stress (1 nm or 8.3 MPa) at 60° flexion compared to Figure 7.14 (b) (0.45 nm or 3.75 MPa). However, this result is subject to the condition that the cadavers, and hence the loading conditions, were different.	Figure 7.22.2 (b): Higher value of peak stress compared to Figure 7.22.2 (a). This result is subject to the condition that the loading conditions were different.

From the results obtained in the cadaveric study and the comparison with results reported in literature, it can be said that the sensor can be used during total knee joint

replacement surgery to estimate the pressure distribution at the tibiofemoral interface. The sensor can be successfully used to correct small misalignments, which may not be obvious at first sight.

7.6 Conclusion

The cadaveric testing has proved that the sensor can be used as it is meant to be during the surgical process. It is practical and simple to use as it slides in and out of the tibial tray. After surgery, it can be washed, sterilized and prepared for use again.

This chapter also shows that the sensor is reliable for quantitative and statistical biomechanical studies of the tibiofemoral interface. The repetition of contact in the gait cycle (angles of flexion) can be seen from the regions registering high wavelength shift as well as definite regions of negative shift (very obvious in the case of cadaver 2). *The significance of using this pressure-mapping sensor for alignment during surgery can be understood by glancing at the flexion maps of cadaver 1 and 2 and the difference in the stress distribution that can arise.* Proper orientation of the implants can change stress distributions to a large extent, reducing subsequent complications of the knee, patella and the hip.

Chapter 8

Conclusions and recommendations

This chapter summarizes the development of the sensor, main conclusions drawn from this work and highlights some of the results obtained. It presents a few suggestions for future work.

8.1 Summary

It is necessary to have a better understanding of the factors contributing to the requirement of an early revision surgery to improve the longevity of total knee arthroplasty. The study of the contact pressure is crucial to the design of the prostheses, which influences the long-term survival of the implants.

Proper alignment of the implants during the surgery can improve their survival and prevent subsequent complications due to malalignment. A tibiofemoral pressure mapping sensor can provide a guide to the surgeons during the process of arthroplasty. For developing this sensor, both condylar grooves of the tibial spacer were covered with sensing points to observe the influence of the various loads. The intrinsic sensor lies below the contact surface and any deformation affecting the sensor can be monitored. Also, the fibre sensor quality does not degrade with repeated measurements. One of the advantages of this sensor is that it can be contoured for different designs of the prostheses.

The sensor developed in this research work gives a measure of the actual impact on the polymer insert (tibial spacer), as it is inbuilt unlike other surface mounted pressure mapping systems. As the whole sensor is covered with medical grade PMMA, it is suitable for any in-vivo studies as well.

To fabricate the sensor, a sampled chirped grating was designed to determine the magnitude and location of the load simultaneously. The grating was then embedded in fibre-reinforced composite to give the sensor. This sensor conforms to the shape and curves of the condylar grooves in the tibial spacer. The grooves were thinly layered with silicone rubber and the embedded sensors were attached on both sides. The whole sensor was coated thinly with a viscous mixture of PMMA.

The spread of the force was studied by recording the behavior of the sub-gratings in the condylar grooves. The wavelength shifts of the various sub-gratings generate a map of the pressure distribution, which was displayed with Matlab.

The pressure was calculated from the applied load and the estimated contact area. The contact area was approximated using Hertzian mechanics for a convex solid body of one material (femoral component) in contact with a concave solid body of another material (PMMA tibial spacer). An approximate theoretical model was derived.

The pattern generated by the tibial spacer matches those of the damaged knees. It identifies that the spread of the force lies beyond the actual contact area. This property could only be predicted by FEA and not with the existing sensors. Though the immediate pressure impact is on the contact area only, long-term degradation of the tibial spacer depends on the entire effect as shown by the sensor. This fact should be taken into account when the prostheses are designed. Another advantage of fibre-optic sensors is the fact that they can have a lifetime of many years. This property can be exploited to carry out survivability studies of the prostheses. Any permanent deformation of the PMMA (or similar polymer) can be monitored.

Assuming that the interrogation system used has an accuracy of 15 pm and a resolution of 1 pm, it can be concluded that the sensor has an accuracy of 0.125 MPa with a sensitivity of 121 pm/MPa. The sensor resolution is about 8 KPa.

8.2 Limitations

The failure of this sensor has not been discussed so far because any physical failure of the sensor can be overcome by manipulation of the design parameters; there is a trade-off between the range and sensitivity. The inherent failure lies in the wavelength spacing between the sampled sub-gratings. It may not be easy to distinguish overlapping peaks unless specific peak resolution techniques or calculation are used. This would saturate the range of the sensor between 0.125 MPa – 16 MPa. However, a failure test was done with a vertical load of 300 N applied with the force gauge on to the femoral implant for 5 seconds. The spectra of the affected sub-gratings shifted beyond 3 nm but

even after repeating the 300 N loading twice, the sensor regained the original reflected spectrum. Further testing with the sensor showed no change in the sensitivity or response.

The resources available, the phase mask and interrogation system, have resulted in the final optimized design of the sensor. A chirped phase mask with a smaller chirp rate would allow the user to squeeze more sub-gratings along the length of the fibre but it would also reduce the free bandwidth of the sub-gratings. Theoretically, if a demultiplexing interrogation system is available, all the sub-gratings can be fabricated on one strand of optical fibre.

8.3 Conclusions

The following conclusions can be drawn from the results obtained during this research. The embedded sampled chirped grating is useful as a distributed sensor. It provides a simple method to multiplex a large number of gratings in a couple of centimetres. The sampling generated from a linear chirp allows the correlation of the physical position of the sub-gratings with the spacing in the reflected wavelengths. An array of sampled chirped gratings creates a distribution of sensing points that can be used as a pressure map. Incorporating this embedded sampled chirped grating array into the tibial spacer provides a sensor that can map the pressure distribution at the tibiofemoral interface.

The shape of the knee joint hinders the use of standard pressure-mapping systems across the interface. The requirements of a sensor include range, flexibility, and

biomedical safety for in-vitro and in-vivo use. This pressure-mapping sensor is suitable for in-vivo use to correct malalignment during total knee arthroplasty. It can also be used, in-vitro, to study the deformation of the polymer tibial implant. Long-term studies of the tibial spacer can predict the degradation, which is one of the main causes of failure of total knee arthroplasty. The density of sensing points is created by sampling a chirped grating, and can be increased. The concept of the sensor can be applied to other complicated orthopaedic interfaces across the hip and shoulder, where knowledge of the pressure distribution is required.

8.4 Recommendations for future work

It is clear that the sensor developed is a viable spatial pressure probe. Further work could be done to improve its performance and application. Possible areas to explore include:

1. As in the case of the tibial spacer, the sensor can also be used for studies of the acetabular socket component in the hip joint [101] and the glenoid component in the shoulder joint [102].
2. The array of sampled chirped gratings can be explored for other areas of ergonomics and pressure mapping.

3. The sampled chirped grating also has the potential to measure the translation of an object over it. The interrogation system should be fast to detect the spatial resolution of a few millimetres.

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Software used:

IFO_Gratings (Integrated & Fiber Optical Gratings Design Software) Version 4.0,
Optiwave Corporation, Canada.

JL Analyzer, AutoFEA Engineering Software Technology, Inc., California, USA.

Matlab 6.1, The Mathworks, Inc.

APPENDIX 1

Fibre optic glossary

The optical fibre is a circular waveguide with a core and cladding. Light is guided in the core.

The chirped grating is a periodic grating in the core of the fibre that reflects a range of wavelengths. It acts as a filter that reflects back specific matched wavelengths. This “matching” is dependent on the range of periods in the grating.

The reflection spectrum contains the matched wavelengths.

The sampled chirped grating is a selective removal of parts of the chirped grating to make it discontinuous.

The reflected spectrum shows many peaks or bands.

When a load is applied to one section of the sampled chirped grating, only the strained sub-gratings will show a shift in their central wavelength.

The effect of a load on an array of sub-gratings; the sub-gratings affected by the stress distribution will show proportional wavelength shift.

APPENDIX 2

GLOSSARY OF TERMS RELATED TO THE KNEE JOINT

(Source: <http://edheads.org/activities/knee/kn-glossary.htm#t1>)

Accessed: 7 November 2004

Anterior:

Closer to or at the front of the body.

Anterior Cruciate Ligament (ACL):

The ligament that connects the tibia to the femur at the center of the knee. Its function is to limit rotation and forward motion of the tibia.

Articular Cartilage:

The specific cartilage that covers the moving surfaces inside the knee such as the tibia and the femur, as well as the underside of the patella.

Cartilage:

A smooth material that covers bone ends at a joint to cushion the bone and allow the joint to move easily without pain.

Collateral Ligaments:

Ligaments that run along the sides of the knee and limit sideways motion.

Condyle:

A rounded projection at the end of a bone that anchors muscle ligaments and articulates with adjacent bones.

Femur:

The thigh bone or upper leg bone.

Hamstring Muscles:

The muscle group located on the back of the thighs; they allow the knee to flex, the thigh to extend and the leg to be drawn inward.

Intramedullary Canal:

The canal that runs up the center of the femur.

Lateral:

Farther from the midline of the body (near the side).

Ligaments:

Elastic bands of tissue that connect bone to bone.

Medial:

Closer to the midline of the body (near the middle).

Meniscus:

Pads of cartilage that further cushion a joint, acting as shock absorbers between two bones. Meniscus can be found on both the lateral (on the side) and medial (near the middle) side of the knee joint.

Osteoarthritis:

The most common type of arthritis affecting the knee. It is a chronic disease and is characterized by destruction of cartilage, overgrowth of bone, bone spur formation and impaired function. This type of arthritis occurs when bone rubs against bone and occurs in most people as they age.

Osteophytes:

Abnormal projections of bone, also known as bone spurs. Usually caused by increased stress on the ends of the bones.

Patella:

The kneecap; a flat triangular bone located at the front of the knee joint.

Patella Femoral Arthritis:

Arthritis that is primarily focused around the kneecap (patella) and femur (thigh bone).

Patellar Ligament:

This ligament helps secure the patella over the front of the knee joint.

Primary Total Knee Replacement:

A "Primary" Total Knee Replacement refers to the first time a patient receives a knee replacement. The surgeon alters the femur, tibia and patella and fits those bones with prosthetic components.

Posterior:

Closer to or at the back of the body.

Posterior Cruciate Ligament (PCL):

The ligament located just behind the anterior cruciate ligament. It limits the backward motion of the tibia.

Quadricep Muscles:

The muscle group located on the front of the thighs; they extend the legs.

Rheumatoid Arthritis:

An inflammatory disease that involves the lining of the joint (synovium). The inflammation generally affects the joints in the hands and feet and tends to occur equally on both sides. Over time, cartilage and bone becomes eroded and the joints become very deformed.

Synovial Membrane:

This membrane produces lubricating fluid (synovial fluid), which contributes to the smooth movement of the knee.

Tendons:

Tough cords of tissue that connect muscles to bone.

Tibia:

The shin bone or the larger bone of the lower leg.

Valgus:

An abnormal position in which part of a limb is twisted outward away from the midline, opposite of varus. Also known as knock-knee.

Varus:

An abnormal position in which part of a limb is twisted inward toward the midline, opposite of valgus. Also known as bowleg.

APPENDIX 3

This appendix determines the location and expected wavelength shift when the point of loading is not exactly on any sub-grating but in between two sub-gratings identified by w_{max1} and w_{max2} . The Matlab program is given below. The inputs are w_{max1} , w_{max2} (central wavelengths of adjacent sub-gratings) and $max1$, $max2$ (corresponding wavelength shifts). The output is the location of the load. The second part of the program gives the expected maximum wavelength shift at the load location.

```
% Input wavelength shifts, max1 and max2, and central wavelengths, wmax1 and
wmax2%

sum = max1+max2;

num = abs(wmax1-wmax2);

displacement = num*max2/sum;

if wmax1 < wmax2

    location = wmax1 + displacement

elseif wmax1 > wmax2

    location = wmax1 - displacement

end;

a = abs(wmax1-location);

b = abs(wmax2-location);

A = [a b];

B = [max1 max2];
```

plot(A,B)

%This plot generates a straight line, which when extrapolated gives the y-intercept as the maximum wavelength shift%

APPENDIX 4

This appendix calculates the values of E_x , E_y and E_{fx} for a four layer unidirectional composite stack with the top layer at 90 degrees. The following Matlab program uses the material properties of the fibre-reinforced composite prepreg as the inputs.

%Input E1, E2, G12, (in GPa), ν_{12} , THETA as X%

E1 = 42;

E2 = 15;

G12 = 2.3;

ν_{12} = 0.227;

ν_{21} = (ν_{12} *E2)/E1;

DEN = 1-(ν_{12} * ν_{21});

Q11 = E1/DEN;

Q12 = (ν_{12} *E2)/DEN;

Q21 = Q12;

Q22 = E2/DEN;

Q66 = G12;

Q = [Q11 Q12 0; Q21 Q22 0; 0 0 Q66];

%For a certain angle of orientation THETA or X the transformations are as follows%

X = pi/2;

q11 = Q11*(cos(X)^4) + Q22*(sin(X)^4) + 2*(Q12+2*Q66)*(cos(X)^2)*(sin(X)^2);

$$q_{12} = (Q_{11} + Q_{22} - 4*Q_{66}) * (\cos(X)^2) * (\sin(X)^2) + Q_{12} * ((\cos(X)^4) + (\sin(X)^4));$$

$$q_{22} = Q_{11} * (\sin(X)^4) + Q_{22} * (\cos(X)^4) + 2 * (Q_{12} + 2*Q_{66}) * (\sin(X)^2) * (\cos(X)^2);$$

$$q_{16} = (Q_{11} - Q_{12} - 2*Q_{66}) * (\cos(X)^3) * (\sin(X)) - (Q_{22} - Q_{12} -$$

$$2*Q_{66}) * (\cos(X)) * (\sin(X)^3);$$

$$q_{26} = (Q_{11} - Q_{12} - 2*Q_{66}) * (\cos(X)) * (\sin(X)^3) - (Q_{22} - Q_{12} -$$

$$2*Q_{66}) * (\cos(X)^3) * (\sin(X));$$

$$q_{66} = (Q_{11} + Q_{22} - 2*Q_{12} - 2*Q_{66}) * (\cos(X)^2) * (\sin(X)^2) +$$

$$Q_{66} * ((\sin(X)^4) + (\cos(X)^4));$$

$$q_{21} = q_{12};$$

% When X is 0 or 90, then q16 and q26 are zero. Thickness in mm%

$$P = [q_{11} \ q_{12} \ q_{16}; \ q_{21} \ q_{22} \ q_{26}; \ q_{16} \ q_{26} \ q_{66}];$$

$$t = 600e-003;$$

$$z_0 = 300e-003;$$

$$z_1 = 150e-003;$$

$$z_2 = 0;$$

$$z_3 = -150e-003;$$

$$z_4 = -300e-003;$$

% Here we assign the stacking sequence%

$$Q_1 = P;$$

$$Q_2 = Q;$$

$$Q_3 = Q;$$

$$Q4 = Q;$$

$$A = P*(-150e-003) + Q*(-450e-003);$$

$$B = 0.5*(Q1*(z1^2 - z0^2) + Q2*(z2^2 - z1^2) + Q3*(z3^2 - z2^2) + Q4*(z4^2 - z3^2));$$

$$D = 0.333*(Q1*(z1^3 - z0^3) + Q2*(z2^3 - z1^3) + Q3*(z3^3 - z2^3) + Q4*(z4^3 - z3^3));$$

$$Ast = \text{inv}(A);$$

$$Bst = -(\text{inv}(A).*B);$$

$$Cst = B.*\text{inv}(A);$$

$$Dst = D - (B.*\text{inv}(A).*B);$$

$$Aa = Ast - (Bst.*\text{inv}(Dst).*Cst);$$

$$Ex = 1/(t*Aa(1,1));$$

$$Ey = 1/(t*Aa(2,2));$$

$$Da = \text{inv}(Dst);$$

% Also, it may be noted that the stacking sequence influences B and D (in this case, D, as the terms of z do not cancel).%

% Efx is the flexural modulus%

$$Efx = 12/(t^3 * Da(1,1));$$

APPENDIX 5

This appendix shows a comparison of the theoretical model and the experimental wavelength shift. The Matlab program uses the geometrical dimensions and the material properties of the femoral implant and the tibial spacer sensor as input. It calculates the contact area between the two bodies. It also calculates the values of wavelength shift based on the analytical model and plots the values with the experimental data.

%THEORETICAL

D1=7.6e-002; %diameter

D2=8e-002;

v1=0.32; %Poisson's ratio

v2=0.3;

E1=107e+009; %Elastic modulus of femoral implant

E2=3e+009; %Elastic modulus of PMMA

Kd=D1*D2/(D2-D1);

Ce=((1 - v1^2)/E1) + ((1 - v2^2)/E2);

sig=5:1:50; %arbitrary chosen values to match experiment

a =0.721*((Kd*Ce)^0.333).*((sig).^0.333);

plot(sig,a);

b1=3.9e-004:0.004:4.1e-004; %Equal to a when load nearly zero. Evaluated from

plot(sig,a).

%The value for rubber layer thickness is 1 mm and for composite layer is 600 um (4 times 150 um).

shift = 1550e-009*0.78.*sig./(pi.*b1.*a*(1/600e-006)*((Er*1e-003)+(Efx*600e-006)));

pres = sig./(pi.*a.*b1);

%EXPERIMENTAL

M=[5 10 15 20 25 30 35 40 45 50];

a2 =0.721*((Kd*Ce)^0.333).*((M).^0.333);

b4=3.9e-004:0.02:4e-004; %Equal to a when load nearly zero. Evaluated from
plot(sig,a).

Mpres = M./(pi.*a2.*b4);

Mshift =[0.14 0.24 0.36 0.44 0.5 0.6 0.66 0.72 0.8 0.88];

Mshift1=[0.18 0.28 0.38 0.46 0.54 0.6 0.68 0.74 0.8 0.88];

Mshift2=[0.18 0.26 0.34 0.44 0.50 0.58 0.66 0.72 0.80 0.84];

plot(pres,shift,Mpres,Mshift.*1e-009,Mpres,Mshift1.*1e-009,Mpres,Mshift2.*1e-009)

APPENDIX 6

This appendix gives the method to plot the pressure maps with the measured wavelength shifts using Matlab software. It generates a three dimensional colour map with the X, Y axes denoting the positions of the sub-gratings and the Z axis showing the effect of the load in terms of wavelength shift. Similar maps can also be obtained in terms of MPa.

```
[X,Y] = meshgrid(-10:5:10, -12:6:12);

%for 30 degrees%

Z2 = [ 0 -0.1 -0.16 0.44 0.02; 0.12 0.06 0.12 0.1 0.08; 0.08 0.02 0.12 0.04 0.1; -0.08 -0.06
0.02 -0.02 0; 0.02 -0.1 -0.28 -0.1 0;];

%for 40 degrees%

Z3 = [ 0 0.02 -0.04 0.08 0.08; 0 0.06 0.24 0.1 0.08; 0.08 0.04 0.12 0.04 0.08; -0.04 -0.08 -
0.04 0 0; -0.08 -0.16 -0.16 -0.04 0;];

%for 50 degrees%

Z4 = [0 0.2 0.06 -0.24 -0.06; 0 -0.12 -0.16 -0.12 0.04; -0.04 -0.04 -0.08 0.04 0.04; -0.04 -
0.04 -0.04 0 0; -0.12 -0.12 -0.18 -0.18 0];

%for 60 degrees%

Z5 = [0 0.84 1.04 0.48 -0.3; -0.04 0.04 0.08 -0.16 -0.04; 0.04 0.08 0.08 0.04 0.08; 0 -0.08
0 -0.04 0; -0.18 -0.12 -0.12 -0.18 0];

subplot(2,2,1)

surf(X,Y,Z5')

colormap(jet(20))
```

shading interp

axis([-10 10 -12 12 -0.3 1.05 -0.3 1.05])

colorbar

axis tight

axis equal

subplot(2,2,2)

surf(X,Y,Z4')

colormap(jet(20))

shading interp

axis([-10 10 -12 12 -0.3 .6 -0.3 .6])

colorbar

axis tight

axis equal

subplot(2,2,3)

surf(X,Y,Z3')

colormap(jet(20))

shading interp

axis([-10 10 -12 12 -0.3 .6 -0.3 .6])

colorbar

axis tight

axis equal

```
subplot(2,2,4)
```

```
surf(X,Y,Z2')
```

```
colormap(jet(20))
```

```
shading interp
```

```
axis([-10 10 -12 12 -0.3 .6 -0.3 .6])
```

```
colorbar
```

```
axis tight
```

```
axis equal
```

APPENDIX 7

This appendix illustrates the data collection and interpretation. The readings are tabulated for separate gratings. In the correct layout sequence, the readings give the discrete map *directly*.