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A needle-tip embedded fibre optical force sensor
for tissue identification

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Doctor of Philosophy

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Abstract

The identification of biological tissue types during needle insertion is subject to a lack of precise monitoring techniques, especially in an intensive magnetic resonance environment. One approach to improve this situation is to attach optical sensors to the surgical needle. In this research, a temperature-compensated fibre optical tip force sensor based on the Fabry-Perot Interference (FPI) principle is developed and embedded into the tip of surgical needles. This was designed for tissue identification which it fulfilled via analysing the tip force of the needle during insertion.

A novel structure for FPI signal temperature-compensation is proposed, based on the provision of another FPI sensor which is only sensitive to temperature change. After sensor fabrication, two FPI sensors are embedded into the needle’s tip according to the designed structure. The experiments with varying temperature and force demonstrate the designed sensing needle could detect temperature and force signals. The system signal processing is based on interference intensity modulation where the FPI light intensity has a cosine relationship with applied force. An algorithm is proposed to turn light intensity into its intensity phase signal which is linear with applied force variation.

The sensing needle is then calibrated via a temperature chamber and commercial force sensors. The results show that the sensing needle successfully overcomes the influence of temperature and captures the applied force at its tip. In addition, as the reference FPI sensor could provide temperature information, the sensing needle is also calibrated in regard to temperature sensing.

An optical circuit for the FPI sensors is built and its size is optimised to fit a control box, with connection terminals to the FPI sensor, a portable laser source, and a computer. The
entire system is portable, economical and practical for Minimally Invasive Surgery (MIS). A Graphical user interface (GUI) is developed to display tip force and temperature information.

Tissue identification is the main application of the sensing needle. To achieve this objective, a series of tissue experiments are conducted on phantom tissue and porcine tissues. A needle insertion platform is designed for the tissue insertion experiments. A database of tip forces versus tissue types is then obtained based on the tissue insertion experiments for use in tissue identification. In addition, the factors influencing tip force during needle insertion are also investigated. Based on the above results, tissue identification experiments are performed, the results of which indicate the possibility of tissue identification in specific body locations.

In vivo experiments on mice are designed and conducted to further verify the concept of temperature compensated tip force sensing and tissue identification. The results of these experiments confirm the ability of the tip force sensing needle to identify tissue types, though some internal organs of mice are difficult to be penetrate and detect. An improved tip force sensing needle is fabricated for application to epidural space identification. The needle insertion experiments on porcine spinal samples show that the improved sensing needle could detect the epidural space effectively.
Dedication

To my family with love
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Abbreviations

FPI                Fabry-perot interference
MIS                Minimally Invasive Surgery
MRI                Magnetic Resonance Image
FBG                Fibre Bragg grating
CMOS               Complementary Metal-Oxide-Semiconductor Transistor
MEMS               Micro Electro Mechanical Systems
PCF                photonic crystal fibre
PMMA               Poly methyl methacrylate
CT                 Computed Tomography
LOR                Loss of Resistance
ID                 Inner diameter
OD                 Outer diameter
SM                 Single mode
Symbols

$A_1$  
Transmission loss factor on surface 1 due to the surface imperfections

$\alpha$  
The loss factor of the cavity induced by diffraction in the air cavity

$L$  
FPI cavity length

$R_1$  
Power reflection coefficients of the reflected mirror

$R_2$  
Power reflection coefficients of the reflected mirror

$\lambda$  
Light wavelength

$\Delta L$  
The optical path-length difference between two reflected light beams

$I_{FP}$  
The light intensity of FPI

$n_{air}$  
The refractive indice of the two media forming the boundary

$L_{ini}$  
Initial change of FPI cavity length

$l_t$  
Intensity of reflected light

$\varepsilon_x$  
Strain in axial direction

$d$  
Distance between the two mirrors at zero load

$\Delta d$  
The change in cavity length

$F_p$  
Needle penetration force

$F_s$  
The force required to deform the tissue at the tip of the needle

$F_f$  
The friction force along the needle

$F_c$  
The force required to cut through the tissue

$E_R$  
The long-time modulus of tissue stiffness

$x_p$  
The tissue deformation at the needle tip

$\tau_1$  
Intrinsic time scales characterizing the nature of the stress relaxation and creep

$\tau_2$  
Intrinsic time scales characterizing the nature of the stress relaxation and creep

$A_t$  
The contact area between the needle tip and the soft tissue

$G_c$  
The fracture toughness required to advance the needle tip per unit area of crack
$E_T$  The tissue stiffness per unit length
$V$  Insertion velocity
$z$  The elastic deformation of surface asperities
$\sigma_0$  The stiffness coefficient of the microscopic deformations during the pre-sliding displacement
$\sigma_1$  The damping coefficient associated with $z$
$\sigma_2$  The viscous damping coefficient
$\phi$  The intensity phase
$I$  The measurement of the light intensity
$F$  The axial loading force
$G$  Needle gauge
$\Delta d_F$  The FPI cavity length change induced by both force change and temperature change
$f_c(\Delta F)$  The FPI cavity length change induced by both force change
$f_c(\Delta T)$  The FPI cavity length change induced by temperature change
$\Delta d_f$  The effective cavity length change induced by force loadings
$\Delta d_R$  The cavity length change of the reference sensor due to temperature variation
$k$  The coefficient accounting for the difference in the temperature influence
$I_n$  Light intensity
$\Delta \phi_F$  The measured phase change of the force sensor
$\Delta \phi$  The measured phase change
$k$  The coefficient used to calibrate the difference in the thermal influence of the two sensors
$\alpha$  Coefficient of intensity phase-force relationship
Chapter 1

Introduction

1.1 The need for force sensors at the tip of medical instruments

In the medical field, force sensing feedback is critical to surgeons to detect tissues’ physical properties, so that they may apply appropriate force during surgical operations. Unfortunately, force sensing is severely constrained in minimally invasive surgeries (MIS) [1]. For example, during spinal surgery, surgeons need to tap the probe into the pedicel bone [2]. They monitor the procedure now only through 2-D image techniques and feeling with their hands. Another example is retinal microsurgery, which requires micro forces that are hard to sense with human hands; applied forces are very important for these important but fragile human organs. Currently, the most successful applications, the ZEUS and da Vinci surgical systems, provide only limited visual feedback but no force feedback [3].

Some experimental studies[4, 5] have shown that, without force feedback, tissue trauma and unexpected damage to healthy organs increase during surgery. When force feedback is incorporated into teleoperated systems can reduce robotic force by 30% to 60%, peak
forces by a factor of 2 to 6, operating time by 30% and error rates by 60% [6]. This shows the need for force sensors to be integrated into surgical instruments.

Force sensing is one of our main means of interacting with the environment. A force sensor can detect contact between itself and an object, and measures static or dynamic force magnitudes [7-9]. Force sensor are widely used in many areas [7], such as in the agriculture and food industry [10], the prosthetics field [11], environmental studies [12], in the robotic industry especially [13-15], and in the biomedical field [16, 17].

From 1980-1984, Harmo made a review of research into force sensors and projected that they had huge potential and many applications in the future [18-20]. However, it was not until the late 20th century that researchers started to pay more attention to force sensing technologies after major advances in computing and robotics technologies. The possible reasons for this situation are examined by Tiwana, et al [21].

There are various positions in a medical instrument that could integrate force sensors, such as on actuation mechanisms driving a joint, instrument shafts outside or inside the patient’s body, and instrument tips [1]. In the past, force sensors were mainly attached to instruments outside the patient’s body as the sensors were large [22]. However, if detection is focused on small forces, the results will be significantly influenced by trocar, inertial and friction forces. Therefore, the best solution is to integrate force sensors into tool tips [23].

Magnetic resonance imaging (MRI) devices produce strong electromagnetic interference, and they are usually employed in MIS to help surgeons guide the operation procedure. Conventional electric sensors cannot function normally in such an environment, but optical sensors are immune to electromagnetic interference and have a high degree of sensitivity. Optical sensors have been studied for over three decades, and a fair amount of theory and knowledge has been accumulated. Fibre optical sensors are sensitive to various parameters such as strain, current, voltage, temperature, force, and electric and magnetic field. The ability to be miniaturised, their light weight, immunity to electromagnetic interference, and ability to measure distance are also advantageous, which makes them a good alternative candidate for applications in extremely harsh
environments, such as those that require liquid temperature and pressure measurement, those that have high temperatures or intense electromagnetic interference, etc.

1.2 Research motivation

The main motivation for this research was to design a practical fibre optical force sensor and integrate it into the tip of surgical needles, with the capability of identifying tissue types during in vivo needle insertions.

Currently there are optical sensors which are suitable for embedding into surgical instruments, such as fibre bragg grating (FBG) force sensors and Fabry-Perot interference (FPI) force sensors, which have the required small size and immunity to MRI influence. Despite their advantages, a number of challenges have been revealed, such as the nonlinearity of the interference light intensity signals, and temperature change influence, most of which cannot be solved through the methods that are currently available. New approaches must, therefore, be found to resolve these two obstacles for the development of integrated tip force sensor’s in needle applications. The factors that influence tissue identification also need to be investigated and clarified. This research has undertaken to ensure the development of temperature compensated tip force sensing for needle insertions in tissue identification applications.

1.3 Objectives and scope

To provide reliable tip force information to surgeons, the objective is to develop a miniature fibre optical force sensor which is suitable for integrating into the tip of medical instruments that has a good temperature influence compensation. The goal of the research is the design, fabrication and experimentations of a temperature-compensated fibre optical tip force sensing needle for the purpose of tissue identification during in vivo needle insertions.

In order to design a practical FPI force sensor, the following sub-objectives need to be achieved:

- **Tip force sensor design** – In order to design an FPI sensor that is suitable for embedding in a needle tip, the FPI sensor’s working principle and the challenges that
need to be overcome must be fully understood. Firstly, the concept of tip force sensing and temperature influence were validated via mathematical calculation, simulation, and simply designed insertion experiments. Then the miniature FPI sensor design was proposed with its temperature compensation solution. A fibre optical circuit and signal processing algorithm also were also achieved.

- **Sensor calibration** – To enable the tip force sensing needle with a force sensing ability, the fabricated needle must be calibrated. Instrumentation for calibrating temperature and force character were prepared. A temperature chamber was designed for temperature experiments which could provide an environmental temperature of up to human body temperature. Different commercial force sensors were prepared for to calibrate the sensing of force, and their relevant working platforms were also designed.

- **Sensing needle capability characterization for tissue identification** – To ensure the ability of the needle to identify tissue, the performance of the sensing needle in different working conditions must be studied, such as needle advance rate and displacement, automated and manual modes of needle insertion, needle-tissue frictional force, needle size and tip shape, and tissue types, etc. An insertion platform for locating tissue and driving the needle was designed for the insertion experiments. Phantom human tissues from SynDaverTM Labs, USA, were prepared for the insertion experiments, as well as porcine tissues from the supermarket.

- **In vivo experimental validation** – In vivo experiments were very important to promote the validation of the designed sensing needle. Mice with tumours were used in this experiment. In vivo tissue identification and tumour localization were conducted by manual insertion. For this application, the tip force sensing system was compressed into a portable system.

- **Epidural space identification** – Epidural space identification is a critical issue where a lack of tip force sensing makes surgery more time-consuming and dangerous. The tip force sensing needle was modified in structure to resolve this problem precisely. For experimental tissue, porcine spine and spinal cords were bought from the supermarket.
1.4 Contributions

The fibre optical tip force sensing needle and its tissue identification ability are the focus of this research. The scientific and application contributions can be outlined as follows:

- The design of the tip force sensing needle

A tip force sensing needle is developed for tissue identification. It provides a solution to clinicians in terms of force sensing at the needle tip during surgery. Contributions include:

  - Tip force concept and temperature influence are validated from numerical calculation and simulations
  - Temperature compensation of optical signals is achieved via adding a reference FPI sensor. The compensation structure is designed for needle applications
  - Fabrication of the FPI sensor and its embedding into the needle tip
  - Experimental characterisation and validation of temperature compensation

- FPI signal processing and system

The FPI light intensity signal has a special cosinoidal relationship with force loadings. Thus, it is very important to discover a signal processing algorithm to overcome issues brought about by this relationship. The contributions of this signal processing and system are:

  - Optical circuit for temperature compensation design
  - An algorithm to change light intensity to its intensity phase which has a linear relationship with force loading change

- Ex vivo and in vivo tissue identification

The tissue identification capability of the sensing needle was characterized, and then tissue identification experiments were carried out. A modified needle design was
developed for epidural space identification. The needle was then further assessed via in vivo mice experiments on mice. The experimental outcomes include:

- Investigation of the factors that influence tissue identification
- Tissue identification exploration of phantom tissue and porcine tissue
- The fabrication and assessment of the modified tip force sensing needle for epidural space identification
- Further validation of tissue identification in in vivo experiments on mice

1.5 Thesis outline

The research work is documented in the order of the objectives described in the previous section. The thesis contains seven chapters in total, and the following gives a brief description of the content of the thesis:

Chapter 2 presents a literature review on the areas relating to fibre optical sensors. The review reveals problems with existing medical applications. A brief introduction to electrical force sensors and fibre optical sensors is presented to justify the choice of medical instrument tip haptic sensing. FPI sensors are introduced in detail, including the working principle and current temperature compensation solutions. Needle insertion researches are also reviewed, with regard to modelling and applications. Following this review, the contributions of the research in this thesis are outlined.

Chapter 3 describes the design of the optical force sensor. It was embedded into the tip of a surgical needle for tip force sensing during surgery. The concept was first tested on a preliminary design without temperature compensation. The temperature compensated tip force sensing needle was then fabricated and calibrated using commercial electrical force sensors, followed by initial tissue experiments for demonstration. The temperature sensing function of the design was also calibrated for the sensing application.

In Chapter 4, the tip force sensing needle is further calibrated for its tissue identification capability. Two kinds of tissue, phantom human tissue and porcine tissue, were used in experiments. The influence of various factors was investigated, and data fields of forces
versus types of tissue are established for tissue identification, based on phantom and porcine tissue insertions. Using this database, this design is validated via tissue identification experiments.

Chapter 5 described the in vivo validation of the tip force sensing needle. Repeated needle insertions were carried out on live and dead mice for the purpose of tissue identification.

Chapter 6 describes a modified structure design for the tip force sensing needle for a specific application, epidural space identification. An 18G (G: Gauge, needle gauge for comparing outer diameter of medical needles) tip-force sensing needle was designed, and characterized to assess the feasibility of epidural space identification. A 16G tip force sensing epidural needle was then fabricated and validated experimentally.

Finally, Chapter 7 provides a conclusion for the overall research work. The contributions of the research are presented and likely directions of future work are provided.
Chapter 2

Literature review

2.1 Needle insertions

Needle insertion is necessary in a variety of procedures such as interventional radiology [24], neurosurgery [25], brachytherapy [26], regional anaesthesia [27], biopsies [28], drug delivery [29] and blood sampling [30]. It also represents one of the least invasive ways to access the internal organs of patients [31]. An example is epidural anaesthesia, a regional anaesthesia used for pain relief that can be performed at different locations along the spine depending on the surgery [32]. In England alone, about 280,000 epidural anaesthesia are performed yearly within the National Health Service [33].

To assist needle insertion, image-guidance techniques, such as ultrasound, computed tomography (CT), and magnetic resonance imaging (MRI), are adopted. However, it is still difficult to avoid human error and damage critical organs. For example, epidural space identification is heavily relied on the clinician’s manual skills, namely, “loss of resistance” (LOR) [34], where air or physiological liquid is injected during insertion via a syringe. The fluid leaks little when crossing the preliminary layers, but empties quickly after reaching the epidural space due to the sudden drop in pressure in the epidural space [35]. Even with the help of image-guidance techniques, the procedure is still challenging,
and may cause damage to the spinal cord if the clinician continues to push the needle after having reached the epidural space [36]. For this reason, clinicians often become hesitant to advance the epidural needle due to the unknown position of the needle tip and the loss of feeling for the tip force when the needle is inserted deeper because of increase of the frictional force between the needle and surrounding tissues [37], which results in a more difficult and time-consuming procedure.

In other procedures, tumours, nerves, vessels, and hard organic structures buried inside other tissues may not be easily detected by medical imaging systems [38], but a tactile sensor may be able to detect them. Tactile sensing can be undertaken in real time, much faster than vision based techniques, and can be more sensitive than a surgeon’s haptic sense [39].

To enhance the precision of needle insertion, three research directions have received considerable attention: the interaction between needle and tissue, the image navigation techniques, and needle force sensing. Of these, the first field is concerned mainly with modelling and simulations. The objective is to control the advancing direction of surgical needles to avoid damage to vulnerable organs such as viscera, vascular tissue, or nervous tissue. As biological tissue is opaque rather than transparent as are phantom tissues used in simulations, image guidance techniques are often used for needle procedures, such as ultrasonography, CT and MRI [40]. The force sensing of needle insertion helps the clinician understand what type of tissue the needle is touching and where the needle’s tip lies precisely, along with anatomical knowledge or medical image guidance. Most of the current force sensors are placed on the shaft or base of surgical needles [41, 42], which are subjected to frictional force along the needle.

2.1.1 Needle interaction with soft tissue - mathematical model

Needle interaction with soft tissue often results in sudden layer rupture, which could enable penetration of adjacent multiple tissue layers, putting sensitive organs in danger. It also increases needle placement error during surgery. Therefore, it is useful to study the mechanics of dynamic needle interaction with soft tissue.

The force acting on the needle, $F_p$, consists of three parts, and can be expressed as,
\[ F_p = F_s + F_f + F_c \]  

(2.1)

Where \( F_s \) concerns the force required to deform the tissue at the tip of the needle, \( F_f \) is the friction force along the needle, and \( F_c \) is the force required to cut through the tissue. These forces are concerned with any soft tissue that a needle is about to enter, as shown in Figure 2.1.

![Diagram showing needle interaction forces](image)

Figure 2.1 Needle-tissue interaction forces during needle insertion. (a) Puncturing the tissue at the contact point between the needle tip and the tissue surface. (b) Tissue cutting, including crack propagation, into the tissue in response to needle displacement. (c) Needle-tissue friction force applied tangentially to the needle shaft. [43]

The mechanics of needle insertion could be simplified to the standard linear solid model to predict the force-deformation response [43], where the tissue is modelled as a Newtonian damper and two Hookean springs, one in parallel and one in series, as shown in the following diagram.

![Simplified model diagram](image)

Figure 2.2 Simplified model to represent needle interaction with a viscoelastic tissue

At the stage of tissue boundary deformation, before tissue rupture, the puncture force \( F_p \) could be calculated by

\[ F_p = E_R X_p \frac{\tau_2}{\tau_1} \]  

(2.2)
where $E_R$ is the long-term modulus of tissue stiffness, $x_p$ is tissue deformation at the needle tip, $\tau_1$ and $\tau_2$ are intrinsic time scales characterizing the nature of stress relaxation and creep [44].

In the next stage of crack initiation, the puncture force $F_p$ could be expressed by

$$F_p = A_t G_c \frac{T_2}{\tau_1}$$

(2.3)

where $A_t$ is the contact area between the needle tip and the soft tissue, $G_c$ is the fracture toughness required to advance the needle tip per unit area of crack. It indicates that maximum puncturing force during needle insertion has no relationship to insertion velocity.

While the cutting force after the crack stage $F_c$, when the needle tip is completely inside of the soft tissue, could be estimated according to

$$F_c = \frac{4 \tan(\frac{\pi}{2}E_R^2)}{E_T}$$

(2.4)

where $E_R$ is the tissue’s long-term stiffness modulus, $E_T$ is tissue stiffness per unit length, and $\alpha$ is the angle of the needle tip. It shows that cutting force during deep insertion tends to be a constant value. While friction force, $F_f$, at this stage is related to insertion velocity $V$, given by

$$F_f = \sigma_0 z + \sigma_1 \dot{z} + \sigma_2 V$$

(2.5)

where $z$ is the elastic deformation of surface asperities, $\sigma_0$ is the stiffness coefficient of the microscopic deformations during pre-sliding displacement [43], $\sigma_1$ is the damping coefficient associated with $z$, and $\sigma_2$ is the viscous damping coefficient.

2.1.2 Latest developments of force sensing integrated needle applications

To study or improve needle placement in soft tissue, three types of needle locations for the force sensor were chosen, the tip, the shaft, and the base of the needle. For sensors embedded inside the needle, FBG and FPI are the main working principles used.
Sensor attached at the end of the needle

Among all needle sensor applications, 65.4% of the applications (about 42 out of 64 from 2010 to 2017) were this type. As sensor size is not restricted in this application, most of the sensors were based on electrical forces. Figure 2.3 shows a typical sensor placement for this type of needle application. Only a few studies were based on fibre optical FBG sensors or FPI sensors, as shown in Figure 2.4 and Figure 2.5, respectively.

Figure 2.3 Typical structure for a needle-end attached force sensor [45]

Figure 2.4 Schematic of FBG-based force sensor with a spinal needle [46]

Figure 2.5 FPI based force sensor for a needle application [47]
Force sensors at the base of the needle can only monitor a hybrid force of the cutting force at the needle tip and the friction force on the needle’s shaft. For deep tissue insertions, friction force could lead to unreliable force sensation on the needle’s tip. A typical insertion force signal is shown in Figure 2.6, where it is difficult to measure the real-time absolute force loaded at the needle’s tip, as needle shaft friction force increases significantly during insertion.

![Figure 2.6](image1.png)  
**Figure 2.6** Typical needle insertion force based on a sensor at the end of needle

- **Sensor embedded in the needle shaft**

![Figure 2.7](image2.png)  
**Figure 2.7** Typical FBG sensor embedded in needle shaft (a). Dimension of the dual force sensing instrument (in cm) (b). The cross-section view of the instrument shaft (c). Photo of the dual force sensing instrument [49]
FBG is one of promising techniques for embedding in a needle shaft. As an FBG sensor is very sensitive to lateral force changes, for the purpose of soft needle insertion navigation, it is employed for detecting needle bending for a shape-changing measurement. A typical design of the application is shown in Figure 2.7.

- **Sensor embedded at the needle tip**

Only a few applications embedded the force sensor at the needle’s tip, with FBG and FPI being the only available sensors [50, 51], as shown in Figure 2.8 and Figure 2.9, respectively. However, none of them have a temperature compensation design for in vivo needle insertion.

![Figure 2.8 FBG based force sensor embedded at needle tip [50]](image)

![Figure 2.9 FPI-based sensor embedded at needle tip [51]](image)

Apart from the above sensor applications, one design with sensors at the base of the needle, also has the capacity for measuring the force loaded at the needle tip, as shown in Figure 2.10.

![Figure 2.10 Tip force sensing using sensors at the end of the needle [37]](image)
2.1.3 Tissue identification by force sensing

Tissue identification in MIS has begun to attract attention in recent years. Only a few attempts were carried out in this area using force sensing instruments. Various special force sensing palpation probes were developed for minimally invasive tumour localization. Perri, et al. [52] developed a hand-held tactile-sensing probe to assist in intraoperative tumour localization, with an accuracy of 61%. Liu, et al. [53] proposed a force-sensitive wheeled probe for the localization of tissue abnormalities during minimally invasive surgery. Their results were effective and repeatable in regard to tumour localization, although a sensitivity threshold does exist.

![Diagram of force-sensing probe](image)

Figure 2.11 Current tissue identification needle applications [54, 55]
Tissue identification using force-sensing needles received attention due to the development of optical sensing techniques. FBG and FPI are the most promising sensing principles in terms of tip-force sensing applications. Two applications are shown in Figure 2.11. Beekmans, et al. [54] designed an FPI force sensing element mounted at the end of a needle with a 5 mm diameter for tissue identification, having some limitations such as large diameter and influence of sample debris. Carotenuto, et al. [55] embedded an FBG force sensor into an epidural needle for epidural space identification. Experiments on phantom organs showed advantages compared with current LOR approaches, but having limitations such as temperature influence.

2.2 Force sensor requirements in MIS

In minimally invasive surgery (MIS), surgical instruments with small diameter are used and gain access into a patient’s body through incisions of small diameter of about 3 - 15 mm [56]. Compared with traditional open surgery, MIS has many advantages, such as reduced operation time, less soft tissue damage, less pain and faster recovery [57]. These features and technological advances have significantly increase the development of these surgical techniques since the introduction of laparoscopic cholecystectomy - the first kind of MIS in late 1989 [58, 59].

However, this technique also brings difficulties for surgeons. It could endanger a patient’s safety if the needle is inserted incorrectly. Indirect visual perception and sensation of force are the main causes of this problem [60]. In the past 30 years, researchers have mainly focused on reducing the visual limitations [61-67], but few studies have looked at resolving a surgeon’s force sensing problem [60].

Due to MIS’s unique working environment; a magnetic resonance environment and very limited working area, force sensors should have specific requirements.

- **Miniaturization**

In MIS, operations are performed using surgical instruments to enter the patients’ body through several small incisions that about 1 cm in length [3]. According to the scale of access, the force sensors should be designed to be smaller than 1 cm in diameter. As they will be integrated into surgical tool tips, such as surgical needles or probes, the smaller
the diameter of the sensor, the less influence they will have on the operation of the surgical instruments.

- **Force sensing range**

As to the interaction force required with different tissue types, the force range should vary. For example, a 0 - 3 N force range would be sufficient for needle insertion into human organs such as kidney or liver [68-73], while cutting human muscle with a surgical knife requires at least 35 N [74]. Since all the force sensors are integrated or simulated outside the tissues, the actual peak amplitude of the tip force should be less than 3 N, taking into consideration friction force.

- **High sensitive and safety**

To detect forces in real-time, with different tissues precisely, force sensors should be sensitive to static and dynamic force and capable of a wide force sensing range at high resolution. Sensors should also be safe for surgical procedures, introducing no of electrical current that may cause injuries such as loss of output control or chemical reactions with substances in the patient’s body.

<table>
<thead>
<tr>
<th>Tissue resource</th>
<th>Tissues</th>
<th>Force range (N)</th>
<th>Deviation (N)</th>
<th>Insertion velocity (mm/s)</th>
<th>Needle information</th>
<th>Other information</th>
</tr>
</thead>
<tbody>
<tr>
<td>Live Porcine [69]</td>
<td>Liver &amp; Skin</td>
<td>3.73</td>
<td>0.59</td>
<td>As constant and as smooth as possible</td>
<td>18G 15.24 cm biopsy needle</td>
<td>Manual insertion</td>
</tr>
<tr>
<td></td>
<td>Liver</td>
<td>0.7</td>
<td>0.29</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Kidney</td>
<td>0.74</td>
<td>0.54</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Pancreas</td>
<td>0.83</td>
<td>0.28</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Liver &amp; Skin</td>
<td>1.89</td>
<td>0.36</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Liver</td>
<td>0.59</td>
<td>0.17</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Kidney</td>
<td>1.22</td>
<td>0.34</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Liver</td>
<td>2.304</td>
<td>0.8286</td>
<td>15</td>
<td></td>
<td>Robotic insertion</td>
</tr>
<tr>
<td></td>
<td>Skin</td>
<td>0.29</td>
<td>0.05 &amp; 0.06</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

| Bovine within 3 - 6 hours of death [70] | Skin | 0.34 | 0.05 & 0.07 | 3 | 18G 15.24 cm biopsy needle | Room temperature |
| Skin on lower | | | | | | |

Table 2.1 Needle insertion required force range
<table>
<thead>
<tr>
<th></th>
<th>arm and abdomen</th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Human volunteers</td>
<td>Skin on lower arm and abdomen</td>
<td>0.24</td>
<td></td>
<td>0.06</td>
<td>2</td>
<td>27G needle,</td>
</tr>
<tr>
<td>[71]</td>
<td></td>
<td>&amp;</td>
<td>0.23</td>
<td></td>
<td></td>
<td>45 &amp; 90 degree</td>
</tr>
<tr>
<td></td>
<td></td>
<td>0.27</td>
<td></td>
<td>0.06</td>
<td>19</td>
<td>angle tip</td>
</tr>
<tr>
<td></td>
<td></td>
<td>&amp;</td>
<td>0.26</td>
<td></td>
<td></td>
<td>-</td>
</tr>
<tr>
<td></td>
<td></td>
<td>0.8</td>
<td></td>
<td></td>
<td>2</td>
<td>30G, 45 &amp;</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>90 degree angle</td>
</tr>
<tr>
<td></td>
<td></td>
<td>1.4</td>
<td></td>
<td></td>
<td>19</td>
<td>tip</td>
</tr>
<tr>
<td>Vivo human skin[72]</td>
<td>Skin on lower arm and abdomen</td>
<td>6</td>
<td></td>
<td></td>
<td></td>
<td>φ0.3 mm</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>sharp-tip</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td></td>
<td>0.9</td>
<td></td>
<td></td>
<td></td>
<td>φ0.6 mm</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>sharp-tip</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td></td>
<td>0.46</td>
<td></td>
<td></td>
<td></td>
<td>φ0.3 mm</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>flat-tip</td>
<td>-</td>
</tr>
<tr>
<td>Porcine within 2</td>
<td>Heart</td>
<td>0.9</td>
<td></td>
<td>5</td>
<td></td>
<td></td>
</tr>
<tr>
<td>hours of death[73]</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>In vivo rabbit[73]</td>
<td></td>
<td>0.7</td>
<td></td>
<td>250</td>
<td></td>
<td>φ1.1 mm</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>tri-tip</td>
<td>2 kHz</td>
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<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>sample rate</td>
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</tr>
</tbody>
</table>

- **Low MR interference**

Due to the visual limitations of MIS, Magnetic Resonance Image (MRI) devices are usually used to monitor the surgical procedure. Therefore, the sensor should have the capacity to work within a strong MR environment.

- **No influence on the magnetic field and radio wave emitting devices**

The force sensors should not affect other surgical instruments, especially the MRI devices which are sometimes the only option for visual feedback during operations. This means force sensing technologies should not have a strong magnetic field of their own or emit radio waves.

**2.3 Overview of force sensing mechanisms**

The possible existing solutions for the micro force sensors can be grouped into the following five categories: capacitive, resistive, piezoelectric, strain gauge, and optical force sensing mechanisms [7, 21, 57, 75-77].
2.3.1 Traditional electric force sensors

- **Resistive and piezoresistive force sensors**

Resistive force sensing is the most common of the force sensing technologies. Resistive force sensors are simple to construct, reliable, have adjustable resolution, are maintenance free, and economical [78]. As one type of resistive force sensor, piezoresistive force sensors are based on changes in resistivity when applied force deforms it. The resistivity change is easy to observe in the form of voltage variation (or current) by fixing current (or voltage) in a simple resistive circuit. Metals such as nickel and platinum alloys, semiconducting materials, such as silicon or germanium, and elastomer are the most common materials used for piezoresistive force sensors [21, 79].

In 1986, Prudenziati, and Morten [69] made a review of the piezoresistive properties as strain-related transducers formulated since 1973 when piezoresistive strain sensitivity was first observed. The work mainly examined correlations between gauge factors, composition and structure of thick-film resistors, and change in strain sensitivity caused by various factors such as peak firing temperature and the nature of the materials. In the following decades, resistive and piezoresistive force sensors were often created in the form of strain gauges, as they were much smaller and had higher resolution, which was more suitable for the requirement of high integrated and sophisticated instruments.

- **Capacitive force sensors**

As with resistive force sensors, the capacitive force sensors are based on a normal capacitive circuit. Capacity change is obtained by a change in voltage when the distance between the inside plates of the capacitor changes due to applied force.

Capacitive sensors are very robust and can be used in high temperature and harsh environments [80]. They are also have good resolution and a wide dynamic range, and can be in a very small when utilizing the MEMS technology [21, 80]. They are very sensitive in detecting extremely small deflections of structures without any direct influence in temperature [81, 82]. However, capacitive sensors are usually limited to situations that require precise force measurement within a small range [1].
• **Piezoelectric force sensors**

Piezoelectric materials are also useful for force sensing technology. These materials can generate a voltage potential when deformation of the material occurs [1, 21, 83-85]. These sensors are highly sensitive even to small deformations, and very reliable as they do not require any electrical power input [84], and they also allow a greater range of applications compared to other sensors [1]. The piezoelectric effect can be found in natural crystals, such as quartz, artificially polarized ceramics, and certain polymers, such as polyvinylidene fluoride (PVDF) [86].

However, these sensors are used only for measuring dynamic force rather than static pressure, as the output signal decreases over time [1, 83, 84]. Another disadvantage is signal noise affected by temperature change, which could decrease the accuracy of the sensor if it has no temperature compensation [1].

• **Strain gauges**

The strain gauge is the most common sensing element used in many force sensing applications. To provide precise force measurement, a flexible structure is usually bonded to the strain gauge. Metal foils are usually applied to the surface of a device, deformation, caused by applied force through the gauge’s displacement or the change of the electrical resistance of strain gauge [1, 57]. This technology is widely used in robotics and medical applications [87-91]. Force sensors using strain gauges can be made very small. The minimum size of an individual strain gauge is 3 mm by 6 mm [92]. These sensors can be stable and robust, but they are sensitive to magnetic resonance environments as well as temperature variations [57].

However, as force sensing is based on the measurement of structural deformation, there is always a balance between the stiffness of the material and the sensitivity of the measurement [93]. If a stiff structure is used, it tends to have a low measuring sensitivity, which causes difficulties in force measurement. In addition, hysteresis is also a factor which restricts the sensitivity of the force measurement of strain gauges [1].
2.3.2 Optical force sensors

Optical force sensors were introduced in the 1960s, mainly driven by the needs of medical progress [76]. For example, some surgical instruments and surgeries, such as endoscopes, intravascular and cardiac surgery, need integrated, miniature force sensors that can work within a strong magnetic resonance environment, which was a big challenge for other types of sensors. Four main categories of optical force sensors are employed according to their varying modulation principles: intensity, phase, wavelength and polarization modulation [94]. Over the past three decades, due to increasing necessity from the medical world and extensive researches in optic sensor technologies, optical force sensors and their applications are expanding very rapidly.

- **Intensity-based force sensors**

  In the early 1960s, intensity modulated optical sensors were introduced [95, 96]. They resolved the force sensing problem in surgical applications. Intensity modulated force sensor suggested since the 1960s are reviewed by Paulo [97].

  Generally, intensity-based force sensors have a simple structure and only require modest signal processing through a detection of changes in optical intensity either in transmission or reflection [94]. The multimode optical micro-bend force sensor, proposed and demonstrated in the 1980s [98, 99], is a successfully commercialized intensity-based sensor. The resolution of this force sensor could reach 0.1% or higher. However, many drawbacks are still existing in this technology, such as accuracy limitations due to hysteresis, optical source power fluctuation and fibre loss [100]. Moreover, intensity-based micro-bend force sensors need to be large due to the mechanical micro-bending mechanism [94].

- **Wavelength-modulated force sensors**

  A fibre bragg grating (FBG) force sensor is a typical wavelength-modulated force sensor and is widely used in many areas such as automotive industry, civil engineering, undersea oil exploration, biomechanics and rehabilitation sensing applications [94, 101, 102]. Since the first FBG sensor was demonstrated by Hill in 1978 [103], FBG sensors have been widely used in measuring temperature, strain and pressure [104]. FBG force
sensors have been used for measuring hydrostatic pressure, with a resolution of 0.5% [105, 106], and force feedback in ophthalmic surgery and retinal microsurgery with a sensitivity of 9 $mN$ and 0.25 $mN$ respectively [107, 108]. However, technical difficulties, such as temperature cross-sensitivity, long-term reliability issues, errors introduced through source power fluctuations and changes caused by fibre loss, still exist and block the way for FBG sensors becoming practical applications [94].

- **Polarization-modulated force sensors**

The Faraday and the photoelastic effect are the two main physical effects applied in polarization-modulated force sensors. While the Faraday effect is based on sensors that measure the magnetic field, the photoelastic effect is more suitable for development into force sensors as it transfers the force directly into a polarization property change [94]. The first photoelastic effect based optical force sensor was introduced by Spillman in 1982 [109]. Optical force sensors based on photoelastic effect could achieve 0.2% in accuracy if they have a good self-compensation mechanism, which is the limitation for practical applications in harsh environments as self-compensation has to be built for the individual sensing location [94].

- **Interferometry-based force sensors**

Since the 1980s, interferometric based sensors have been exploited and introduced into force sensing techniques to overcome the above flaws [76]. These are the Mach-Zehnder, Michelson and Fabry-Perot interferometers. The first two interferometer-based sensors are mainly used for detecting acoustic pressure in the early stage of optical force sensing development [110-114]. But these require a long sensing fibre to gain the desired sensitivity. This introduces a temperature instability problem. Moreover, these two types of sensors also suffer from a polarization fade problem. The above problems render them unsuitable for sensing applications requiring long-term stability. The Fabry-Perot interference (FPI) based sensor is a novel and promising technology in force sensing applications.

Although optical fibres have promising potential for force measurement, there are limitations in utilizing relative materials. For example, the accuracy of intensity based
force sensors will be affected by alterations to the light signal caused by bending and misalignment. In this situation, prevention or compensation methods such as a reference fibre have to be considered [115]. Additionally, optical fibres are more fragile compared with other sensing materials and need precise interfaces with other components. Small fibre bending and temperature change could cause signal noise and signal fluctuation, while large deformations may damage the fibre sensors.

From the 0, it can be seen that all types of force sensors in the literature could be designed on a very small scale having with high resolution. However, the various sensors possess diverse advantages and drawbacks. The best choice of a suitable force sensor should be made according to the specific working environment.

<table>
<thead>
<tr>
<th>Table 2.2</th>
<th>The pros and cons of various sensing techniques [21, 57, 82, 84]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sensing technology</td>
<td>Modulation methods</td>
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<tr>
<td></td>
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<tr>
<td>Piezoresistive</td>
<td>Resistance change</td>
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<td>Capacitive</td>
<td>Capacity change</td>
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<tr>
<td>Piezoelectric</td>
<td>Strain polarization</td>
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<tr>
<td>Strain gauges</td>
<td>Resistance change</td>
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<td></td>
<td></td>
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<tr>
<td>Principle</td>
<td>Sensitivity/Resolution</td>
</tr>
<tr>
<td>----------------------------</td>
<td>------------------------</td>
</tr>
<tr>
<td>Piezoresistive [116]</td>
<td>10.361±0.267 uN/V</td>
</tr>
<tr>
<td>Piezoresistive [117]</td>
<td>0.13 mV/mA-MPa</td>
</tr>
<tr>
<td>Piezoresistive [118]</td>
<td>-</td>
</tr>
<tr>
<td>Piezoresistive [119]</td>
<td>100 nN</td>
</tr>
<tr>
<td>Capacitive [71]</td>
<td>-</td>
</tr>
<tr>
<td>Capacitive [120]</td>
<td>-</td>
</tr>
<tr>
<td>Capacitive [121]</td>
<td>1.35 mV/uN</td>
</tr>
<tr>
<td>Capacitive [122]</td>
<td>3 uN</td>
</tr>
<tr>
<td>Piezoelectric [123]</td>
<td>7 pc/uN</td>
</tr>
<tr>
<td>Piezoelectric [124]</td>
<td>25 mN/40mN</td>
</tr>
<tr>
<td>Piezoelectric [125]</td>
<td>230 Pa</td>
</tr>
<tr>
<td>Strain gauges [126]</td>
<td>0.3 N</td>
</tr>
<tr>
<td>Strain gauges [127]</td>
<td>0.5 mN</td>
</tr>
<tr>
<td>Strain gauges [128]</td>
<td>0.05 N</td>
</tr>
<tr>
<td>Strain gauges [129]</td>
<td>1 nN</td>
</tr>
<tr>
<td>Optical [130]</td>
<td>0.1569 μm/kPa</td>
</tr>
<tr>
<td>Optical [131]</td>
<td>0.533 kPa</td>
</tr>
<tr>
<td>Optical [132]</td>
<td>0.133 kPa</td>
</tr>
<tr>
<td>Optical [133]</td>
<td>0.01 N</td>
</tr>
</tbody>
</table>

Table 2.3 Property comparison of sensing technology designs
**2.4 Fibre optical force sensor based on FPI**

From 1899 to 1901, Fabry and Perot formulated the FPI theory and published their findings [134, 135]. FPI-based applications have a very simple structure, with two partially reflecting mirrors that form a cavity. The reflected light from the two mirrors produces multiple interferences the phase of which can be changed by changing the FPI cavity’s length [136].

Based on the ways in which the cavity is created (see Figure 2.1), FPI sensors have two types of structure: intrinsic FPI sensors and extrinsic FPI sensors. In intrinsic FPI sensors, the cavities are formed within the fibre by fusion spliced fibres of the same diameter and cleaved or coated end faces. Mirror coatings on the fibre end faces have better reflectivity than fibres with cleaved end faces [137]. In extrinsic FPI sensors, the cavities are made outside of the fibres, and the fibres are the medium for transmitting light into the cavities and receiving the reflected light from the second mirror [136, 138], which could be a cleaved or coated fibre end or a metal mirror. The cavities can be bonded to the input optical fibre through splicing techniques or epoxy resin. The cavities can also be formed by a mechanical structure (see the design in [139]). Typical intrinsic and extrinsic FPI sensor configurations are shown in Figure 2.12.

![Figure 2.12 Typical intrinsic and extrinsic FPI sensor configurations [137]: (a) Intrinsic FPI structure (b) Extrinsic FPI structure](image)

25
2.4.1 Reasons for employing the FPI principle

The requirements of surgical force sensor, make optical-based force sensors ideal for MIS, compared to traditional electronic force sensors. Specifically, optical sensors use non-electrical signals to detect the contact force, are immune to MR influence, are less influenced by temperature change and do not impact on MRI devices [140, 141]. Traditional electronic force sensors are greatly affected by the MR environment and temperature change [141, 142]. Optical sensors can also be very small, mechanically stable, sterilizable, and do not pose a spark source hazard for flammable environment applications [94, 139]. Therefore, optical-based force sensors are well suited to be incorporated into conventional surgical instruments.

Compared to other optical force sensing methods, the FPI force sensing technique has advantages. For example, FBG sensors, one of the best solutions for force sensing in an MR environment, also have high sensitivity and can be miniaturized, and have been a favourite option on many robotic devices and health monitoring systems [143]. However, they require controlled working environments, and require high manufacturing costs and spectral analysis instruments [140]. In addition, temperature cross-sensitivity and the necessity for temperature compensation is another drawback when FBG sensors are applied within surgical devices [141]. Based on an intensity and phase measuring method, FPI sensors have a simple structure and are easy to design and manufacture [144], can relying on a simple interference pattern based on the voltage measurement, and they can be used temperatures up to hundred degrees [145].

2.4.2 Fabrication methods

2.4.2.1 Air-gap cavity manufacturing methods

Cleaving, fusion splicing, micromachining, selective etching and photolithography patterning, Complementary Metal-Oxide-Semiconductor (CMOS) transistor and Micro-Electro-Mechanical Systems (MEMS) are the main methods cited in studies for fabricating FPI cavities [145].
• **Cleaving and fusion splicing**

Cleaving and fusion splicing techniques are the most common methods used in fabricating FPI sensors. Specifically, cleaving is applied to produce reflective mirrors and form suitable diaphragm thicknesses or cavity lengths. To assemble the FPI sensor, fusion splicing is used to integrate cavities and reflecting mirrors into the fibre ends.

By applying the fusion splicing method, Rajan, et al. used HC-1550 fibre and LMA-10 PCF fibre to fabricate an FPI force sensor based on a hollow core photonic-crystal fibre (PCF) and a solid core PCF [141]. A small section of the PCF was spliced to a single mode fibre pigtail through a fusion splicing process. A standard splicing machine was used to collapse the central region of the PCF to form a micron sized tapered region of 0.3 \( \text{mm} \) length and a 19 \( \mu \text{m} \) thickness at the waist. By controlling the diameter and the length of the tapered waist, the interferometer can be adjusted.

• **Femtosecond laser patterning**

Femtosecond laser patterning was also introduced to produce FPI pressure sensors. Zhang, et al. [94] used this technology to produce an external diaphragm for FPI pressure sensors. The authors pointed out that this technique could form a very short cavity, which is important in FPI sensing, as the longer the cavity length, the greater the temperature interference. Similarly, Liu, et al. used a laser-cut technique to fabricate a flexure made of Nitinol to bond the mirror and the tool shaft to form an FPI force sensor used for vitreoretinal microsurgery [139].

• **MEMS technology**

MEMS technology was applied by Ahmadi, et al. [146, 147] to fabricate a flexible silicon membrane for building an FPI optical force sensor used in MIS cardiac annuloplasty surgery. They modelled a sensor using a single-mode fibre which was integrated into the catheter via MEMS techniques.

• **Semiconductor processes and micromachining**

Semiconductor processes and micromachining techniques were used by Wolthuis et al. to fabricate an FPI sensor, with a size of 300×300×275 \( \mu \text{m}^3 \), used for blood pressure
testing [148]. Totsu, et al. [131] also used micromachining techniques to fabricate their FPI sensors. They used an aluminium mirror as the diaphragm and a polyimide spacer to form a cavity in the sensor.

- **Etching technology**

Etching technology was introduced by Cibula et al. to develop an FPI pressure sensor [132]. The inside cavity of the sensor was created by wet etching with dilute acid, the diaphragm being part of the fibre instead of being attached to the fibre end. Anisotropic etching techniques were applied by Tohyama, et al. [149] to produce a diaphragm-cavity in an optical force sensor to measure inflation pressure.

- **Polishing machine**

A polishing machine was used by Bae, et al. [150] to form a 45° angled end face on a fibre as a part of FPI sensor, with a thin layer of silver to enhance light reflection. The sidewall of the polished fibre was coated with a positive photoresist (AZ4620, Shipley) by soft baking for five minutes at 95 °C.

#### 2.4.2.2 Mirror coating and light reflecting materials

Table 2.4 shows the FPI cavity fabrication information from current researches. Most of the micro sized FPI cavities use diaphragms as secondary reflecting mirrors. The materials use are central for sensor development as the sensor’s performance will be affected by the material’s properties such as thermal expansion, thermal conductivity or mechanical strength. Metals, ceramics, and glass are the most common materials used in sensor design [151].

As to metals, stainless steel, aluminium, copper, and Kovar are common choices used in previous research. These are much easier to fabricate, compared with ceramics and glasses, but ceramics are more suitable for the harsh environments that have high temperature and corrosive chemicals. However, they are expensive and more difficult to manufacture. While metals and ceramics have different thermal properties with optical fibres, the expansion induced by temperature changes will lead to greater stress when they are combined with optical fibres. From this point of view, glass is an ideal material for sensor design especially when the FPI cavity is bonded to fibres. Besides, these materials
are other reflecting materials that have been applied by researchers, such as SU-8 and Poly methyl methacrylate (PMMA) [152]. Coating materials and their thicknesses from previous research are listed in Table 2.4.

<table>
<thead>
<tr>
<th>Reflecting materials</th>
<th>Cavity length(μm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>TiO₂ [148]</td>
<td>1.4-1.7</td>
</tr>
<tr>
<td>Gold [153]</td>
<td>50-100</td>
</tr>
<tr>
<td>Glass [123]</td>
<td>-</td>
</tr>
<tr>
<td>Metal [154]</td>
<td>15.23</td>
</tr>
<tr>
<td>Silicon [155]</td>
<td>20</td>
</tr>
<tr>
<td>TiO₂ [155]</td>
<td>2.5</td>
</tr>
<tr>
<td>Silicon [146]</td>
<td>-</td>
</tr>
<tr>
<td>Silver [150]</td>
<td>13.8</td>
</tr>
<tr>
<td>Glass [156]</td>
<td>75</td>
</tr>
<tr>
<td>Polymide [152]</td>
<td>100</td>
</tr>
<tr>
<td>Silver [157]</td>
<td>110</td>
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<tr>
<td>Stainless steel [139]</td>
<td>-</td>
</tr>
<tr>
<td>Glass [140]</td>
<td>15.8</td>
</tr>
<tr>
<td>Ai [158]</td>
<td>1.4</td>
</tr>
<tr>
<td>Glass [159]</td>
<td>100</td>
</tr>
<tr>
<td>PD film[160]</td>
<td>20</td>
</tr>
<tr>
<td>Si [161]</td>
<td>60</td>
</tr>
<tr>
<td>Glass [162]</td>
<td>20-30</td>
</tr>
<tr>
<td>Glass [145]</td>
<td>8 &amp;25</td>
</tr>
<tr>
<td>Glass [163]</td>
<td>20-30</td>
</tr>
</tbody>
</table>

2.4.2.3 Cavity length design

Cavity length has a great bearing on FPI force sensors. It has an effect on visibility of FPI signal intensity, dynamic range and measurement precision [164]. The visibility of broad band signals diminishes, and their intensity becomes flat when the sensor’s air-gap is larger than 60 μm. The author suggested an air-gap range from 10 - 30 μm maintains good signal visibility. Some FPI cavity lengths suggested by researchers are listed in Table 2.4.

2.4.3 Signal processing methods

A spectrum phase trace algorithm and interferometric-intensity detection are two main methods used to detect the changes in laser interference. The first method has high resolution, but the response frequency is limited [104] and it requires very expensive
spectrum analysis instruments. The intensity-based detecting method has a large bandwidth, and is suitable for dynamic force detection [165].

2.4.3.1 Working principle of FPI-based force sensors

The FPI sensor has a very simple structure, with two partially reflective mirrors that form an air cavity. The reflected light beams from the two mirrors produce multiple interferences which can be changed in interference intensity by changes in the cavity length [136]. A typical layout of a sensor is shown in Figure 2.13. It usually consists of one single mode fibre with a coated cleaved end, a tube, a metal diaphragm coated with reflective material. They are assembled using different methods, such as super glue or thermal-fusion techniques. The light propagates through the cavity and is reflected by the coated fibre end and coated diaphragm, and travels back along the fibre and forms interference fringes. If force is applied, the deformation will be produced and the air cavity length will change. Thus, reflected interference light will be changed in terms of phase and intensity.

![Typical FPI sensor structure](image)

**Figure 2.13** Typical FPI sensor structure

2.4.3.2 Spectrum phase based detection method

A wideband light source or a wavelength tunable light source is used in this method to produce an interference signal. Then spectrum instruments, such as a power meter or spectrometer, are applied to recognise peaks in the interferometry and their changes [166]. The cavity length change induced spectrum shift is shown in Figure 2.14.
Through the spectrum shift $\Delta L$, the change in cavity length could be calculated. The interference pattern of the reflection spectrum is described as follows:

$$I_{FP}(\lambda) = R_1 + (1-A_1)^2 (1-R_1)^2 (1-\alpha)^2 R_2 + 2\sqrt{R_1 R_2} (1-A_1)(1-R_1)(1-\alpha) \cos(4\pi L_{air}/\lambda-\pi)$$

(2.6)

Where $A_1$ is the transmission loss factor on surface 1 due to surface imperfections, $\alpha$ is the loss factor of the cavity induced by diffraction in the air cavity, $L$ is the FPI cavity length, $R_1$ and $R_2$ are the power reflection coefficients of the two reflective mirrors, and $\lambda$ is the light’s wavelength.

**2.4.3.3 Interferometric-intensity detection method**

Using this method, the force sensor needs a fixed wavelength light source with a narrow bandwidth to create interference fringes. A photodiode is then used to convert the light intensity signals into electrical signals, such as voltage or current. To avoid fringe direction change ambiguity, the sensor based on this method is designed to limit fringe change within a linear range of one fringe, as is shown in Figure 2.15.
In an intensity based force sensor, initial air-gap control is very significant. It is necessary to set the initial air-gap length for the intensity change to start from the beginning or middle of the linear range of the fringe.

\[
\phi = \frac{2\pi L}{\lambda} + \pi = \pi\left(\frac{2L}{\lambda} + 1\right)
\]  

(2.7)

\[
\phi_{opt} = \frac{1}{2}(2k + 1)\pi, \quad k \in N
\]  

(2.8)

Set the initial change from the start: \( L_{ini} = \frac{k}{4}\lambda_0, \quad k \in N \) and \( \lambda_0 \) is in microns.

Set the initial change from the middle: \( L_{ini} = \frac{2k-1}{8}\lambda_0, \quad k \in N \) and \( \lambda_0 \) is in microns.

The relationship between signal intensity and cavity length could be calculated by the following equation.

\[
I = I_1 + I_2 + 2\sqrt{I_1 \cdot I_2} \cos\left(\frac{2\pi \varepsilon_x d}{\lambda}\right)
\]  

(2.9)

Where \( I_i \) is the intensity of reflected light, \( \varepsilon_x \) is the strain in the axial direction, \( d \) is the distance between the two mirrors at zero load, and \( \lambda \) is the wavelength of the light source.
2.4.4 Temperature influence and compensation

Various optical fibre force sensing techniques have been investigated, among which the extrinsic FPI sensor is thought most suitable for in-vivo, embedded applications [167]. However, an FPI sensor is sensitive to temperature and becomes of little use in the MIS environment, where temperatures can change up to 38 °C. To design a practical FPI sensor, several temperature compensation techniques have been developed, but these are not feasible for embedded applications [163, 168-171].

For intrinsic FPI force sensors, temperature influences can be ignored. Some researchers have investigated the temperature properties of their sensing designs, and the results are positive in this respect [131, 145, 155, 172, 173], with most designers of intrinsic FPI force sensors not mentioning any temperature cross-sensitivity problems. However, some researchers do suggest that using integrated FBG sensor to detect temperature to eliminate the small influence [160, 174].

A ambient temperature does have an impact on extrinsic FPI cavity length, induced by thermo-optic effect and thermal expansion [175], especially for unsealed extrinsic FPI sensors [176]. The longer the cavity length, the greater the temperature influence. A force sensor used in a surgical environment will be subject to different temperatures from room temperature (usually 20 °C) to a patient’s body temperature (about 37 °C), sometimes even higher (more than 200 °C) due to some heating medical instruments [141]. Thus, temperature influence cannot be ignored.

Different solutions have been proposed to eliminate thermal influence on optical fibre sensors. Three main types of temperature compensation approaches have been found in
previous research. The first means is setting a reference temperature sensor. For example, an FBG sensor was employed as a temperature compensator [174] as it was not exposed to the mechanical forces of the design.

Another way is mechanical self-compensated structures that consist of various materials with different thermal coefficients, which is the most common choice for temperature compensation. Through precise calculation, the FPI mirror could stay at the same distance when the temperature changes as the design could enable part of the structure to have an equivalent thermal effect in the opposite direction. For example, different materials with different thermal expansion coefficients were used to form an air cavity which would reduce cavity length changes induced by temperature. The sensor could then compensate itself as the temperature changed. The experimental results show that cavity length changes induced by temperature were reduced from 0.142 to -0.045 μm/°C. Two self-compensated FPI mechanical structure designs are shown in Figure 2.17 and Figure 2.18.

![Figure 2.17 Temperature compensation using self-compensated mechanical structure [177]](image)

![Figure 2.18 Temperature compensation using self-compensated mechanical structure [178]](image)

The last type of temperature compensation is realized through miniaturizing the FPI cavity. Many FPI force sensors minimise temperature cross-sensitivity by designing as small an air cavity length as they could, mostly at the level of micro meters, to avoid the
accumulating thermal effect. Three examples are illustrated in Figure 2.19 are the typical designs presented in previous research. The tiny cavity was mainly fabricated by laser micromachining technology. This sort of the FPI sensor is still influenced by temperature but at a minimal level which can be regarded as signal noise.

![Figure 2.19 Temperature compensation by fabricating a miniature FPI cavity](image)

(a) (b) (c)

2.4.5 State-of-the-art FPI applications in MIS

Recently, new manufacturing methods have been applied and some special FPI applications have been designed as MRI-compatible force sensors for use in MIS.

In 2004, Peirs, et al. fabricated a micro FPI force sensor integrated into a needle driver [23]. It is designed to sense tri-axial force on the driver’s end, using three optical fibres and a flexible structure having a diameter of 5 mm, as shown in Figure 2.20. The optical fibres are glued into the core and the secondary FPI cavity mirrors are also glued at the front with a 100 μm air-gap between the polished mirrors and fibres’ end. It has a axial force range of 2.5 N and a radial directional range of 1.7 N range in the radial direction. And its resolution is 0.01 N. The sensitivity of the sensor differs from the values in the design due to manufacturing errors. The authors also suggest that the repeatability of each sensors’ properties cannot be guaranteed.
In 2005, Totsu, Haga, and Esashi exploited an FPI application for detecting force in MIS, as shown in Figure 2.21 [131].

The diameter of the pressure sensor is only 125 μm. It has the sensitivity of -0.25 mmHg\(^{-1}\) and a resolution of 4 mmHg for pressures ranging from -100 to 400 mmHg. The sensor was integrated into an injection needle with a diameter of 0.6 mm. Utilizing a goat as an experimental subject and a spectrometer, they observed the results of the sensor. Although dynamic blood pressure was successfully observed, influence of temperature changes, leading to expansion of the air in the FPI cavity, also emerged. The designers used a vacuum FPI cavity to solve this problem. Another problem for the design is the low visibility of the reflection spectrum when using multimode optical fibre, even though the air-gap length was already very small (about 2 μm). They believe utilizing a single-mode optical fibre will improve the sensitivity of the sensor.
In 2010, a design for measuring the pressure of fluids was proposed [150], as shown in Figure 2.22. They turned the optical axis by 90° through a 45° angled fibre end face. An FP cavity is formed on the sidewall of the fibre. The sensor exhibits a pressure range of 13 - 98 kPa, with a sensitivity of 0.0013 μm/kPa. In their experiment, a broadband light source and a spectrometer were used as signal resource and processing respectively. A small drift induced by temperature change, about 0.04% of the initial cavity length, was observed during the experiment. To resolve temperature influence, they suggested a temperature sensor or an additional FP cavity with a different initial cavity length could be used to provide temperature information to compensate for the effect of temperature influence.

In 2012, a miniature FPI force sensor embedded in a stainless steel surgical tool tip was developed [139], used for vitreoretinal microsurgery, as shown in Figure 2.23. The two mirrors, the cleaved fibre end and the tool shaft end, are bounded by a Nitinol flexure. The super-elastic property of Nitinol was used to form a displacement when force is applied. It has a 0.5 mm outside diameter, its range is from 41×10^3 to 75×10^3 MPa and it has the sensitivity of 0.25 mN. They noticed that temperature variations, mechanical vibration, and air/fluid flow would have an influence on the accuracy of the force measurement. To reduce the noise created by the environment, the system was covered with an enclosure and the data was collected in a relatively short time. They also believed that laser relative intensity noise (RIN) and quantum noise exists in the signal. Another problem was sealing the tool to protect FPI sensor from liquid in the organs.
In 2013, Shang, et al. designed an MRI-compatible miniature uniaxial force sensor based on FPI for sensing the axial force of a needle tip during needle placement [140]. It has more than two mirrors, but only two are the main reflecting mirrors. The detectable strain ranges are from ±1000 \( uN \) to ±5000 \( uN \), with a resolution of 0.01% of full range. However, the sensor is not embedded inside the needle, but is located in a piezoelectrically actuated robot.

In 2013, Luo, et al. [180] and Ngajikin, et al. [158] designed their FPI force sensors using MEMS technologies. Luo, et al. used a standard TSMC 0.35 \( \mu m \) 2P4M CMOS process to produce dual tunable resonant cavities. Cavities with an air-gap of 0.64 \( \mu m \) and an oxide membrane of 2.64 and 0.9 \( \mu m \) thickness were created, and a light source with a wavelength ranging from 450 to 800 \( nm \) was applied in their experiments. Ngajikin, et al. also fabricated an FPI force sensor through MEMS technologies. It was designed for blood pressure measurement.
In their design, a diaphragm $1 \mu m$ in thickness, with an FPI cavity $1400 \text{ nm}$ in length was attained. By using a $550 \text{ nm}$ optical light source, they gained an analytical result with high linear response ranging from 0 to $40 \text{ kPa}$ and sensitivity of $1.83 \text{ nm/kPa}$ (spectrum shift/pressure). However, they only performed theoretical analysis of the force sensing mechanism and simulation of its fabrication in their paper.
The various properties of these designs are summarised in Table 2.5. The designs in published papers suggested that the FPI sensors could be designed to be extremely small and have very high resolution and sensitivity, making them suitable for integrating into surgical tools’ tips. However, based on a search of published papers on Google Scholar, there are still no commercial integrated FPI force sensor applications now available.

Table 2.5 Properties of state-of-the-art FPI sensor designs

<table>
<thead>
<tr>
<th>Authors</th>
<th>Force ranges</th>
<th>Resolutions or sensitivities</th>
<th>Cavity length (μm)</th>
<th>Reflecting materials and thickness</th>
<th>Other information</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peirs, et al.</td>
<td>Axial: 2.5N</td>
<td>Resolution: 0.01 N</td>
<td>100</td>
<td>-</td>
<td>Sensor diameter: 5mm</td>
</tr>
<tr>
<td></td>
<td>Radial: 1.7N</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Totsu, Haga, and Esashi</td>
<td>-0.98-3.92 N/cm²</td>
<td>Sensitivity: 0.25 nm/mmHg⁻¹</td>
<td>2</td>
<td>Aluminium, 100 nm</td>
<td>Sensor diameter: 125 μm</td>
</tr>
<tr>
<td></td>
<td></td>
<td>resolution: 0.044 N/cm²</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Authors</td>
<td>Force (N/cm²)</td>
<td>Sensitivity</td>
<td>Sensor Material</td>
<td>Sensor Diameter</td>
<td>Notes</td>
</tr>
<tr>
<td>--------------</td>
<td>---------------</td>
<td>-------------</td>
<td>----------------</td>
<td>----------------</td>
<td>--------------------------------------------</td>
</tr>
<tr>
<td>Bae, et al.</td>
<td>1.31-9.8</td>
<td>0.009 μm/psi</td>
<td>13.8</td>
<td>150 μm</td>
<td>Silver, 150 nm</td>
</tr>
<tr>
<td>Liu, et al.</td>
<td>4.1-7.5</td>
<td>0.25 mN</td>
<td>-</td>
<td>0.8 mm</td>
<td>Stainless steel, Commercial sensor from FISO Multimode fibre sensor, OD=0.23 mm</td>
</tr>
<tr>
<td>Shang, et al.</td>
<td>±1000-±5000 μN</td>
<td>0.01% of full range</td>
<td>2.3</td>
<td>Semi-reflected materials</td>
<td>SiO₂, 2.64 μm and 0.9 μm</td>
</tr>
<tr>
<td>Luo, et al.</td>
<td>-</td>
<td>-</td>
<td>0.64</td>
<td>SiO₂, 2.64 μm and 0.9 μm</td>
<td>No</td>
</tr>
<tr>
<td>Ngajikin, et al.</td>
<td>0-4</td>
<td>1.83 nm/kPa</td>
<td>1.4</td>
<td>Al, 1 μm</td>
<td>OD=0.18 mm</td>
</tr>
</tbody>
</table>

2.5 Summary and conclusion

In MIS, surgeons lose most of the traditional instrument force sensing of open surgery due to the influence of friction forces, extremely small forces and indirect contact induced by the utilization of robotic instruments. Without adequate force control, surgeons are severely constrained in reducing important organs’ injuries. Therefore, giving force feedback to surgeons through integrating force sensors into surgical instruments is important. In this chapter, current force sensing mechanisms have been reviewed, including their capabilities, advantages and limitations. Among those sensing principles, optical force sensing methods are the most suitable for application to surgical environments with MRI devices as optical signals are unhindered by MR interference.

Recently, force sensors based on FBG and FPI are regarded as the most promising designs suitable for MIS. Many research groups have intensively studied related issues, and much research work is still in progress. Some excellent FPI force sensor designs have been reviewed and detailed. However, there are still many constrains that need to be resolved in FPI sensor design. For example, the intensity based FPI force sensor is restricted in detection within 1/8 wavelength change, and the influence temperature also needs to be compensated. In addition, before these force sensors can be considered practical, some obstacles, such as hardware design, signal processing, surgical systems integration, and the human-device interface still need to be overcome.
Chapter 3

Temperature-compensated optical fibre force sensing at the tip of a surgical needle

3.1 Introduction

An FPI sensor based on interferometric intensity-phase modulation is proposed for in vivo force sensing at the tip of a surgical tool for MIS. In this chapter, in order to develop a tip force sensing sensor for tissue identification, the design requirements for needle application were determined according to routine needle surgery. The main obstacle to the application, temperature influence, was confirmed via numerical calculation, simulations, and temperature experiment. To resolve this problem, a temperature compensation design for needle application was then proposed. The sensor fabrication procedure was described, as well as the fabrication of a temperature-compensated tip force sensing needle. To calibrate the sensing needle, a needle insertion platform was
designed. The signal processing system is also given, together with the algorithm for temperature compensation.

After the tip force sensing needle was produced, a preliminary experiment for the concept of tip force sensing was carried out. A tip force sensing needle without temperature compensation was fabricated and calibrated. Insertion experiments on phantom tissue were conducted, and the results show the potential of the design. After this, a temperature compensated tip force sensing needle was fabricated and calibrated. The performance of the sensing needle was then assessed through phantom tissue insertions.

3.2 Sensor Design and Fabrication

3.2.1 Design considerations

Aside from the influence of magnetic resonance, sensor specifications, such as sensing range, resolution, and sensor size, are central and need to be in line with human tissue properties and the instrumental requirements of MIS.

In vivo human skin insertion forces were studied [71, 181], which were undertaken on arm skin and abdomen skin, using puncture needles with outer diameters from 0.3 mm to 0.6 mm and tip angles of 45° and 90°. It was found that the needle needs 0.2 to 6 N to break the skin layer, and the force increases as the diameter of the needle and insertion speed increase. Insertion experiments on animal internal organs, such as liver, kidney and heart (from bovine, porcine and leporine) were carried out, and the breaking force was found to range from 0.2 to 2.3 N [69, 70, 73].

In this study, therefore, the force sensing range was specified to be 0 - 8 N. As a surgeon can perceive a minimum force of around 0.5 N [182], the sensing resolution was set lower than 0.5 N. Taking into account a MIS incision size that is about 3 - 12 mm [3, 183], the sensor was assumed to have a diameter less than 2 mm. The sensor specifications are summarized in Table 3.1.
Table 3.1  FPI Force Sensing Requirements

<table>
<thead>
<tr>
<th>Sensor specifications</th>
<th>Expected values</th>
</tr>
</thead>
<tbody>
<tr>
<td>Force type</td>
<td>Tip axial force</td>
</tr>
<tr>
<td>Sensing range</td>
<td>0 - 8 N</td>
</tr>
<tr>
<td>Sensor resolution</td>
<td>&lt; 0.5 N</td>
</tr>
<tr>
<td>Sensor size</td>
<td>&lt; 2 mm</td>
</tr>
</tbody>
</table>

3.2.2  FPI force sensing principle

A typical FPI sensor (Figure 3.1) has two cleaved single mode fibres embedded inside a glass capillary by epoxy. There is a micrometre level cavity between the two fibre ends. An incident light is delivered into the fibres and four percent of the incident light is reflected by each cleaved fibre end. Consequently, the two beams of reflected light interfere with each other, resulting in an interfered light that can be detected by a photodiode sensor or a spectrometer.

![Figure 3.1 A typical FPI sensor](image)

When an axial loading is applied, the cavity length varies, so does the intensity of the interfered light, which is described by [140],

\[ I = I_1 + I_2 + 2 \sqrt{I_1 I_2} \cos(\phi) \]  \hspace{1cm} (3.1)

where \( I \) is the intensity of the interfered light, \( I_1 \) and \( I_2 \) are the light intensities of the two reflected incident light beams, and \( \phi \) is the intensity phase, which is the phase difference between the two reflected lights.

The intensity phase can be calculated by,

\[ \phi = \frac{2 \pi (\Delta L)}{\lambda} \]  \hspace{1cm} (3.2)
\[ \Delta L = 2 \left( d + \Delta d \right) \tag{3.3} \]

where \( \Delta L \) is the optical path-length difference between two reflected light beams, \( d \) and \( \Delta d \) are the initial cavity length and the change in cavity length, respectively and \( \lambda \) is the incident light wavelength.

The interference intensity of the sensor is a sinusoidal function of the intensity phase. Consequently, the measurement of the intensity \( (I) \) gives rise to the change in cavity length \( (\Delta d) \) and further the axial loading force \( (F) \), given a sensor and the incident light.

A simulation based on Equation (3.1) was performed, and the result is shown in Figure 3.2. One light intensity value corresponds to more than one cavity length if the cavity deformation is larger than 0.1 \( \mu m \). Therefore, a light intensity variation tendency should be considered when experiencing peak values of light intensity. Previous studies \cite{165} proposed avoiding peak value by setting the sensing range to a linear range between two peak amplitudes.

![Figure 3.2](image)

Figure 3.2 Light intensity change simulation due to cavity length change

This phenomenon was also confirmed by the numerical simulation. The sensor is embedded into a medical puncture needle that has an inner diameter of 1.0 \( mm \) and a length of 80 \( mm \). According to Table 2.1, the force required for the needle to penetrate the skin and tissues ranges from 0 to 3 \( N \). Thus, an axial force of 3 \( N \) is applied to the needle tip simulated by finite element analysis. Two different conditions were simulated, and the difference is that in the first simulation, the needle can only be deformed in the axial direction while the needle has three degrees of freedom in the second simulation. Figure 3.3 shows the axial deformations of the glass capillary inside the needle are around
0.24 μm and 0.54 μm respectively. It reveals that deformation is generated by both axial force and needle curve. The simulation result also indicates that the sensor inside the needle tends to experience the peak value under 3 N force loading according to Figure 3.2.

Figure 3.3  Finite element analysis results: (a) axial freedom only analysis (b) three degrees of freedom analysis

3.2.3 FPI sensor fabrication

Figure 3.4  FPI sensor fabrication materials and instruments: (a) Single mode fibre (b) Three-hole fibre stripper (c) Epoxy (d) Puncture needle and glass capillary (e) Fibre cleaver
FPI sensor fabrication and embedding the sensor into the instrument are the two main tasks in the tip force sensing needle fabrication. Figure 3.4 shows the materials and instruments for sensor fabrication.

The cavity is formed by two ends of a cleaved fibre, which is cut by a fibre cleaving instrument and fixed inside a glass capillary. One fibre with a connector was cut and cleaved into two parts and fixed with epoxy into a glass capillary of 400 μm in outer diameter to form a 12 μm FPI air cavity. To produce a FPI qualifying sensor, the procedure is separated into six steps which are illustrated in Figure 3.5. The capillaries are cut into about 5 mm length to form FPI cavities (Step 1). One part of the coating-removed single mode (SM) fibre is inserted into the cut-capillary (Step 2), and then the end face is cleaved at an angle of 90 degrees, using a precision cleaver (F1lynx, Fis Lynx, US) (Step 3). After this, the fibre is pulled back into the middle of the capillary and fixed by epoxy (Thorlabs, US). Another SM fibre is cleaved (Step 4) and inserted into the capillary without touching the capillary end face (Step 5). The two cleaved fibre ends form an FPI cavity, where the cavity length can be adjusted to a desired value with the help of a microscope (Step 6). The cavity length is essential to FPI force sensors. It has an influence on intensity visibility, dynamic range and measurement precision [164]. The visibility of broad band signals diminishes, and their intensity becomes too weak to detect when the sensor’s cavity length is larger than 60 μm. It was suggested that the cavity
length range from 10 - 30 $\mu m$ to enable the intensity calibration and keep the signal observable.

After sensor fabrication, the sensor can be embedded into the tip of a puncture needle. Attention should be paid in this step due to the fragile nature of fibre and friction force inside the needle (Figure 3.6).

3.2.4 Temperature influence and proposed compensation solution

- Temperature influence validation via simulation

To evaluate the influence of temperature, a simulation was performed with a needle with an FPI sensor embedded at its tip, as shown in Figure 3.6. The needle was inserted from a room temperature of 23 °C into an empty chamber of 37.5 °C at a speed of 10 mm/s for a duration of 5 s.

![Simulated tip force sensing needle structure](image)

Figure 3.6 Simulated tip force sensing needle structure

Figure 3.7 is results of the simulation of needle temperature change and the FPI cavity deformation induced after 5 s of environment temperature change, shown in longitudinal section mode. It was found that the temperature change results in a cavity length change of 1.3 $\mu m$ over an initial cavity length of 12 $\mu m$, when incident light $\lambda = 1550$ nm and $I_1 = I_2 = 10$ $\mu w$. As the only variable of the environmental settings is environment temperature, the simulated result verified the influence of temperature on this application.
Figure 3.7  Simulated temperature change (a) and the FPI cavity deformation induced (b)

- **Temperature influence validation via experiments**

This phenomenon was also observed experimentally. Two tip force sensing needles with the same structure were inserted from room temperature into a chamber with a temperature of 37 °C, as shown in Figure 3.8 (a). The FPI light intensity variations are displayed in Figure 3.8 (b), showing that the temperature changing influence of temperature is significant.
According to the experimental results, in a similar situation, the cavity length change caused by a $6 \, N$ force loading was only around $0.25 \, \mu m$, which induced a light intensity change that much smaller than $30 \, \mu w$ compared with Figure 3.8 (b). This implies that the thermal influence is much greater and must be compensated for.

- **Temperature compensation solutions**

Two main types of temperature compensation designs were proposed, as shown in Figure 3.9. One is sensor miniaturization, for the purpose of reducing temperature induced deformation to a minor range. The other is adding another reference FPI sensor to provide temperature information.
Both types of solutions were investigated. Figure 3.10 is the tip force sensing needle based on solution Figure 3.9 (b). After examining the FPI signal, it was found that the temperature response was successfully reduced, however, the force-induced signal also reduced proportionally. In addition, it was found that temperature influence became significant again after fixing into the needle by epoxy, as shown in Figure 3.11. This is due to the larger coefficient of thermal expansion of the epoxy and steel, with $15-50 \times 10^{-6} /{\text{k}}$, compared with quartz glass, with $0.5 \times 10^{-6} /{\text{k}}$. Therefore, this temperature compensation solution does not settle the problem, especially for embedded applications.

![Figure 3.10 Exploration of one miniaturization solution](image)

![Figure 3.11 Temperature influence before and after epoxy fixation](image)

(a) Temperature increase
(b) Temperature decrease

Figure 3.12 shows the sensor fabrication for the exploration of introducing a reference FPI sensor. One of the FPI sensors is attached at the needle tip as a force sensor, and the reference sensor is only subject to temperature variations. The reference sensor has two
different placement options, hung as a cantilever or mounted on a metal support beam. To compare the outcome of the two structures, the needle was first fabricated as shown in Figure 3.12 (a), and then a piece of metal was attached for support as shown in Figure 3.12 (b). The signals of both situations were collected and compared.

![Figure 3.12 Exploration of reference sensor solution](image)

Figure 3.12 Exploration of reference sensor solution

Figure 3.13 shows the signals from the two designs: the reference sensor hanging in the air (a) and the reference sensor mounted on a support beam (b). Experimental insertions, from a room with a temperature of 23 °C into the chamber of 37.5 °C, were performed and the results are shown in Figure 3.13.

![Figure 3.13 Sensor placements for temperature compensation sensor (a) the reference sensor hung as a cantilever (b) the reference sensor mounted on a metal support beam](image)

Figure 3.13 Sensor placements for temperature compensation sensor (a) the reference sensor hung as a cantilever (b) the reference sensor mounted on a metal support beam

It was found that the placement (b) gave rise to a greater fluctuation in the intensity magnitude of the reference sensor. As a small change of the intensity signal is likely to be submerged by noise, therefore, structure design (b) was employed in this study.
The two FPI sensors can be placed into a needle in parallel or in series to each other. The simulations conducted with a temperature change of 23 °C to 37.5 °C show that the reference sensor being in series with the force sensor created a significant time delay in the cavity change. In this study, the two sensors were placed in parallel. To compensate for temperature influence, two identical FPI sensors were placed at the tip of a needle (14 gauge), as shown in Figure 3.14, one of which serves as a reference sensor and is subjected only to temperature variation, the other serving as a force sensor being subject to both temperature and axial loading at the tip of the needle. Each sensor has an outside diameter of 0.4 mm and an initial cavity length of 30 μm.

![Design sketch](image)

(a) Design sketch

![Image of the needle with the two sensors embedded](image)

(b) Image of the needle with the two sensors embedded

Figure 3.14 Placement of the two FPI sensors for temperature compensation

In this design, for the force sensor, the FPI cavity length change $\Delta d_f$ is induced by both force change $f_1(\Delta F)$ and temperature change $f_2(\Delta T)$, while the reference sensor is only influenced by temperature $f_3(\Delta T)$. Consequently, temperature influence on the force sensor can be compensated for.

$$\Delta d_{f} = f_1(\Delta F) + f_2(\Delta T)$$  \hspace{1cm} (3.4)

$$\Delta d_f = \Delta d_{f} - k\Delta d_R = f_1(\Delta F) + f_2(\Delta T) - k f_3(\Delta T)$$  \hspace{1cm} (3.5)

where $\Delta d_f$ is the effective cavity length change induced by force loadings, $\Delta d_R$ is the cavity
length change of the reference sensor due to temperature variation, and $k$ is the coefficient accounting for the difference in the temperature influence in the two sensors, which can be calibrated experimentally.

3.2.5 Signal processing system

Figure 3.15 Temperature-compensated FPI sensing system, (a) Schematic optical circuit, (b) Image of the physical optical circuit and its optimized optical circuit box
Figure 3.15 shows the schematic optical circuit of the temperature-compensated FPI sensing system. A 2 mw laser with a wavelength of 1550 nm is the source of incident light. An optical isolator prevents reflected light returning to the source. The laser is split into two channels by a 50/50 splitter, which is transmitted to the two FPI sensors. The interfered light of the two FPI sensors is then measured by two photodiode power sensors, and the measurements are further processed by a computer. After testing the circuit, it was assembled in a designed box, shown in Figure 3.15 (b).

3.3 Experimental Setup and Signal Processing

3.3.1 Experimental setup

Figure 3.16 shows an experimental platform constructed for characterisation of the temperature-compensated FPI force sensing system.

![Experimental Platform](image.png)

Figure 3.16 Needle insertion experimental platform

It consists of a framework, a linear actuator, a temperature controlled chamber, and a needle with the FPI sensors embedded. Phantom skin tissue can be put inside the chamber.
and is held by two acrylic plates. The linear actuator drives the needle into the chamber at a constant velocity to perform tissue insertion.

3.3.2 The algorithm for temperature compensation

The compensation algorithm takes into account two working scenarios: a constant temperature environment, and a changing temperature environment, under the conditions of FPI cavity length \( d \) of about 30 \( \mu m \), reflected light \( I_1 = I_2 = 5 \mu w \), and an input light wavelength \( \lambda = 1550 nm \).

3.3.2.1 Constant temperature condition

Based on Equations. (3.1) - (3.3), the relationship between FPI interference intensity change and the FPI cavity length change is shown in Figure 3.17.

![Figure 3.17 Interference intensity change versus the cavity length change](image)

While Equation. (3.1) may be used to estimate \((\Delta d)\) and the axial loading force \((F)\) by the measurement of \( I \), the sensing range has to be restricted to a quarter of the intensity between one valley and one peak [169, 184, 185]. Otherwise, in the vicinity of a peak, the intensity \((I)\) alone cannot be used to distinguish whether or not the cavity length increases or decreases, which leads to a signal ambiguity problem.

In this case, the sensing range can be set to a linear range as shown in Figure 3.17 to avoid the signal ambiguity. As mentioned earlier, a 6 \( N \) force causes a 0.25 \( \mu m \) decrease of the
cavity length and, consequently, the cavity length change range can be set to 29.9 - 30.2 \( \mu m \).

### 3.3.2.2 Changing temperature condition

Temperature variations can also be sensed by FPI sensors and drive the FPI intensity magnitude out of the linear range. To distinguish the force-induced signal from the temperature-induced signal, in this study, another FPI sensor working as a reference sensor was introduced as depicted in Figure 3.13 (b).

Figure 3.18 is the simulated FPI intensity signals of both sensors, with cavity lengths of 29.97 \( \mu m \) for the force sensor and 29.87 \( \mu m \) for the reference sensor, respectively, when the needle was moved from 23° to 37.5 °C, and an axial force was loaded at the needle tip at 1.7s and then released at 1.9s.

![Figure 3.18 Interference intensity changes due to temperature and force](image)

The reference sensor can be used to estimate the temperature induced intensity signal with the force sensor, as the cavity length difference between the two sensors results in an intensity phase shift, based on Equations. (3.1) - (3.3). Consequently, the difference between the estimated intensity signal and the measured intensity signal of the force sensor is the effective intensity signal due to the loaded forces, as shown in Figure 3.18.
To resolve the signal ambiguity problem at intensity magnitude peaks, the intensity phase analysis is used to estimate cavity length change, which is described as follows.

Once an intensity \( I_n \) is measured, its intensity phase \( \phi \) can be measured by,

\[
\phi = \cos^{-1}(f(I_n))
\]

(3.6)

in which when \( I_1 = I_2 = I \),

\[
f(I_n) = \frac{(I_n - I_1 - I_2)}{2\sqrt{I_1 I_2}} = \frac{I_n}{2I} - 1
\]

(3.7)

according to Equation. (3.1).

The intensity phase \( \phi \) of the force sensor in Figure 3.18 is calculated and shown in Figure 3.19. At the maximum intensity magnitude \( A \), whether the intensity phase increases or decreases can be confirmed by its change rate \( d\phi/dt \).

![Figure 3.19](image)

Figure 3.19  Intensity phase processing for temperature compensation and solving signal ambiguity

The effective phase \( \Delta \phi \) due to actual force can then be found as

\[
\Delta \phi = \Delta \phi_F - k \Delta \phi_R
\]

(3.8)

where \( \Delta \phi_F \) is the measured phase change of the force sensor, \( \Delta \phi \) is the measured phase change of the reference sensor and \( k \) is the coefficient used to calibrate the difference in the thermal influence of the two sensors.
In the end, the axial force \((F)\) is related to effective phase \(\Delta \phi\) via the following intensity phase-force relationship,

\[
F = \alpha \Delta \phi
\]  

(3.9)

where \(\alpha\) can be calibrated experimentally.

### 3.4 Experimental Results

#### 3.4.1 A preliminary experiment for FPI tip force sensing validation

A tip force sensing needle without temperature compensation was fabricated first to demonstrate the concept of tip force sensing. Figure 3.20 shows the fabrication process of the sensing needle.

![Figure 3.20 FPI tip force sensing needle fabrication procedure](image)

A mechanical frame, with an electronic scale (range 0-5000 g, resolution 1 g, corresponding to 50 N and 0.01 N, respectively), was used to test the FPI sensor. The applied force was increased and then kept stable a while after the change. The force was repeatedly loaded onto and removed from the needle tip. The signal was collected and filtered using wavelet transform analysis.

To study the force and light intensity signal relationship, steady axial forces ranging from 0 - 20 N were applied, at a room temperature of 20 °C, to the needle tip and the noise fluctuation was averaged during the data processing procedure. The experiments were
repeated six times, and the results of the fitted relationship curve, with error bands, is shown in Figure 3.21.

![Figure 3.21 Light intensity and force relationship curve fitting result](image)

As the experimental results show, the FPI sensor has the ability to measure force amplitude up to 20 N. However, it works better when the applied force is smaller than 5 N, which meets the MIS force sensing requirement. The resolution is found to be 0.1 N during calibration. A phantom skin insertion test was carried out as shown in Figure 3.22.

![Figure 3.22 Silicon rubber skin phantom insertion procedure: (a) Silicon rubber phantom insertion (b) Processed insertion signal](image)

The purpose of this test was to record the insertion force at the tip of the needle. A typical insertion force signal is shown in Figure 3.22 (b). From the result, some very useful information about the forces applied during insertion is obtained, which includes insertion force and withdrawing force with their amplitudes, and force impact when the puncture needle goes through the first layer. The entire insertion procedure has four stages. The first force peak appears at penetration through the surface, followed by an insertion force
decrease as denoted by (1) in Figure 3.22 (b). Then, when the puncture needle moves deeper into the rest of the layer, the force went up to the next force peak. After puncturing all the layers, the force gradually went down as denoted by (2) in Figure 3.22 (b). The last stage is to pull the puncture needle out, where the force is in the opposite direction as denoted by (3) in Figure 3.22 (b). The experimental results show the insertion forces measured agreed with the data in the literature [13, 14]. The phantom tissue insertion test showed that the sensor is able to identify the entire insertion and extraction procedure.

3.4.2 Calibration experiments for the temperature compensation tip force sensing needle

Based on the sensor placement investigation in Section 3.2.4 and the temperature compensation algorithm in 3.2.5, an optical fibre tip-force sensing needle was finally fabricated and characterised in the following experiments.

3.4.2.1 Temperature compensation calibration

The coefficient $k$ in Equation 3.8, dependent upon how the FPI sensors were embedded in the needle, needs to be calibrated through experimentation. In this study, the needle was inserted into the chamber with a constant temperature of $37.5\degree C$ from room temperature of $23\degree C$ for about two minutes and then removed, this process being repeated seven times. Figure 3.23 is part of the FPI interference intensity signals, and $L$ is labelled as the low temperature of $23\degree C$, while $H$ is the high temperature being $37.5\degree C$.

The FPI intensity signals of the two sensors in the first 30 $s$ after dramatic temperature change, e.g. $23\degree C$ to $37.5\degree C$, were extracted and analysed, taking into consideration duration of real needle surgery. The intensity phase was calculated and then accumulated over time, and a linear relationship was then found between the two accumulated phase signals, as shown in Figure 3.23, indicating a coefficient $k$ of 0.9515.

The sensing range of the cavity length was deduced to be 29.65 - 30.3 $\mu m$, given the initial cavity length of 30 $\mu m$ at $23\degree C$, a needle temperature change of 14.5 $\degree C$ (e.g. $23\degree C$ to $37.5\degree C$ and $37.5\degree C$ to $23\degree C$) and an axial loading force of 0 - 8 $N$. 
3.4.2.2 Compensation effect against different temperatures

To evaluate temperature dependence after compensation, the needle was left in the chamber and heated to six different temperatures ranging from 23 - 45 °C at 6 °C intervals. At each temperature, the chamber temperature fluctuated up and down within 2 °C for fifteen minutes. Temperature compensation (in terms of the intensity phase) was satisfactory, as shown in Figure 3.25.
To check the effectiveness of the temperature compensation solution alongside axial loading, pulse forces were applied periodically at the needle’s tip during the temperature change of 23 - 37.5 °C, as shown in Figure 3.26. The intensity phase of the force sensor is attributed to both the temperature and force, while the intensity phase of the reference sensor is only attributed to temperature. A mean intensity phase error of 0.03 rad was found after temperature compensation.

### 3.4.2.3 Calibration of the intensity phase-force relationship

The FPI sensing system was calibrated using a commercial dynamic force sensor (PCB208C02, PCB Piezotronics, USA) with a sensing range of 0 - 448 N and a resolution of 0.004 N. The intensity phase of the force sensor after compensation and the applied dynamic forces were recorded (Figure 3.27), and then a linear relationship between them was then obtained (Figure 3.28) to represent Equation. (3.9). The results show that the temperature compensated force sensor has a force measurement range of 0 - 8 N with a resolution of 0.3 N at temperatures of 23 - 37.5 °C.
Figure 3.26  Effective intensity phase after temperature compensation

Figure 3.27  FPI signal and applied force detected by dynamic force sensor

Figure 3.28  Calibration of the coefficient $\alpha$ for Equation. (3.9)
3.4.3 Sensor performance

3.4.3.1 Phantom skin tissue insertion at different temperatures

Insertion experiments were carried out at different temperatures to test temperature compensation performance, with synthetic human skin tissue (SynDaver\textsuperscript{TM} Labs, USA) placed in the chamber. After the tissue was heated to a target temperature, the needle was then inserted into the tissue at a constant speed of 7.25 mm/s. Insertions were performed three times at 37.5 °C, four times at 34 °C, twice at 30 °C and three times at 26 °C, respectively. The intensity signals of the two FPI sensors are shown in Figure 3.29 to Figure 3.32.

During insertions, the FPI sensors went through three main stages: force increasing, skin tissue breaking, and needle retraction. As shown in Figure 3.29, the substantial change in the tip force during the first and second stages is indicated by P1, P2 and P3. The small change in the tip force due to retraction is indicated by R1, R2 and R3. The other minor signal fluctuations are due to noise.

The effective force signals after temperature compensation in Figure 3.29 to Figure 3.32 show that force signals were well captured, while the temperature-induced signal variations were compensated satisfactorily. Comparing the FPI interference intensity change induced by a 3 °C temperature variation, as shown in Figure 3.35, a 14.5 °C temperature change led to a more dramatic signal fluctuation shown in Figure 3.29.

![Figure 3.29 Phantom tissue insertion from 23 °C to 37.5 °C](image-url)
Figure 3.30  Phantom tissue insertion from 23 °C to 34 °C

Figure 3.31  Phantom tissue insertion from 23 °C to 30 °C

Figure 3.32  Phantom insertion from 23 °C to 26°C
3.4.3.2 Insertion force versus insertion depth

To study the effectiveness of the tip force sensing system, a commercial force sensor, with a resolution of 0.001 \( N \), was assembled at the base of the needle. A silicon rubber phantom skin with a thickness of 25 \( mm \) was employed to accomplish the depth insertion experiment with an insertion speed of 5.6 \( mm/s \) at a room temperature of 23 \( ^\circ C \). The relationship between insertion force and insertion depth can be found in Figure 3.33.

![Figure 3.33 Insertion force versus insertion depth](image)

It shows that after the surface layer was broken through, the FPI sensor was able to detect tip force reaching a stable magnitude of 2 \( N \) at a depth of 9 \( mm \), while force measured by the base sensor involved both the frictional force between the needle body and the tissue and the tip penetration force. The depth of 9 \( mm \) is where the sensors were placed in the needle from the tip. Before the surface was broken through, the base sensor and the FPI sensor had the same measurements.

3.5 Temperature monitoring during needle insertion

As mentioned above, the fabricated sensing needle also can be used as a temperature sensor. To evaluate the temperature sensing performance of the sensing needle, a temperature calibration experiment was performed in the chamber with a temperature controller, as shown in Figure 3.34.
Figure 3.34  Temperature calibration experiment setup

During the experiment, the sensing needle and a temperature sensor were attached and put into the chamber with a room temperature of around 23 °C. The chamber was then heated slowly to around 40 °C over 7 minutes. The sensing needle and temperature sensor slowly cooled down to room temperature after heating stopped. The temperature calibration results obtained are shown in Figure 3.35 and Figure 3.36.

Figure 3.35  FPI light intensity versus temperature change
3.6 Conclusion

This chapter described a surgical needle with an embedded, temperature-compensated optical fibre sensor for tip force measurement. It employed two FPI sensors, one serving as the force sensor and the other as a reference sensor. The latter is subjected only to temperature variations and provides a temperature reference for the force measurement of the former. An interference intensity-phase algorithm was proposed for sensing force in a varying temperature environment. It transforms the non-linear interference intensity signal into a linear intensity-phase signal. Experimental calibration results show that the sensing system developed has a resolution of 0.3 \( N \) in the measurement range of 0 - 8 \( N \) at a temperature of 23 - 37.5 °C. Phantom skin tissue insertion experiments at different temperatures and insertion depths verified the effectiveness of the temperature compensation solution and tip force sensing, respectively. This temperature-compensated tip force sensing needle has the potential to recognize tissue types in vivo and diagnose abnormal tissues through analyzing tip force and insertion speed.
Chapter 4

Capability characterization of the force sensing needle via ex vivo experiments

4.1 Introduction

The proposed tip-force sensing needle aims to identify tissue types in vivo via its mechanical properties, by placing a fibre optical tactile sensor at the tip of an instrument. So far, there have been few attempts to enable an instrument to have such sensing capability, such as hand-held sensing palpation probes for minimally invasive tumour localisation [52], a force-sensitive wheeled probe for the localisation of tissue abnormalities during minimally invasive surgery [53]. While a fibre optical sensor based on the principle of FBG was embedded into the body of an epidural needle for epidural space identification [55], the method suffers from needle-tissue friction interference and the thermal influence.

This chapter deals with the capability characterisation of the tip force sensing needle mentioned above. To characterise the needle’s force sensing capability, ex vivo
experiments were conducted using both phantom human tissue and porcine abdomen tissue. Various factors relevant to tissue identification are investigated, with data fields of forces versus types of tissue being established for tissue identification. By means of this database, two tissue identification experiments are performed, where the result shows that 66.7% of the multiple layered tissues and 86.7% of the neoplasm tissue are successfully identified.

4.2 Experimental Design

4.2.1 The sensing needle

In the previous chapter, a tip-force sensing needle was fabricated based on a 14G surgical needle with a bevel tip, and an FPI sensor was placed inside the needle tip. The sensing needle can measure an axial force of 0 - 20 N with 0.3 N resolution at the tip, based on the calibrated linear relationship in Chapter 3. In this novel sensing needle design, we managed to exclude time-varying needle-tissue frictional force and compensate for influence of the temperature when the needle goes from outside a body into inside a body.

![Figure 4.1 The tip force sensing needle system, (a) Laser source, (b) Optical circuit box, (c) FPI sensor location, (d) The sensing needle](image)

The sensing needle system was further improved for portability. As shown in Figure 4.1, it has a hand-held laser source (EXFO FLS-300, CAN.), a compact optical circuit and a sensing needle.
4.2.2 Experimental sensing setup

![Experimental sensing setup diagram](image)

To characterise the sensing needle developed, an experimental rig was set up, with its system components being shown as in Figure 4.2. To validate the removal of the needle-tissue friction and study the influence of needle displacement and velocity, an ATI force sensor (ATI SI-20–1, USA) was mounted at the base of the needle that measured hybrid force, involving both the axial tip force and needle-tissue friction during insertions. To provide a human body temperature environment, a temperature control chamber was fabricated to warm the phantom tissue to around 38 °C, with a precision of 0.5 °C after calibrated via a temperature sensor with a resolution of 0.1 °C. A linear actuator with a speed controller (Progressive Automations Inc, PA-02-6-200, CAN.) was used to perform automated needle insertions to observe the effect of the needle advancing rate, the results of which are compared to those from manual insertion.
During insertion experiments, experimental tissue was placed and clamped inside the temperature chamber, as shown in Figure 4.3, and then warmed to around 38 °C. The sensing needle was driven by the linear actuator at a specific speed, ranging from 3.8 mm/s to 14.5 mm/s, and inserted into the warm tissue. As well as experiments undertaken at body temperature, needle insertions were also performed at room temperature for the purpose of comparison or when experimental tissues were too large to be placed in the temperature chamber. The tip force data were then collected by the optical circuit and analysed in an algorithm. The hybrid tip/friction force data from the ATI sensor were also captured to study the time-varying frictional force. Table 4.1 gives the technical specifications of the setup.

Table 4.1  Experimental setup specifications

<table>
<thead>
<tr>
<th>Sensor</th>
<th>Diameter /mm</th>
<th>Sensing range/N</th>
<th>Sensing resolution/N</th>
</tr>
</thead>
<tbody>
<tr>
<td>FPI sensor</td>
<td>0.4</td>
<td>0-20</td>
<td>0.3</td>
</tr>
<tr>
<td>FPI tip force sensing needle</td>
<td>2.1</td>
<td>0-20</td>
<td>0.3</td>
</tr>
<tr>
<td>ATI force sensor</td>
<td>43</td>
<td>0-60</td>
<td>0.02</td>
</tr>
</tbody>
</table>
4.2.3 Tissue samples

Five kinds of phantom human tissues (SynDaver™ Labs, USA) were prepared for the insertion experiments, i.e. human skin, fat, muscle, liver, and multiple layer abdominal tissue (Figure 4.4). The phantom tissues are made from salt, water and fibre and have on the physical properties of human in vivo tissue regarding tensile modulus, penetration force, the coefficient of friction, and thermal conductivity. Also, a piece of phantom tissue made of silicon rubber with a thickness of 50 mm was also used for studying the influence of needle displacement on the tip force. Table 4.2 gives the specification of the phantom tissues used.

![Phantom human tissues](image)

Figure 4.4 Phantom human tissues from SynDaver™ Labs, USA, (a) skin, (b) fat, (c) muscle, (d) skin-fat-muscle multiple layer tissue, (e) liver.

Table 4.2 Phantom tissue specifications

<table>
<thead>
<tr>
<th>Phantom human tissue</th>
<th>Thickness/mm</th>
<th>Typical insertion force /N</th>
</tr>
</thead>
<tbody>
<tr>
<td>Skin</td>
<td>3</td>
<td>4</td>
</tr>
<tr>
<td>Fat</td>
<td>5</td>
<td>2</td>
</tr>
<tr>
<td>Muscle</td>
<td>10</td>
<td>3</td>
</tr>
<tr>
<td>Abdominal tissue plate</td>
<td>18</td>
<td>-</td>
</tr>
<tr>
<td>Liver</td>
<td>Normal human</td>
<td>-</td>
</tr>
<tr>
<td>Liver</td>
<td>liver size</td>
<td>-</td>
</tr>
<tr>
<td>Silicon rubber</td>
<td>50</td>
<td>2</td>
</tr>
</tbody>
</table>

Note: The phantom tissues were made from salt, water and fibre.
For the purpose of comparison, five types of porcine tissue, i.e. skin, fat, muscle, liver, and kidney, were prepared. The skin, fat and muscle were mainly from porcine abdomen, the mechanical properties of which are similar to human tissues [186, 187]. The skin-fat-muscle tissue from porcine shoulder was also prepared for a comparison with the abdomen tissue. To simulate the abnormal tissue, chicken hearts were buried in porcine muscle as a simulated tumour.

4.3 Experiment Results

Needle tip force during insertion can be affected by various insertion factors. For example, different needle advancing rates may result in large force variations in identical tissue. Due to the inability of a human to maintain a constant velocity during needle insertion, automated insertion is likely to lead to a more stable force result. In addition, the tip force may also change due to different tissue types and locations, needle size, and multiple tissue layers, etc. In this study, to understand the characteristics of the sensing needle for use in tissue identification, different experiments with various insertion conditions were carried out.

To characterise the tip force sensing needle for tissue identification, influential factors were investigated first, including needle advance rate, automated and manual insertion, friction force and needle displacement, needle size and tip shape, different between phantom tissue and porcine tissues, etc. Then, to identify tissue types, a tip force bank was established based on identical tissue insertion experiments.

Based on the properties mentioned above, two experiments were performed to assess tissue identification capability of the sensing needle. One was the multiple layer tissue detection, aimed at tissue layer and tissue type detection. The other was neoplasm tissue detection, performed on porcine abdomen muscle with a simulated tumour buried within it.

4.3.1 Typical insertion stages

Figure 4.5 is a typical tip force of insertion procedure measured by the tip force sensing needle, with an advancing rate of 3.8 mm/s on a two-layered phantom human tissue.
Figure 4.5  Typical insertion force versus needle displacement

The first insertion stage, called boundary displacement, starts when the needle reaches the tissue surface, and ends with rupture of the tissue boundary, as shown in Figure 4.5. The tissue boundary deforms under the load applied by the needle tip and moves at the same advancing rate as the needle (the relative advancing rate is zero) [188]. The relative advancing rate starts to increase when the puncture event occurs, where the stress of the tissue around the needle tip is peaked, resulting in the tissue being cracked and the needle tip breaking into the tissue [189].

After the layer rupture, insertion force drops to a lower value to maintain the tip’s cutting force. According to Hing, et al. [190], the needle tip cutting force could be constant during ex vivo porcine liver insertion, apart from some fluctuations due to further layer ruptures. For example, in Figure 4.5, before needle extraction, two layer rupture events were observed during insertion.

4.3.2 Needle advance rate

Needle insertion force (hybrid tip-friction force) was believed to relate to the needle advance rate in previous research, both in phantom tissue [191] and biological tissue [192]. However, some researchers also found that under specific conditions, the hybrid tip/friction force showed no significant effect on advance rates [193, 194]. In addition, studies on the influence of needle advance rates on needle tip force are scarce. Therefore, the relationship between tip force and needle advance rates must be explicitly measured for the purpose of characterisation.
Three needle advancing rates were considered during the experiments, i.e., 3.8 mm/s, 7.8 mm/s, and 14.5 mm/s, which are in line with needle advance rates in routine surgery, such as interventional radiology procedures (about 8.3 mm/s) [195], epidural procedures (0.4-10 mm/s) [196], and prostate brachytherapy (1-20 mm/s) [197], etc. The experiment was carried out on both phantom human skin tissue and porcine abdomen skin tissue. A same number of 10 times of insertions were performed under each insertion speed and tissue type, which means 30 times of insertions were performed on porcine abdomen skin and another 30 times on phantom skin. Based on the 10 tip force readings of each configuration, standard deviation was calculated, the results of which is shown in Figure 4.6.

Figure 4.6 Insertion tip force with various needle advance rates

It can be seen that the tip force has a nonsignificant relationship with needle advance rates, with an average force of about 8.5 N for porcine abdomen skin and 4 N for phantom human skin tissue under the velocity range. It also shows that the standard deviation of tip force in porcine tissue is larger than phantom tissue which has more consistent properties. The result indicates that manual insertion may achieve similar tip force rates at varied constant advance rates.

4.3.3 Automated and manual insertion

To figure out the influence of insertion modes on needle tip force, automated and manual insertions were conducted on phantom human muscle, with an overall time of around 6s. The needle advance rate was set to 3.8 mm/s for automated insertion, and the operator drove the needle as continuously and stably as possible in the manual insertions. Five
automated insertions and six manual insertions were performed on phantom muscle. The tip force signals of both situations are compared in Figure 4.7.

![Diagram a](image.png)

**Figure 4.7** Automated and manual insertion on phantom muscle, (a) Four stages of one typical insertion, (b) automated and manual insertion signals

From the results, different insertion stages during the procedure can be clearly identified in both situations, indicated by a, b, c, and d in Figure 4.7. However, some differences can be found when the stages changed. In automated insertion, boundary displacement (from a to b in Figure 4.7) took less time, and more time was spent in other stages, especially in the last stage (from c to d). As well as this, the stage changing signals of b and c were more clearly in automated insertion than that in manual insertion, due to fewer force fluctuations through using a linear actuator. In automated insertion, through using a constant advance rate, tissue thickness can be precisely derived. For example, the time
duration of b to c multiplied by the advancing rate could be regarded as the exact tissue thickness. In manual insertion, precision is sacrificed when it comes to measuring tissue thickness.

This experiment shows that tissue identification through manual insertion is possible, despite losing thickness measurement of interacted tissue.

4.3.4 Friction force and needle displacement

Currently, needle surgery is subject to varying friction forces along the needle shaft due to increasing amount of contact area during insertion. However, only little research has measured actual tip force during needle insertion via a specially designed tip force sensing needle [51, 198, 199], some researches calculating tip force via simulation or estimation [200].

To demonstrate the avoidance of friction forces by using the tip force sensing needle, the hybrid tip/friction force was measured by using an ATI force sensor for comparison. Figure 4.8 is the force comparison between the ATI force sensor and the tip force sensing needle, from an insertion into a 10 mm phantom human muscle tissue sample. The insertion procedure had four stages, denoted by a, b, c, and d, as shown in Figure 4.8. It shows that the two force sensors had similar values in the stage of boundary displacement (from a to b), until the layer was ruptured at around 6.7 N. However, after insertion through the entire tissue depth denoted by d, the ATI force sensor kept a steady friction force at around 12.5 N, while the tip force from the sensing needle dropped to zero.

The result indicated that the friction force clearly influenced hybrid tip-friction force after tissue boundary rupture, which is likely to result in an uncertainty of force feedback to clinicians, but it can be addressed using the tip force sensing needle.
Another insertion experiment on silicon rubber phantom tissue of greater large-thickness shows more detailed information about the influence of needle displacement on tip force and needle shaft friction force. Figure 4.9 shows the tip and hybrid tip-friction force signals in one needle insertion under automated insertion. The calculated friction force was obtained by subtracting tip force from the hybrid tip-friction force.

During the insertion procedure, tissue rupture happened when tip force increased to about 0.7 \( N \). After that point, the tip force showed differed from the hybrid force, staying stable within a force range of 1.5 - 2.6 \( N \). While the hybrid tip-friction force kept increasing linearly as expected with the length of the needle in the sample and reached a peak force of 11.4 \( N \) at a needle tip displacement of 40 \( mm \), which was close to the full length of the
needle. Friction force appeared at the rupture of the tissue boundary, and increased linearly with the increment of needle displacement.

The result shows that the tip force is relatively constant in identical tissues during needle insertions, which could be used to estimate tissue types. It also demonstrates that the sensing needle has the ability to eliminate the influence of friction force during deep tissue insertions.

### 4.3.5 Needle size and tip shape

To investigate the influence of needle size, four more tip force sensing needles, three bevel tip needles in sizes 17G, 18G, and 19G, and one 18G flat tip needle, were fabricated. The comparison experiments were conducted on phantom tissue. Besides, this insertion force data of other needle size and tip-shapes were also collected from the literature for a comparison to reach a more precise conclusion. The insertion force was sampled at the tissue rupture event which can be regarded as tip force, as there is no friction influence involved at that moment. Table 4.3 shows the tip force data for different needles.

<table>
<thead>
<tr>
<th>Needle type</th>
<th>Mean tip force (N)</th>
<th>Tissue type</th>
<th>Advance rate (mm/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>14G, 18G</td>
<td>10, 2.1</td>
<td>Porcine skin [201]</td>
<td>3.8, 15</td>
</tr>
<tr>
<td>17G</td>
<td>1.18</td>
<td>Phantom skin</td>
<td>3.8</td>
</tr>
<tr>
<td>18G, 17G</td>
<td>0.6, 0.64</td>
<td></td>
<td></td>
</tr>
<tr>
<td>19G</td>
<td>0.31</td>
<td></td>
<td></td>
</tr>
<tr>
<td>18G, 19G</td>
<td>0.6, 1.1</td>
<td>Porcine heart[202]</td>
<td>5</td>
</tr>
<tr>
<td>14G</td>
<td>50</td>
<td>Silicone rubber [203]</td>
<td>0.8</td>
</tr>
<tr>
<td>19G, 25G</td>
<td>18, 5</td>
<td></td>
<td></td>
</tr>
<tr>
<td>22G, 22G</td>
<td>0.14, 0.24</td>
<td></td>
<td></td>
</tr>
<tr>
<td>18G</td>
<td>0.31</td>
<td>Phantom [204]</td>
<td>3</td>
</tr>
<tr>
<td>15G, 10G</td>
<td>0.54, 1.45</td>
<td></td>
<td></td>
</tr>
<tr>
<td>15G, 15G</td>
<td>1.4, 0.54</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

81
The results indicate that a cone needle tip has the smallest tip force of all types of needle tip shape, being similar to the triangular shape. The bevel tip has a slightly larger tip force. Blunt needle tips have the largest tip force, far larger than other types.

Needle size has an estimated influence on needle tip force, increasing with the increment of needle size. We assume that the two needles of different outer diameters have the same force-diameter relationship in all types of tissues. For example, the 17G needle has a tip force twice as high as an 18G needle in phantom skin insertion. Supposing that during insertions of other types of tissue, 17G needles still have a tip force twice as high as an 18G needle. The relationship between needle outer diameter and tip force in Table 4.3 is summarized in Figure 4.10, which shows tip force increased linearly with needle size.

<table>
<thead>
<tr>
<th>Shape</th>
<th>Tip Force</th>
</tr>
</thead>
<tbody>
<tr>
<td>Triangular</td>
<td>0.3</td>
</tr>
<tr>
<td>Cone</td>
<td>0.28</td>
</tr>
</tbody>
</table>

Figure 4.10 The tip force of various sized bevel-tipped needles in porcine skin insertion

**4.3.6 Phantom tissue and porcine tissue**

It can be expected that the tip force sensing needle records different tip forces during insertion into various types of tissues, due to their different mechanical properties. Therefore, a tip force bank for different types of tissues needs to be established for tissue identification. For this purpose, needle insertions on different kinds of phantom human tissue and porcine tissue were conducted. Four insertion velocities were applied to the needle, 3.8 mm/s, 7.8 mm/s, 11.3 mm/s, and 14.5 mm/s. Each kind of tissue was inserted
individually ten times for each needle advance rate. In total, 40 insertion events were executed on every type of tissue sample.

The insertion results in Figure 4.11 and Figure 4.12 show that insertion forces on the identical tissues vary within a specific range for different needle advance rates, but no general tendency was found. Basically, the tip force ranges for phantom skin, fat, muscle, and liver are 2.5 - 4 N, 2.4 - 3 N, 5.4 - 7.3 N, and 4.3 - 5 N, respectively. For porcine abdomen skin, fat, muscle, liver, and kidney, the tip force ranges are 17 - 21 N, 6.6 - 10 N, 9 - 10 N, 3.2 - 6 N, and 3.3 - 5.2 N, respectively. However, distinct force variations were observed between every two adjacent tissue types, i.e. a 10 N force difference was found between porcine skin and fat, and a 3 N difference between porcine fat and muscle. The tip force of porcine tissues was far larger than the force data from published literature, i.e. 0 - 4 N for skin, liver and kidney with an 18G needle [201], which is mainly due to the larger needle diameter in this study. The results also show that some internal organs such as liver and kidney may have very similar insertion tip forces at around 4 - 5 N. However, this would have no influence on tissue identification as they lie in different areas of the body.

It was noticed that force values of abdomen skin exceeded the sensor measuring range of 0-20 N. However, the readings were still regarded reliable results, as they were calculated based on a linear relationship between the applied force and the FPI light intensity phase. The sensing needle can record actually the force larger than 20 N.

![Figure 4.11 Tip force during insertion of various phantom tissues](image)

Figure 4.11 Tip force during insertion of various phantom tissues
Apart from tissue types, the same type of tissue found in different locations also tends to have different tip forces due to different tissue densities. As shown in Figure 4.13, skin and fat in porcine abdomen are obviously sloppier than that in porcine shoulder. The information about insertion location factors is scarce, with very limited results. Podder, et al.[205], Yan, et al.[206], and Langevin, et al.[188] investigated axial forces in different zones of the human body, but, without clear conclusions.

To investigate tip force differences induced by varied tissue location, three types of tissues including skin, fat and muscle from porcine abdomen and shoulder, respectively, were employed in this experiment (Figure 4.13). Each type of tissue was isolated from the others and tested individually. The tissue thicknesses for skin, fat and muscle are about 1 mm, 10 mm, and 20 mm, respectively. Results in Figure 4.14 are the tip forces of each kind of tissue.
An obvious difference was observed from the same type of tissue from different locations. For example, abdomen skin had an average rupture force of 17.2 N, while it was 26.4 N for shoulder skin. Shoulder fat required a relatively high rupture force (around 11.4 N), even higher than that of shoulder muscle (around 6.7 N). Furthermore, two different adjacent tissues, such as abdomen fat and abdomen muscle, still had clearly different tip forces in each location. Therefore, tissue identification is possible through needle insertions at a specific site.

4.3.7 Multiple layer tissue insertion

After having obtained a data field of tip forces for each type of tissue, the sensing needle has the potential to identify the tissue types during insertions. To assess its tissue identification capabilities, insertion experiments on multiple layer tissues need to be done. Two groups of multiple layer tissue were prepared, i.e. phantom skin-fat-muscle-liver tissue and porcine abdomen fat-muscle-liver tissue. Unlike the phantom tissue, the skin was excluded as it requires rupture forces much higher than in vivo human skin, giving rise to a sudden penetration of multiple adjacent tissue layers.

Figure 4.15 is the tip force signal during insertion of phantom tissue with an advance rate of 7.8 mm/s. Figure 4.16 is the tip force sensed during the insertion of the porcine fat-muscle-liver tissue.
During insertion, layer rupture information was the most important signal, which providing very useful information, such as tissue thickness and average tip force. Figure 4.15 and Figure 4.16 show that layer rupture information can be clearly captured by the tip force sensing needle. The layer rupture of skin, fat, muscle, and liver are denoted as a, b, c, and d, respectively, in Figure 4.15 as a good example. It also indicates that after rupture of each tissue layer, tip force always drops to some extent, maintaining a stable value during needle advancement. There is also a clear small force drop after the stable force stage when the needle comes through one type of tissue layer and reaches the next layer, denoted by a', b', and c' in Figure 4.15. Then, further boundary displacement started towards to the next layer rupture. In Figure 4.16, the signal denoted A also suggests that the needle tip was leaving the muscle layer, followed by the boundary displacement of liver.
Tissue type can be confirmed according to a tip force database for various tissues, such as the data in Figure 4.12, along with the help of a knowledge of anatomy. Consequently, a tissue layer sketch can be precisely drawn for clinicians during needle insertions.

4.3.8 Neoplasm detection

Neoplasm is an abnormal growth of tissue, which is commonly referred to as a tumour [207]. Another important application of the tip force sensing needle is to identify neoplasm tissue during insertion.

Previous researches by Swaminathan, et al. [208] suggested that a tumour often has higher toughness than surrounding tissue. Based on this hypothesis, a neoplasm tissue detection experiment was designed as shown in Figure 4.17 (a). Chicken hearts, bought from the supermarket, were positioned inside the porcine abdomen muscle as a simulated neoplasm tissue. The tissue thicknesses for simulated tumour and the muscle layer upon it are about 20 mm and 5 mm, respectively. The needle was then inserted automatically into the multiple layers, and the signal was collected and analysed.

![Figure 4.17 Neoplasm tissue identification, (a) Neoplasm tissue setup (b) Tip force signal of tissue identification](image)

Figure 4.17 is a typical tip force signal of one needle insertion. It shows that the muscle was ruptured at 8.7 N. After 4 mm of needle displacement, another tissue layer was detected, where the layer rupture force was around 10.8 N, which is higher than the muscle layer rupture force. Therefore, it can be confirmed that a different kind of tissue exists beneath the muscle tissue.
4.3.9 Performance assessment

The performance of tissue identification was assessed by the accuracy of tissue identification and neoplasm tissue location, which was defined as the proportion of tests that correctly identified tissue layer rupture or neoplasm tissue rupture. In multiple layer tissue insertions and tumor-in-muscle insertions, the tissue layers were assembled manually before insertions, as known structures. If the sensing needle could identify all of the layers, then it was regarded as a successful identification.

In multiple layer tissue identification, 15 insertions were performed, and 66.7% (10/15) of tissue layers were successfully detected. In neoplasm tissue detection, a higher success rate of 86.7% (13/15) of 15 insertions was observed.

Our experiments show that the main factor influencing tissue identification is that some layer rupture signals exhibit very little difference to those of surrounding tissue, which is hard to recognise. The greater accuracy of tumour identification is due to the fact that simulated neoplasm tissue had an obvious difference in stiffness to the surrounding tissues, giving rise to a clear force variation.

4.4 Conclusion

This chapter described the capability characterisation of a fibre optical tip force sensing needle for tissue identification via ex vivo experiments under simulated human body temperature conditions. To characterise the sensing needle for tissue identification, a series of insertion experiments were carried out on both phantom human tissue and porcine abdomen tissue. Various tip force-relevant factors were investigated, such as needle advance rate, needle displacement, needle size, tissue type and location, insertion methods, etc. It was found that the tip force sensing needle could obviate the influence of friction force. Tip force could be affected by tissue type and location, needle size and tip shape, but shows little connection to needle advance rate within the range of 0 - 15 mm/s, as well as needle displacement. Needle insertion methods, automated and manual insertions, only influence the time of insertion event rather than the insertion forces themselves. However, manual insertion did lose precision in tissue thickness
measurement, compared with automated insertion. Added to this, a tip force database was established for phantom tissue and porcine tissue identification.

Based on these characteristics, to assess the sensing ability of needle to identify tissue, multiple layer tissue detection and tumour tissue concealed within muscle tissue detection were performed. The results show that the tip force sensing needle has the potential to identify various tissue layers and detect tumour through analyzing tip force variations, with accuracies of 66.7% and 86.7% for tissue identification and simulated tumour identification, respectively. The results indicate this fibre optical tip force sensing needle has the potential to perform tissue identification in in vivo needle surgeries.

There are some related works should be carried out for clinical use and potential of the developed sensing needle. For example, the current sensing needle cannot diagnose the tumour as benign or malign condition. It is possible to achieve that once the database are collected covering tip forces of various tumours, if benign and malign tumours have different mechanical properties. Also, in vivo insertion experiment is to be conducted before clinical use of the developed sensing needle.
Chapter 5

Experimental validation of the force sensing needle via in vivo tissue identification on mice

5.1 Introduction

In vivo experiments on animals play a very significant part in medical device development, moving further towards practical application. In this chapter, ex and in vivo tissue identification experiments on mice using a tip force sensing needle were conducted.

Firstly, a new sensing needle with a smaller outer diameter of 1.47 mm and greater length of 80 mm is introduced, with given calibration results. The experimental setup, subject preparation, and needle insertion experiment designs are then described. Two groups of mice experiments are presented. One is ex vivo tissue insertion, conducted on dead mice of different times of death to study the influence of time of death on needle tip force, as
well as influence of temperature. The other is an in vivo tissue insertion experiment conducted on anaesthetized mice for a comparison with ex vivo experiments. A tip force database of the internal organs of mice was also obtained for the tissue identification via insertions of fresh mice organs which were anatomised within one minute after death. Two types of experiments, skin-tumour-skin insertions and mice torso insertions were carried out using manual insertion to assess the tissue identification performance of the sensing needle.

The animal experiments followed protocols approved by the University of Auckland Animal Ethics Committee (AEC), having the AEC reference number: #1781. The operations conducted on alive mice, including mice preparation, tumour cell culture, tumour planting, mice anaesthesia, and in vivo needle insertions, were operated by Xinjian Mao (PhD candidate of Faculty of Medical and Health Sciences, the University of Auckland), who is named on the ethics approval. His research relates to tumour culture and observation. Dead mice were the disposed productions from his experiments.

5.2 Experimental Design

5.2.1 The sensing system

5.2.1.1 New sensing needle characterisation

As mice tissues have low stiffness, a new sensing needle with a smaller diameter needs to be fabricated for better sensation. To optimise the FPI sensor for a better sensitivity, new hollow round quartz capillaries (CM Scientific, UK) with an inner diameter of 0.15 mm and an outer diameter of 0.25 mm were used to fit a needle with a smaller diameter. A 17G tip force sensing needle with a diameter of 1.47 mm and a length of 80 mm was produced for mice tissue identification, as shown in Figure 5.1. The cavity length of the force FPI sensor and the reference FPI sensor are 31.25 μm and 37.5 μm, respectively.
Using the same calibration methods mentioned in Chapter 3, its temperature compensation properties and force-intensity phase relationship were obtained, shown in Figure 5.2 and Figure 5.3, respectively. Figure 5.2 (a) shows the FPI interference light intensity change of two embedded FPI sensors induced by temperature change. Their intensity-phase changes were calculated, as well as their accumulated phase changes, shown in Figure 5.2 (b) and Figure 5.2 (c), respectively. The relationship between the accumulated phase changes of the two FPI sensors was obtained through curve fitting, as shown in Figure 5.2 (d). The temperature compensation then could be achieved based on this.
Figure 5.2  The tip force sensing needle calibration results (a) FPI sensor interference light intensity signals (b) FPI light intensity phase change (c) FPI intensity phase change accumulation (d) Curve fitting the relationship of the accumulated phase changes between the two FPI sensors

By leveraging a commercial ATI force sensor, the relationship between phase change and applied force was finally gained, also via curve fitting, the result of which is shown in Figure 5.3. As the internal organs of mice have very little stiffness, compared to mouse skin, it is necessary to calibrate force under 1 N. Therefore, the sensing needle was calibrated with small force loading, the results of which are shown in Figure 5.4.

5.2.1.2 Experimental setup

The experimental setup is shown in Figure 5.5. Apart from tip force sensing system introduced in previous chapters, a camera was used for recording the needle insertion
procedure. As mice tissues are too soft to be fastened during needle insertions, it was placed on the table or held by hands. The tissue identification was based on manual operations to achieve varying insertion directions.

![Figure 5.5](image)

Figure 5.5 The fibre optical tip force sensing system, (a) Optical circuit box, (b) Sensor input, (c) Laser light input, (d) Laser source, (e) Laptop, (f) Tip force sensing needle, (g) Camera

5.2.2 Mouse preparation and experimental design

- **Mouse preparation**

The tumour cells, the human colorectal adenocarcinoma cell line HCT116 cells from American Type Culture Collection (Manassas, VA), were prepared first. Specific pathogen-free female CD-1 homozygous nude mice (approximately 25g body weight) were used, derived from breeding mice supplied by Charle River Laboratories (Wilmington, MA). About 5,000,000 cultured tumour cells were injected into each mouse at their rear flank, shown in Figure 5.6. After three to four weeks, the surviving tumour tissue had grown to a diameter of about 20 mm.
**Experimental design**

In total, 28 mice with tumours and four mice without tumours were prepared for the needle insertion experiments. For the in vivo experiment, the mice were first anaesthetized, using gas anesthesia as shown in Figure 5.6. In the ex vivo experiments, euthanasia was carried out on mice just before the needle insertion experiments. To study the difference between in vivo and ex vivo insertion, skin-tumour-skin insertion experiments were conducted. Skin-tumour-skin insertions, lateral-direction abdomen insertions, insertions from anus to head, and internal organs insertions were carried out during ex vivo experiments to study various sensing objectives.
5.3 Experimental Results

A series of needle insertion experiments were carried out either ex vivo or in vivo on mice to:

1) Validate the feasibility of tip-force sensing in an in vivo environment
2) Obtain tip force database of different tissues for mouse tissue identification
3) Localize tumours and identify other tissues such as internal mouse organs
4) Assess performance of tissue identification

The sensing needle was manually inserted into the mouse tissue held by hand on a table. To investigate the influence of time of death, needle insertions were performed several times after death, i.e. 10s, 10 mins, 30 mins, and 120 mins.

During the experiments, needle tip forces of various mouse organs, including tumours, were collected to form a tip force database for tissue identification. Various ex vivo tissue identification experiments were performed based on the tip force database. During the in vivo experiments, a tumour localization experiment was repeated to be compared with the ex vivo experimental results in terms of the differences in temperature compensation and tip force sensing.

5.3.1 Tip force change in mouse tissue versus the time of death

The time of death influences two aspects of mouse tissue: the body temperature changes and the tissue decays, both affecting the tip force measurement. Therefore, it is necessary to figure out the influence in these two aspects.

- Temperature change versus the time of death

To investigate the influence of temperature, the mice body temperature was measured with an electrical temperature sensor, based on the DS18B20 digital thermometer with a precision of ±0.5 °C. The temperature was collected from the time of death up to two hours after death. Figure 5.7 shows the temperature measurement setup, where the sensor
was inserted into a mouse from its anus. Temperature change after death over time was recorded in Figure 5.8.

![Figure 5.7 Experiment setup of temperature recording after death](image)

![Figure 5.8 Temperature change post mortem](image)

It shows that the body temperature dropped gradually within 35 mins after death, from about 36 °C to near room temperature, yielding to a polynomial relationship. It also indicates that a mouse one minute after death could be regarded the same as a live mouse in terms of temperature and living organs.

Figure 5.9 is the FPI interference light intensity signal of two skin-tumour-skin insertions and one abdomen insertion on a mouse with a death time of two hours.
The result shows that there was still significant temperature influence that needs to be compensated for after two hours of death due to the temperature difference between the mouse’s body and room temperature.

The above results demonstrate that the sensing needle needs temperature compensation at least within two hours after death of the mouse. It can also be estimated that experimental defrosted animal tissue would take more than two hours to reach room temperature, which would also require temperature compensation if employing fibre optical force sensors.

- **Tip force versus the time of death**

Biological tissue decay tends to change its physical properties. No studies were found that systematically report the difference in needle insertion forces between living and dead tissue [209], with limited studies touching upon this area [210, 211]. Kobayashi, et al. [210] inserted porcine liver both ex vivo and in vivo with a 17G bevel tip needle, and concluded that the stiffness of vivo tissue was two to six times as high as in vitro. However,
Boessenkool, et al. [211] suggested that the insertion force measured in an ex vivo spine insertion was in the same range as living tissue on a male Yorkshire piglet. Therefore, the difference between in vivo and ex vivo insertions needs to be clarified, providing a reference for other animal researches.

To discover the effect of time of death on tip force during needle insertions, needle insertions were conducted on dead mice with times of death ranging from 10s to 120 mins to study tip force changes. Figure 5.10 shows the tip force of skin insertions, which shows there was nonsignificant influence as time of death increased up to two hours.

![Figure 5.10 Tip force changing of mice skin since death in two hours](image)

5.3.2 **Validation of temperature compensation of the sensing needle in ex vivo experiments**

Needle insertions were conducted on mice which were dead for only 10s, with a similar physical condition equivalent to in vivo experiments. Two needle insertions and withdrawals were performed on each mouse on the tumour location. Figure 5.11 and Figure 5.12 are typical insertion signals, where Figure 5.11 is the original FPI light intensity signal, and Figure 5.12 is the processed tip force signal after temperature compensation.
There are two stages in one insertion: the insertion and the needle withdraw, denoted a and a' in the first skin-tumour-skin insertion, respectively, and b and b' in the second insertion. The reference sensor signal in Figure 5.11 shows that temperature influence was very significant, especially at the beginning of the insertion going from the room temperature to the mouse’s body temperature. When the needle was inside the mouse’s body, the temperature did not have much effect on the force signal. The reference sensor also shows the stage changes during the procedure. For example, the point e in Figure 5.11 represents the moment that the needle came out from the skin on the other side to air, therefore, the temperature change tendency flipped. Similarly, the point f indicates
that the needle was being withdrawn back but inside the tumour, therefore, the temperature tendency flipped again. The point g indicates the needle was fully withdrawn from the mouse body to air.

Figure 5.12 demonstrates that temperature compensation gives the sensing needle the ability to overcome the influence of temperature and can capture tip force signal during in vivo needle surgeries. It further validates that the tip force sensing needle has the potential to be used in in vivo applications.

5.3.3 Tip force comparison between in vivo insertion and ex vivo insertion

![Images of mice anesthetization experiment setup and in vivo insertion experiment]

Figure 5.13 Mice anesthetization experiment setup and in vivo insertion experiment, (a) Initial anaesthesia, (b) Anaesthesia setup for mice during insertion experiments, (c) Needle insertion on live anesthitized mouse

In vivo needle insertions is further proof of the temperature compensation effect of the sensing needle. The purpose of this section is to compare the difference between these two conditions, through conducting skin-tumour-skin insertion experiments. As shown in Figure 5.13, mice were first anaesthetised in a box (Figure 5.13 (a)), and then moved to a tube with anaesthetic gas flow, putting the mouse’s mouth inside the tube for continuous anaesthesia during insertion experiments (Figure 5.13 (b)). The operator could then start the in vivo insertion experiments, as shown in Figure 5.13 (c).
Four mice were sampled in this experiment. Figure 5.14 shows a typical FPI interference light intensity signal during the insertion, and its processed tip force signal after temperature compensation is shown in Figure 5.15. The signal of one ex vivo insertion on the skin-tumour-skin with 10s time of death was added into Figure 5.14 for comparison.

![Figure 5.14 In vivo insertion FPI light intensity signal](image1.png)

![Figure 5.15 In vivo tip force signal after temperature compensation](image2.png)

The tip force result indicates that in vivo insertions match well with the ex vivo insertions on mice one minute from time of death in terms of the tip force range and similar original FPI sensor signals, as compared to the added ex vivo signals and those in Figure 5.11 and Figure 5.12. This experiment shows a database based on freshly-killed mice bodies could be regarded equivalent to live mice, which could be used for real-time in vivo tissue identification during needle insertions.
5.3.4 The database of tip force for difference organs

It can be expected that the tip force sensing needle measures different tip forces during insertion of various types of tissue due to their different mechanical properties. A tip force database of different types of tissues is, therefore, necessary for tissue identification. Three mice were killed and dissected for collecting tip force data of various organs, as shown in Figure 5.16.

Figure 5.16 Murine anatomy for internal organ insertion

For this purpose, needle insertions were performed manually on different tissues, including skin, muscle, tumour, and other internal organs. The main organs for insertion are shown in Figure 5.17. Small internal organs, such as heart and lung, were inserted individually three times in each mouse, except skin, muscle, and tumour, with five
insertions. In total, nine insertions were conducted on each kind of internal organ and 15 insertions for skin, muscle, and tumour.

Figure 5.18 shows the tip force database for different tissues and internal organs. The leg muscle required the largest tip force for penetration, with an average force of around 3.45 N. It was followed by skin with 2.53 N, and the largest standard division, and then the heart with 2.53 N. Other internal organs showed a force range of 0.3 - 0.9 N, depending on the tissue type. The tumour showed a tip force difference from the other types of organs, which can be used for tumour identification.

![Tip force database for different tissues and internal organs](image)

**Figure 5.18 Tip force per various murine mouse tissue during insertion**

### 5.3.5 Multiple layer tissue insertion

Three types of multiple layer insertion experiments were designed, skin-leg muscle, the lateral direction of the abdomen, and from anus to head. Skin-leg muscle was conducted in the abdomen after anatomy, while the others were insertions first and then dissection was undertaken for confirmation, as shown in Figure 5.19.
Figure 5.19 Multiple layers insertion experiments, (a) Skin-leg muscle insertion after anatomy, (b) Anatomy after insertion of the lateral direction of the abdomen, (c) Anatomy after insertion from anus to head

Figure 5.20 shows the tip force results for two needle insertions of skin-leg muscle. The needle penetrated the skin first at around 1.2 N and 1.9 N, respectively in the two insertions. It reached leg muscle and broke the muscle layer at around 3.3 N.

Figure 5.20 Murine skin-leg muscle insertion

Figure 5.21 is the insertion result of the abdomen in a lateral direction. It was found that the needle only penetrated the skin, leg muscle and another skin layer on the other side. The dissection confirmed that none of the internal organs were penetrated, which is common in abdomen insertions. The reason is due to that the internal organs tend to avoid needle penetration, as they are too small and soft, especially as they have a protective fluid to keep them hydrated, which enables them to move positions easily when pressed by force.
Another experiment was performed for internal organs identification. The needle was inserted as stably as possible into the mouse body from the anus to head. Figure 5.22 (a) gives a force-time history of one insertion. The result shows that there were several organs experiencing penetration. According to needle displacement and mice organ location, the possible organs are colon, jejunum, liver, lung, and heart. It was confirmed via dissection that the penetrated internal organs were colon, jejunum, liver, and lung.

During the experiment, it was found again that organs are likely to move due to needle tip force. In addition, the internal murine organs may cause force responses, within the

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force database, but the organ slips away from the needle tip, which results in failure in identification.

5.3.6 Neoplasm detection

Neoplasm is an abnormal growth of tissue, which is commonly referred to as a tumour [207]. Ten mice with tumours were employed for neoplasm detection. Within one minute of death, the needle was inserted into the mouse from skin to tumour, coming out of the skin on the other side, as shown in Figure 5.23 (a). The other experiment was carried out for multiple tissue insertion, the needle was inserted into the skin, tumour, and then internal organs in the abdomen, until it broke the skin on the other side of the abdomen, as shown in Figure 5.23 (b).

![Figure 5.23](image1.png)

(a) Skin-tumour-skin insertion, (b) Multiple tissue insertion

Figure 5.23 Two types of insertions for tumour detection, (a) Skin-tumour-skin insertion, (b) Multiple tissue insertion

Figure 5.24 is a typical signal of this procedure. The sensing needle captured the layer breaking force well, and showed stable tip force throughout tumour tissue.

![Figure 5.24](image2.png)

Figure 5.24 Typical signal of skin-neoplasm tissue
A typical signal of the second experiment is presented in Figure 5.25, where shows possible tissue of skin, tumour, cecum, and skin on the other side. The following dissection confirmed the penetration of these organs.

![Figure 5.25 Neoplasm detection in multiple tissue insertion with possible tissue identification, (a) Skin, (b) Tumour, (c) Cecum, (d) Skin](image1)

A lateral abdomen insertion on mice without tumours was also carried out for a comparison. Figure 5.26 is a typical signal of this experiment. It shows a clearly difference with the above experiment, with force information for only skin on both sides of the abdomen. During insertion, it was observed that the shape of the mouse’s abdomen changed significantly, the layers of skin nearly came together during insertion due to the pressure of the needle.

![Figure 5.26 Lateral abdomen insertion on mice without neoplasm tissue](image2)
5.4 Performance assessment

The performance of tissue identification was assessed by the accuracy of tissue identification and neoplasm tissue location, which was defined as the proportion of tests that correctly identified experiencing tissue layer rupture or neoplasm tissue. In skin-tumour-skin insertion experiments, due to the clear force differences between skin and tumour, all the samples were successfully identified. However, in the lateral abdomen insertions, only a few internal organs were penetrated and identified, most internal organs being too small and their positions moving when contacted by insertion force. In the internal organ insertions, from anus to head, only the main organs were penetrated and identified successfully, while smaller organs missed penetration, which was confirmed by dissection.

5.5 Conclusion

In this chapter, an optimised tip force sensing needle with a smaller diameter of 1.47 mm and a longer length of 80 mm was fabricated and calibrated for murine tissue insertion. The concept of tip force sensing and tissue identification were further validated via ex vivo and in vivo insertion experiments on mice.

The experimental setup, preparation of the mice, and experimental designs were introduced firstly. Then, two types of experiments were carried out on the mice: ex vivo tissue insertions and in vivo tissue insertions. Ex vivo tissue insertions were conducted on dead mice with various the time of death to study the influence of time of death on needle tip force, the influence of temperature, as well as tissue identification. The other experiment was in vivo tissue insertion conducted on anaesthetized mice for a comparison with the ex vivo experiments.

The temperature experiment and insertions undertaken two hours after death show that temperature compensation is necessary for mice that had died within two hours due to the internal temperature differences with room temperature. Mice which had been dead for only one minute had very similar characteristics to live mice in terms of body temperature and tip force, which means they are equivalent to in vivo experiment samples.
A tip force database of mouse internal organs was established for tissue identification through dissected mice within one minute of death. Two types of experiments, skin-tumour-skin, and body insertion from two directions were carried out by manually. Based on the results of the above experiments, the performance of the tip force sensing needle was assessed.

It was found that murine skin, leg muscle, heart, tumour, and stomach tissue have greater tip force during insertion, while other internal organs have tiny tip forces, under 0.5 $N$, which is below the threshold of human perception. The tip force sensing needle has the capacity to identify most of the organs penetrated during needle insertion. It was also found that smaller internal organs are likely to move position due to the force applied by the needle tip. The ex vivo and in vivo experiments further demonstrated the potential of the sensing needle for use in in vivo tissue identification applications.
Chapter 6

Epidural space identification on ex vivo porcine spines

6.1 Introduction

Epidural space insertion is a surgery performed on a daily basis, in cases of cervical radiculopathy, lower extremity surgery, labour pain relief, interventional pain management, etc. [212-214]. In 1921, Sicard and Forestier first described a technique, named, loss of resistance (LOR) to identify the epidural space [215]. The epidural needle is inserted to pass through skin, fat, supraspinous and interspinous ligaments, the ligament flavum, and then reaches the epidural space. As epidural space is under lower pressure compared with the surrounding ligament tissues, clinicians can feel resistance force or the pressure change by hand during the procedure, with the leverage of air or saline in a syringe. Currently, LOR and its improved methods are the most common means for epidural space identification.

However, the LOR technique still has drawbacks that need to be overcome. For example, it requires experience to feel the change in resistance during the procedure, and
inexperienced operators can give rise to a 5% - 15% failure rate [216]. The epidural needle can be easily blocked by tissue during insertions, which results in unreliable outcomes. These “blind” attempts can cause spinal cord damage, which can lead to serious complications, including peripheral nerve damage, dural puncture headache, and partial analgesia [217, 218]. Moreover, clinicians often become hesitant to advance the epidural needle due to the unknown position of the needle tip, which results in a more difficult and time-consuming procedure.

It is, therefore, necessary to develop safer techniques to improve the efficacy of the epidural injection and reduce side effects. An integrated force sensor is a new way of detecting the epidural space developed recently. Boessenkool et al. [211] conducted an experimental study on the ligament flavum of piglets to compare force and pressure feedback. The result indicated the sensing of force could lead to a quicker response than pressure feedback. Currently, only a few researchers have mentioned mounted force or pressure sensors for epidural space identification [55, 211, 219, 220]. Most of them were integrated at the base or along the epidural needle, which is subjected to greater friction force induced by surrounding ligaments. To the best of author's knowledge, there are no force sensors embedded at the needle tip.

To provide clinicians with better real-time visual signals, this chapter presents a novel tip force sensing needle with a fibre optical sensor mounted at its very tip. Based on surgical needle requirements, an 18G tip-force sensing needle was firstly designed, and then characterised via a commercial force sensor. To assess the feasibility of epidural space identification, a trial experiment on porcine ligaments/tendons was conducted. Repeated porcine spinal insertion experiments were performed on epidural space identification. The results show that the sensing needle can provide reliable force information to distinguish epidural space from surrounding tissues.

### 6.2 Sensor design

#### 6.2.1 Design specifications

Sensor specifications, including sensing range, resolution, and sensor size, are significant and need to be in line with epidural space insertion surgical properties and the
instrumental requirements of MIS. According to Boessenkool, W [211], ligament flavums have insertion force ranges from 6 - 8 N using an 18G epidural needle, which is the greatest insertion force compared with surrounding tissue. In the epidural space, fat is the main tissue between the ligament flavum and the spinal cord, which requires the smallest insertion force of around 0.5 N. Consequently, using an 18G needle, the force sensing range was specified to be 0 - 10 N, with a force resolution of less than 0.5 N. Considering the anaesthesia function of the needle in epidural surgical applications, the diameter of FPI sensor should be designed to be as small as possible to save space for medical fluid. As well as this, to distinguish the layer breaking signal, the force sensor should be embedded as close as possible to the needle tip. The sensor’s specifications are summarised in Table I.

<table>
<thead>
<tr>
<th>Sensor specifications</th>
<th>Expected values</th>
</tr>
</thead>
<tbody>
<tr>
<td>Force type</td>
<td>Tip axial force</td>
</tr>
<tr>
<td>Sensing range</td>
<td>0 - 10 N</td>
</tr>
<tr>
<td>Sensor resolution</td>
<td>&lt;0.5 N</td>
</tr>
<tr>
<td>Sensor size</td>
<td>&lt;0.3 mm</td>
</tr>
<tr>
<td>Sensor position</td>
<td>Exact needle tip</td>
</tr>
</tbody>
</table>

6.2.2 The epidural sensing needle design

Fabrication of the sensing elements consists of three subtasks in this chapter, which are: FPI sensor fabrication, embedding the sensor in a bevel tip needle, and embedding the sensor in an epidural needle. The FPI sensor fabrication and the bevel tip force sensing needle fabrication are the same procedures as those described in Chapter 3, while the epidural tip force sensing needle has a slightly different texture from the former in terms of the FPI sensor placement.

The two types of sensing needle have different purposes, the bevel tip needle is designed for experimental trials to investigate the feasibility of epidural space identification, while the other is for simulated testing of daily surgical procedures.
6.2.2.1 FPI tip force sensing needle prototype

To investigate the feasibility of epidural space identification, a preliminary test was designed. An FPI sensor was embedded into a bevel tip needle, shown in Figure 6.1. To detect the epidural space as soon as possible, the sensor was attached as close as possible to the needle tip. In this design, a 0.5 mm distance from the capillary end to the needle tip was set during fabrication. Theoretically, friction force has little effect on the tip-force signal after the 0.5 mm distance has been passed. If this property could be validated, the sensing needle would have the potential to detect epidural space within 1 mm of entry into the epidural space.

![Figure 6.1 Sensor design and tip-force sensing needle fabrication](image)

As the FPI sensor can be severely affected by temperature variation, temperature compensation is necessary for epidural identification in a practical application. After the preliminary test, another sensing needle was designed, based on an epidural needle used in current routine surgery. In this study, a reference FPI sensor was used, similar to that mentioned in Chapter 3, to provide temperature information, except that the sensing structure was optimized for epidural identification. As shown in Figure 6.2, the epidural needle has a special tip that protects vulnerable organs such as the spinal cord as much as
possible. This structure results in a change in the temperature compensation design in regard to sensor placement inside the needle. The FPI sensor used as the force sensor was moved forward to the exact tip of the needle, compared with the previous design. Another difference from the previous design is that this FPI force sensor was fixed by epoxy as the last step of fabrication, while it was the first step in previous chapters.

![Sketch of the temperature compensated tip-force sensing epidural needle](image)

**Figure 6.2** Sketch of the temperature compensated tip-force sensing epidural needle

One concern about this design is that the fore-and-aft sensor placement could lead to a time delay for temperature compensation between the two FPI sensors at the beginning of insertions, as mentioned in Chapter 3. This influence could be eliminated when the insertion depth is beyond the positions of the two sensors, as the time delay is very short, and the spinal insertion procedure takes much longer than most needle procedures the needle is kept at a constant, slow advance rate by the operator during a real spinal insertion.

### 6.2.2.2 FPI tip-force sensing system architecture

The overall sensing system is shown in Figure 6.3, including a laser source (b), optical circuit (c) and the sensing needle (a). It is designed to be a portable system, easily to be attached to other medical system.
6.2.3  Tip-force sensing epidural needle characterisation

The temperature compensated tip force sensing needle needs to be calibrated in terms of temperature compensation and the relationship between force and FPI light intensity phase, before tissue insertion experiments. To calculate the relationship of between the two FPI sensors in regard to temperature induced signals, the tip-force sensing needle was repeatedly put inside and withdrawn from a chamber which had a temperature of 37°C. The temperature signals of both FPI sensors is shown in Figure 6.4, where (a) is the temperature-induced light intensity change, and (b) is the accumulated FPI intensity phase change. From the sensors’ response to varying temperatures, a linear relationship between the two FPI sensors was obtained, as shown in Figure 6.5.
Figure 6.4  FPI sensor signal and signal phase change information, (a) FPI sensor intensity signal induced by temperature change, (b) Accumulated FPI sensor signal phase change

Figure 6.5  Phase change relationship fit result between FPI sensors
As a linear relationship could be found between insertion force and optical intensity phase of the FPI signal as mentioned in Chapter 3, sensor calibration can be obtained via curve fitting the two signals. A force signal is gained from a commercial ATI force sensor (ATI SI-20-1, USA) attached at the base of the needle, with a resolution of 0.02 N and a force sensing range specified as 0 - 60 N. A linear relationship between the two signals was then obtained via applied tip forces, as shown in Figure 6.6. The sensing needle showed a sensitivity of 0.1 N during the calibration.

![Figure 6.6](image)

**Figure 6.6** Linear relationship between force and FPI intensity phase

Figure 6.7 shows a force comparison between the FPI sensor based sensing needle and the ATI force sensor after temperature compensation. It demonstrates that the sensing needle could provide reliable force signals.

![Figure 6.7](image)

**Figure 6.7** Force results of FPI sensor compared with ATI force sensor
6.3 Epidural space identification

The experimental setup and a proposed insertion path sketch for porcine spinal insertion are shown in Figure 6.8. The sensing needle was operated manually, advanced as slowly and as steadily as possible. The objective is to identify the epidural space before contact between the needle tip and the spinal cord.

The last tissue layer before entry into the epidural space is ligament flavum, which has much greater stiffness than surrounding tissue, with a penetration force of around 8 N for both in vivo and ex vivo insertions on porcine spine using an 18G needle [211]. As the thickness of human epidural space is around 4 - 7 mm in the upper thoracic region and the lumbar region [221-223] after passing the ligament flavum, the epidural space needs to be detected within 4 mm of the needle tip breaking the ligament layer.

![Figure 6.8](image)

**Figure 6.8** Experimental setup for porcine spine insertion (a) Laser source, (b) Monitoring interface, (c) Optical circuit, (d) Epidural sensing needle, (d) Recording camera

Two types of experiments were carried out for performance assessment of the tip-force sensing needle. A preliminary test was carried out to first explore the feasibility of epidural space identification by using a bevel tip force sensing needle at room temperature.
After that, another experiment of porcine epidural space identification was conducted with the novel temperature-compensated force sensing epidural needle at simulated body temperature.

6.3.1 Preliminary test by the bevel-tip sensing needle

Pig feet have ligaments and tendons which have similar physical properties to ligament flavum. Therefore, this was employed for a preliminary evaluation of ligament flavum insertion. The insertion experiment was performed using a bevel tip sensing needle. Needle insertion and a typical signal are shown Figure 6.9. The foot was partially dissected first, and its ligaments/tendons were isolated from surrounding tissue as shown in Figure 6.9 (a).

The sensing needle was then inserted into the ligaments and stopped 1 mm after tissue fracture for a short moment, then continued up to 20 mm after layer fracture. Figure 6.9
(b) is a representative force signal of this insertion procedure, showing that the tip force sensing needle clearly detect the event of passing ligament within 1 mm after layer fracture. Only a little friction force after 1 mm from the tip affects the tip force signal. This experiment demonstrated that the sensing needle has the potential to detect the epidural space at around 1 mm after the needle tip reaches the area.

An epidural space identification experiment was then carried out on porcine spinal sections, shown in Figure 6.10. Considering the needle length and the difficulty in locating entry positions without a visual system, insertion started from muscle rather than skin. It then passed through the ligament flavum, and reached epidural space. To correctly locate the ligament flavum, the operator firstly examined the sample between two spinous processes to find the correct insertion locations. The sensing needle was then inserted as slowly and as smoothly as possible into the spine, at an average advance rate of around 3 mm/s. The needle passed through the ligament flavum layer, epidural space, dura mater, spinal cord until it hit the vertebral body where the needle cannot enter. A representative insertion signal sample of this procedure can be found in Figure 6.10.

Figure 6.10 Porcine spine sample insertion and its insertion signal

A small drop of force at around 1 N indicated the first tissue layer breaking event. The needle then experienced several force fluctuations at needle displacements of 3 - 5 mm due to the unstable advance rates of manual operation. A final sharp drop of force signal clearly showed that the needle tip had passed the ligament flavum layer. After entry into
the epidural space, the tip-force maintained a stable value until gradual force increases occurred at around 4 mm after entry. Based on spinal structure, the tissue in this position was inferred to be dura mater, the entry to the spinal cord. At the last stage of insertion, the needle stopped at the vertebral body with a very large force reading.

6.3.2 Porcine epidural space identification with the epidural sensing needle

The porcine spine sample was warmed to 37 °C using laboratory water bath equipment (SWBR27 SHEL LAB, US) with a temperature precision of +/-0.2 °C, as shown in Figure 6.11. The warm spine sample was then taken to the experimental table for manual insertions. The insertion procedure was the same as in the preliminary test. A representative insertion signal for this procedure is displayed in Figure 6.12. From the tip force results of the spine insertion procedure, it can clearly be seen that the epidural space could be detected before spinal cord insertion. There is enough space (about 3.8 mm) and time (around 1.3 s at 3 mm/s) for clinicians to recognise the target. The experiment also indicated that the special tip of the epidural needle is effective for protecting the spinal cord from insertions.

Figure 6.11 Spine sample preparation and experimental setup, (a) Laboratory water bath, (b) Porcine spine segment, and (c) Needle insertion
During insertions, it was found that maintaining a stable needle advance rate is very difficult, especially during fracture of the ligament flavum. A large proportion of the samples had sudden penetration directly from the epidural space to the vertebral body. The operator still needs lots of practice in driving the needle stable and slowly. That being said the epidural needle does have the ability to provide reliable tip force feedback that could identify the epidural space during stable insertion.

6.4 Conclusion

This chapter presents two tip force sensing needles with an optical sensor mounted at the precise tip. The first one is based on a bevel tip needle, designed for preliminary experiments to investigate the feasibility of epidural space identification, the other is based on an epidural needle, with temperature compensation for simulated testing as per daily surgery. The sensing needles were fabricated and characterised based on realistic requirements. Results showed the sensing needle has a force sensing range of 0 - 10 \( N \), with a sensitivity of 0.1 \( N \).
To assess the feasibility of epidural space identification, insertion experiments on porcine ligaments and spinal sections were conducted. The results showed that bevel tip force sensing needle could provide reliable force information to distinguish epidural space 1 mm after entry of the epidural space. The spine sample insertion experiment was also conducted using the epidural tip force sensing needle. Its results also indicate that it can identify epidural space, although it requires better control to avoid sudden large depth penetration during insertion.
Chapter 7

Conclusion and future work

This thesis has presented novel temperature-compensated tip force sensing needles and their tissue identification applications. The sensing needle is based on the FPI sensing principle, with a temperature compensation design through the addition of another FPI sensor as a reference sensor that is only sensitive to temperature variation. A series of experiments were designed and carried out to test real-time tissue identification during needle insertion. This chapter reemphasises the overall outcomes and contributions of this research, followed by a discussion of the research’s perspectives and its direction for future work.

7.1 Research outcomes

The overall outcomes of this research are: (1) the design, fabrication, and characterisation of a tip force sensing needle, (2) an optical circuit for FPI sensors, (3) signal processing
of temperature compensation based on an intensity-phase algorithm, and (4) a tissue identification concept and experimental validation on different tissues.

7.1.1 Sensor design, fabrication, and calibration

Temperature interference in FPI force sensing was verified via theoretical calculation, simulation, and temperature experiments. To resolve the problem, a novel structure for FPI signal temperature compensation was proposed, based on providing another reference FPI sensor which is only sensitive to temperature change. The fabrication procedure was designed to be effective, economical and the FPI sensor size was miniaturized as much as possible so it was suitable for embedding in small diameter needles. The experiments with varying temperature and force demonstrated this sensing needle design could detect both temperature and force signals.

The sensing needle was calibrated via commercial force sensing equipment. The results show that the sensing needle successfully overcomes the influence of temperature and captured the amount of force applied to its tip.

In addition, as the reference FPI sensor could provide temperature information, the sensing needle was also calibrated in terms of temperature sensing.

7.1.2 Signal processing of non-linear FPI interference signals

A multiple FPI sensing circuit was built for the two FPI sensors embedded inside the needles. The whole circuit was optimised from a large circuit board with a high powered supplied laser in a control box, with connection terminals to the FPI sensor, a portable laser source, and a computer. The entire system was able to be small, portable, economical and practical for MIS.

As the system’s signal processing is based on interference intensity modulation, it needs to deal with non-linear optical signals. FPI interference light intensity has a cosine relationship with applied force. Therefore, an algorithm was proposed to turn light intensity to its intensity phase signal which is linear with applied force variation.

It was found that the temperature-induced signals of the two FPI sensors have a linear relationship with each other during temperature experiment, on which the temperature
compensation algorithm is based. The hybrid temperature-force experiment results showed that temperature compensation is effective.

7.1.3 **Tissue identification by tip force sensing needle**

Tissue identification is the main purpose of this research. To achieve this objective, a series of tissue experiments were conducted, including phantom tissue, and porcine tissue. A database of tip force versus tissue type was then established for tissue identification. Factors influencing tip force during needle insertion were also investigated.

Based on above results, tissue identification experiments were conducted, the results of which indicate the possibility of tissue identification in specific body locations.

In vivo experiments on mice were designed and conducted to verify the concept of temperature compensation and tissue identification. The results confirmed the ability of the tip force sensing needle, though some of the internal organs of the mice were too soft to be detected.

7.1.4 **Other applications**

An improved tip force sensing needle was fabricated for the application to epidural space identification. Two types of needle, a bevel tip needle and an epidural needle, were considered as candidates. The experiment by using porcine spinal samples showed that the improved sensing needle could detect the epidural space effectively.

7.2 **Contribution to the scientific development of tip force sensing techniques**

- **Sensor design and calibration**
  - A new structure for resolving temperature compensation,
  - A new algorithm to process non-linear FPI interference intensity signals,
  - A low-cost portable optical sensing system design, and
  - A GUI for displaying tip force and temperature readings
- **Tissue identification experiments**
  - Tissue identification concept via tip force sensing,
  - Investigation of factors influencing tissue identification,
  - Exploration of tissue identification, including phantom tissue and porcine tissue,
  - In vivo experiment validation of tissue identification, and
  - Modified tip force sensing design for epidural space identification and experiments

### 7.3 Limitations and future work

This research has successfully demonstrated that the tip force sensing needle has the ability to overcome the influence of temperature and detect tip force during needle insertion, further to achieving tissue identification. After the concept is validated, the next step would be the development of the sensor’s accuracy.

In line with this objective, more research needs to be undertaken to further improve the system developed thus far for needle application, including:

- The algorithm of non-linear signal processing needs to be modified to achieve better sensor resolution. Sensor resolution is related to three aspects, which are laser source fluctuations, temperature compensation, and intensity-phase transference. An additional channel of light signal reading might be needed for reading the laser source changes to aid compensation. Based on a more precise laser source, temperature patterns of the two FPI sensors must be studied further.

- System miniaturization is another direction to improve the tip force sensing system. The light intensity reading sensor - the light power meter could be designed and fabricated at a much smaller size for integration. The entire system could be miniaturized to a super “pen” size, which would be very portable.

- The GUI could be more intelligent, displaying tissue type and tissue size or shape in real time during needle insertion, rather than charts or readings. Achieving this goal
would make the system more operator friendly. In addition, more functions, such as a significant organ damage alert, could be added to the GUI.

- It was found that epoxy cannot bear long term heating, and its properties might change properties over time. Therefore, it would be better to use laser fusing techniques to form the FPI cavity with the capillary.

- The current FPI signal reading is quite weak, compared with the input light. Although the light intensity reading sensor could still read the signal, it would be better to treat the cleaved end, with a coating reflective to enhance reflected light intensity.

- Compared with needles used currently in surgery, these needles could not hold two sensors inside. It may be possible to use intrinsic FPI fabrication techniques to achieve a smaller sensor size.

- When a long needle has a significant pliability, it will have a significant effect on the FPI sensor, as it pulls on the fibre making the FPI cavity length change. Therefore, integrating the sensor into a longer needle needs to overcome this issue.

FPI sensors also could be used in other environments, such as soft robotics, for the force sensing applications and shape changing applications, or other materials that need force, temperature, and shape change monitoring.
Publication List

Peer Reviewed Journal Publications


Peer Reviewed Conference Publications


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