Design of Novel Sensors and Instruments for Minimally Invasive Lung Tumour Localization via Palpation

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A thesis submitted in partial fulfillment of the requirements for the degree in Master of Engineering Science  
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DESIGN OF NOVEL SENSORS AND INSTRUMENTS FOR MINIMALLY INVASIVE LUNG TUMOUR LOCALIZATION VIA PALPATION

(Thesis format: Integrated Article)

by

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A thesis submitted in partial fulfillment of the requirements for the degree of Master of Engineering Science

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Abstract

Minimally Invasive Thoracoscopic Surgery (MITS) has become the treatment of choice for lung cancer. However, MITS prevents the surgeons from using manual palpation, thereby often making it challenging to reliably locate the tumours for resection. This thesis presents the design, analysis and validation of novel tactile sensors, a novel miniature force sensor, a robotic instrument, and a wireless hand-held instrument to address this limitation. The low-cost, disposable tactile sensors have been shown to easily detect a 5 mm tumour located 10 mm deep in soft tissue. The force sensor can measure six degrees of freedom forces and torques with temperature compensation using a single optical fiber. The robotic instrument is compatible with the da Vinci surgical robot and allows the use of tactile sensing, force sensing and ultrasound to localize the tumours. The wireless hand-held instrument allows the use of tactile sensing in procedures where a robot is not available.

Keywords: minimally invasive surgery, force feedback, haptic feedback, tactile sensor, force sensor, fiber Bragg grating, robotic palpator, hand-held palpator, tumour localization, palpation, da Vinci robot, sensorized instrument
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1D One Dimensional
2D Two Dimensional
3D Three Dimensional
ADC Analog to Digital Converter
CDC Capacitance to Digital Converter
CT Computed Tomography
DOF Degree Of Freedom
EBRT External Beam Radiation Therapy
EDM Electric Discharge Machining
FEA Finite Element Analysis
FOS Factor Of Safety
ICS Inter-Costal Space
LED Light Emitting Diode
MEMS Micro-Electro-Mechanical Systems
MIS Minimally Invasive Surgery
MITS Minimally Invasive Thoracoscopic Surgery
MRI Magnetic Resonance Imaging
NPT National Pipe Tapered Thread
NSCLC Non-Small Cell Lung Cancer
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<td>PCB</td>
<td>Printed Circuit Board</td>
</tr>
<tr>
<td>PET</td>
<td>Positron Emission Tomography</td>
</tr>
<tr>
<td>PID</td>
<td>Proportional Integral Derivative (controller)</td>
</tr>
<tr>
<td>PPS</td>
<td>Pressure Profile Systems</td>
</tr>
<tr>
<td>PVDF</td>
<td>Polyvinylidene Fluoride</td>
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<td>VATS</td>
<td>Video-Assisted Thoracoscopic Surgery</td>
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Chapter 1

Introduction

1.1 Motivation

Lung cancer is the most prevalent form of cancer in both men and women [1]. It is often treated by surgical resection of the tumour, quite frequently via Minimally Invasive Thoracoscopic Surgery (MITS). MITS is significantly better than open surgery for the patients due to less trauma, pain and scarring, faster recovery, and lower risk of infection [2]. However, when compared to open surgery, it greatly increases the challenges faced by the surgeon due to the loss of direct sight of, and physical contact with, the area being operated on. One such challenge is the inability of the surgeon to use the sense of touch (tactile feedback) for manually palpating the lung to accurately locate sub-surface tumours. There are very limited solutions available to tackle this problem, and hence there is motivation to develop novel sensors and instruments that can increase the success rate in localizing sub-surface tumours and also reduce the uncertainty in the exact location of the tumours. Further information about lung cancer, its treatment options, the use of MITS, the scope of this project and an overview of the thesis is presented in this chapter.
1.2 Lung Cancer

The focus of this thesis is on the development of sensors and instruments to aid in the treatment of lung cancer. Relevant background information about lung cancer is presented in this section to put the research presented in this thesis into context.

1.2.1 Anatomy

The lungs are a pair of soft, spongy, cone-shaped organs located in the thoracic cavity on either side of the heart [3]. The basic lung anatomy is presented in Fig. 1.1. The primary function of the lungs is to allow the vital exchange of oxygen and carbon dioxide between the blood and the atmosphere. The lungs are individually encased in two layers of serous pleural membranes that can keep one lung functioning even if the other is deflated due to an injury or deliberately during a surgery. The lungs are divided into lobes, three in the right lung but only two in the left to accommodate the heart. Air reaches the lungs via a network of airways that begins as the trachea at the throat and divides into the left and right bronchus before entering the respective lungs. The bronchi divide into secondary and then tertiary bronchi before dividing into bronchioles that terminate at the alveolar sacs – the location of gas exchange [3]. This branching of the airways is critical in allowing damage to and the resection of small sections from the surface of the lung or entire lobes without significantly affecting the rest of the lung.

1.2.2 Presentation

The lungs are very delicate structures that can be easily infected or injured and become non-functional due to constant exposure to pathogens and pollutants in the atmosphere. Lung cancer typically presents itself as fast growing malignant tumour nodules located beneath the surface of the lung [5]. These tumours typically originate on the walls of the
A cancer is characterized by the uncontrolled growth and spread of abnormal cells. About 80% of lung cancers are classified as Non-Small Cell Lung Cancer (NSCLC) that present themselves as stiff tumours that grow and spread slower than their Small Cell counterparts, making treatments more effective [6]. The cause of lung cancer can be external, such as smoking, chemicals and radiation, or internal, such as hereditary mutations, hormone imbalance and immune system reactions. If lung cancer is not treated promptly by controlling, removing or annihilating the abnormal cells, it can metastasize to the surrounding organs and eventually cause death [5]. Current medical knowledge has no proof that lung tissue can regenerate, and hence it is important to maximize the preservation of healthy lung tissue when resecting or annihilating the tumour.

### 1.2.3 Prevalence

Cancer causes more deaths worldwide than AIDS, tuberculosis and malaria combined, and lung cancer is the most prevalent form of cancer, accounting for 20% of cancer related deaths. In 2012 worldwide, there were 1.8 million new lung cancer cases and 1.6 million
lung cancer deaths [1]. It is estimated that there will be 26,600 new lung cases and 20,900 lung cancer deaths in Canada alone in 2015 [7]. Earlier detection of cancer significantly increases the chances of survival due to the smaller size of the tumours, which can be removed more easily, and the lack of metastasis of the cancer to other organs [8]. The high prevalence of lung cancer justifies the efforts to develop new tools for its treatment.

1.2.4 Diagnosis

Lung cancer is typically diagnosed by imaging the chest to verify the presence of cancerous nodules. The imaging techniques used typically are radiography (plane X-ray imaging) and computed tomography (CT). Magnetic Resonance Imaging (MRI) and Positron Emission Tomography (PET) are also sometimes used for the diagnostic imaging. When lung cancer is suspected, radiography is often used as the first modality to detect the presence of thoracic nodules due to its lower cost and radiation exposure. However, radiographs may miss nodules smaller than 2 cm in diameter, and hence CT is used directly for people with a higher probability of having lung cancer [5]. For the final confirmation images, CT has surpassed radiography as the preferred modality since it can produce three-dimensional images that present the size and location of the tumours with good accuracy.

CT and radiography both use X-rays to image the internal structures of the body, and rely on the fact that the tumours have a higher X-ray attenuation due to higher density and show up as brighter spots in the lung on the images. MRI is sometimes used since it does not utilize ionizing X-rays and provides better contrast. MRI works especially well to detect cancer that has metastasized to the brain or spinal cord [5]. However, the higher cost and lower availability of MRI limits its use. PET is a nuclear medicine imaging technique that provides three-dimensional functional information about the tumours that can be combined with the structural information from a CT scan for better localization.
1.2 Lung Cancer

(identifying the location and margins of a structure) and treatment planning [5]. It uses the fact that tumours are more metabolically active than healthy tissue and take up more of the radioactive tracer, and hence appear brighter in the images. Its benefits include the ability to provide information about the aggressiveness and metastasis of the tumours [9].

1.2.5 Treatment

Non-Small Cell Lung Cancer (NSCLC) is classified from Stage I to Stage IV in order of increasing severity and metastasis. Fortunately, NSCLC is typically diagnosed at early stages with only 20% of the NSCLC cases being diagnosed at Stage III or IV [10]. Stage IV NSCLC is typically treated using only chemotherapy and/or radiotherapy due to heavy metastasis and multiple nodules, but surgery is a feasible option for earlier stages [5].

Chemotherapy predominantly uses platinum-based drugs and targeted molecular agents to attack and kill cancerous cells. It relies on the fact that cancer cells are more metabolically active and take up more of the poisonous drugs than healthy tissue, which in turn kills them [10]. This also means that these drugs do cause systemic damage to the body as a side effect including long term cognitive impairment. Radiotherapy uses radiation from focussed external beams or surgically implanted radioactive seeds (brachytherapy) to kill the cancer cells. Radiotherapy harms surrounding healthy tissue at a lower level than chemotherapy; however, only a small percent of patients with NSCLC are suitable for radiation therapy [11].

The use of chemotherapy and radiotherapy alone without surgery in Stage IV NSCLC has very low success with only about a 1% 5-year survival rate. On the other hand, the use of these treatments in a combined modality therapy with surgery (pre-operatively and/or post-operatively) has shown up to a 15% increase in the 5-year survival rate in lower stages of NSCLC, indicating that surgery plays a major role in the treatment [12].
1.2 Lung Cancer

1.2.6 Traditional Open Surgery

Surgery is the preferred treatment for early stages of NSCLC with no or very little metastasis [13]. Different procedures can be performed depending on the size and location of the tumours as well as the patient’s health and functional lung capacity. The three typical procedures are segmentectomy (removal of a small cancerous segment of the lung, also known as wedge resection), lobectomy (removal of an entire lobe), and pneumonectomy (removal of an entire lung) [5]. A review of several different studies has shown that when surgery is performed at earlier stages, the 5-year survival rates range from 40% to 85% depending on the size and aggressiveness of the tumour, and the type of surgery [9].

Traditionally, surgical resection of lung tumours was performed through an open thoracotomy. In this procedure, the patient is positioned on his/her side and an incision is made on the patient’s side between the ribs where the tumour is located. The incision extends from between the scapula and the spine on the posterior side for about 15 cm to 25 cm to the anterior side [14]. The ribs are then spread apart using a rib spreader to gain direct access to the lung, which is a major cause of lengthy post-operative pain. The lung being operated on is collapsed prior to making the incision using an endotracheal tube that blocks airflow to that lung and allows the lung to deflate naturally in 20–30 minutes. Once deflated, an adult lung is roughly 16 cm × 9 cm × 4 cm in size and is extremely soft and delicate [15]. This deflation process limits the use of pre-operative imaging in locating the tumour during surgery since the size and location of the lung changes dramatically. The surgeon localizes the tumours by using visual cues and by manually palpating the deflated lung with gloved hands trying to feel for a spheroid lump in the tissue, because tumours typically have a higher stiffness than the surrounding healthy tissue. This approach to locate tumours works very reliably for tumours greater than 1 cm in diameter, which is typically the case. Once localized, the tumour(s) are excised and the incisions in the lung are stapled shut. Finally
the chest cavity and surrounding tissue is closed with sutures with one or more tubes coming out of the chest to allow for air and fluid drainage for about 3 to 5 days [15].

1.3 Minimally Invasive Thoracoscopic Surgery

The specific lung cancer treatment approach that this thesis focuses on improving is Minimally Invasive Thoracoscopic Surgery (MITS). Relevant background information about MITS, including the use of surgical robots and the limitations that this research is trying to overcome is presented in this section.

![Comparison between the incisions in open and minimally invasive surgery: (a) traditional open thoracotomy, (b) minimally invasive thoracoscopic surgery.](image)

1.3.1 Overview

With the advancement of technology in the field of healthcare, a revolutionary technique known as Minimally Invasive Surgery (MIS) evolved in the 1980’s that has become very popular as an alternative to open surgery [16]. When performed on thoracic organs instead
of open thoracotomy, it is known as MITS. MITS is also known as Video-Assisted Thoracic Surgery (VATS) due to the use of the video feed from an endoscopic video camera called a thoracoscope, rather than using direct sight to perform the procedure. In MITS, three, or sometimes four, small 1–2 cm incisions (called ports) are made in the side of the chest after deflating the lung rather than a single long 15–25 cm incision, as can be seen in Fig. 1.2 [15]. The thoracoscope is typically inserted through an incision in the 6th–8th intercostal space (ICS, between the ribs), and other two 5–12 mm diameter laparoscopic (minimally invasive) instruments, such as graspers, scissors, retractors, staplers, electrocautery probes, etc., are inserted through incisions in the 3rd–4th ICS at the midclavicular line and the 5th ICS near the border of the scapula [15]. The incisions are made using a device with a sharp tip known as a trocar. The inner shaft with the sharp tip is removed after pushing the trocar through the chest wall to leave behind a tube that allows for the insertion of minimally invasive instruments without constantly rubbing against the chest wall tissue, thereby reducing friction and protecting the chest wall from further damage.

With some guidance from preoperative images, and possibly intraoperative ultrasound images, the surgeon uses visual feedback from the thoracoscope and limited kinaesthetic force feedback from a laparoscopic grasper to localize the tumours to be excised. The surgeon then uses a stapling instrument to excise the tumour along with a 1–2 cm margin of healthy tissue to account for localization uncertainty, while simultaneously closing the incision in the lung [15]. The excised tissue is sealed in a retrieval bag before it is pulled out through one of the ports to reduce the chances of contamination with tumour cells. Due to the limitations of this process, the tumours typically excised using MITS are no more than 3 cm in diameter and are located towards the periphery of the lung [17]. This means that the lung cancer must be detected at an early stage for the patient to qualify for MITS. Fortunately, with better imaging and screening techniques, about 60% of lung cancer cases today are detected early enough to qualify for MITS [17].
1.3 Minimally Invasive Thoracoscopic Surgery

1.3.2 Benefits and Challenges

The significantly smaller incisions in MITS provide several benefits to patients. Post-operative pain is significantly reduced due to less trauma to the chest wall and the elimination of the use of a rib spreader [18]. The smaller incisions heal faster, thereby decreasing the length of hospital stay, and they also result in less scarring which is preferred by patients for cosmetic reasons. On average, MITS patients return to normal activity within 10 weeks versus 16 weeks for open thoracotomy patients [16]. MITS has also been shown to preserve better lung function due to lower collateral damage to the chest wall. Another significant advantage of MITS is that it can be performed on weaker and older patients who have very low chances of recovering from an open thoracotomy, and hence it increases the treatment options available for these patients [18].

Although there are several benefits to performing MITS for patients, it also comes with several challenges for the surgeons. The most significant challenge is the inability of the surgeon to use manual palpation to localize the tumour, the quick and reliable technique that surgeons depend on during open thoracotomy [19]. The surgeon can barely insert a single finger through an MITS port which is insufficient for proper palpation. The use of two-dimensional video from a thoracoscope instead of direct sight creates a loss of depth perception for the surgeon [16]. Since the surgeon typically handles two laparoscopic instruments, an assistant manoeuvres the thoracoscope which limits the speed at which the field of view can be adjusted. Another big challenge is the limited and reversed motion of the instruments due to insertion through a fixed point on the chest wall, referred to as the remote centre of motion (RCM) [16]. The RCM makes the instruments function like a Class 1 lever with the fulcrum at the RCM, and this reverses the tip motion relative to the handle. It also scales the lateral motion and forces on either side of the fulcrum creating a distorted perception for the surgeon. These effects greatly increase the complexity for surgeons and
make MITS considerably more difficult to learn and perform than open thoracotomy.

1.3.3 **Intraoperative Tumour Localization**

The problem of intraoperative tumour localization in MITS has been investigated for three decades now and several solutions have been proposed and tested clinically, some of which have become the standard of care today. Proposed solutions include interpolating from preoperative CT scans, inserting markers using CT guidance before surgery, using laparoscopic ultrasound probes, and surgeons feeling for the tumours using a laparoscopic grasper and visual cues from the thoracoscope. There are also several instruments and techniques that have been recently developed by researchers to aid the surgeon in tumour localization by somewhat restoring the surgeon’s sense of touch. However, these new technologies have not yet reached widespread clinical evaluation, and are discussed in more detail in the next chapter.

**Preoperative CT Scans:** As discussed earlier, CT is the primary imaging modality used for the diagnosis of lung cancer. CT scans provide very accurate information about the size and location of the tumours in the inflated lung. But this information becomes inaccurate for direct intraoperative use since the lung is deflated during the surgery [20]. The change in the shape and position of the lung during deflation is quite non-linear, however, experienced surgeons can still use the preoperative information to interpolate mentally and approximately locate the tumour down to the correct lobe. In case of a lobectomy, this may be sufficient, but for a segmentectomy in which the preservation of healthy tissue must be maximized, better tumour localization is required. Ideally, CT scanning must be performed intraoperatively, but the high cost and complexity limits the availability of CT scanners in the operating room.

**CT Guided Markers:** Another solution that has been attempted clinically is the insertion
of a marker under CT guidance before the surgery to help locate the tumour. The marker is placed using a percutaneous needle that is inserted close to the tumour in the lung through the chest wall. Two common markers used in the past are a one foot long guide-wire with a small hook on the inserted end [21], and 0.5 ml of injected methylene blue contrast dye [22]. The challenges with marker placement are the risk of pneumothorax, hemothorax and/or pulmonary haemorrhaging due to the needle, and the availability of CT scanner in a facility where percutaneous needle insertion can be performed [23]. Even after successful placement of the marker, the guide-wire can dislodge while transporting the patient, and the dye can diffuse to an extent that it becomes unusable. A study by Santambrogio et al. has shown that the failure rate in locating tumours using methylene blue and guide-wires were 13% and 47%, respectively, resulting in most clinical centres avoiding the use of markers today [23].

**Intraoperative Ultrasound:** Using laparoscopic ultrasound probes for intra-thorascopic ultrasound imaging is a common technique for intraoperative lung tumour localization. Diagnostic ultrasound imaging is a non-invasive technique that uses the reflection of 1–10 MHz ultrasonic waves from tissue boundaries to create an image of the internal organ structure [24]. Ultrasound machines today are relatively inexpensive, portable and create real-time images permitting their use intraoperatively. A laparoscopic ultrasound probe is introduced in the thoracic cavity through one of the three ports after the thoracic cavity is filled with isotonic saline solution [25]. The probe is then placed directly on the lung surface and the area with the suspicion of having a tumour is scanned until the tumour is found. The main challenge with using ultrasound here is that the residual air in the alveoli and bronchioles of the deflated lung causes significant artefacts in ultrasound images since air–tissue interfaces reflect sound significantly [26]. The poor quality and distorted images obtained limits the use of ultrasound in this application. Furthermore, it is very difficult to identify tumours smaller than 1 cm in diameter using ultrasound alone.
Laparoscopic Grasper: The simplest and most widely used approach is the use of a laparoscopic grasper with its jaws closed to push on the lung tissue directly to palpate it. The surgeon uses his/her feel of the force exerted on the instrument handle and the visual feedback about tissue deformation, texture and other characteristics from the thoracoscope to approximate the tissue stiffness and locate the tumour [16]. This, however, is a skill that is very difficult for the surgeons to learn since the palpation has to be done with instruments that are difficult to use due to motion reversal, distorted kinaesthetic force feedback due to friction and deformation of the tissue and the ribs at the trocar, and complete lack of tactile feedback [27]. There is a risk that the surgeon can permanently damage the delicate lung tissue if large palpation forces are accidentally applied due to distorted force feedback. Experiments have shown that excessive palpation forces are often applied when no measured force feedback is available [27]. The long length of the instruments also worsens ergonomics and amplifies hand tremors, resulting in excessive surgeon fatigue in comparison to open surgery [16]. Tumours can also be incorrectly detected because it is difficult to differentiate a tumour from any other hard underlying structures such as airways and blood vessels due to a lack of tactile shape perception [16]. Tumours larger than 2 cm exhibit a distinct lump that can be easily identified via visual feedback alone, and verified with a grasper, but smaller subsurface tumours pose a bigger localization challenge. These problems motivate the development of a better solution for this problem.

1.3.4 Advent of Surgical Robots

By the early 1990’s, technology had become sufficiently advanced to enable the introduction of robots to aid surgeons in performing surgeries with better precision and reliability. The RoboDoc, an orthopaedic surgery robot introduced in 1992, and the AESOP, a voice-command based endoscope manipulating robot introduced in 1994 were the first
surgical assistance robots used clinically. The da Vinci Surgical Robot, introduced in 1997
by Intuitive Surgical Inc., was the first complete robotic minimally invasive surgery system
to be commercialized [28]. Today, it is the only minimally invasive soft-tissue surgery sys-
tem with the clearance to operate on humans in North America. In 2014, approximately
570,000 surgical procedures were performed with the da Vinci system worldwide.

The da Vinci system overcomes several limitations of MITS by using computer en-
hanced instrumentation. The system consists of a master console from which a comfort-
ably seated surgeon controls the motion of up to four surgical arms using a pair of con-
trollers [29]. The master console provides the surgeon with a three-dimensional stereo-
scopic view of the surgical site enhancing depth perception. The surgical arms originate
from a mobile surgical arm cart that is positioned adjacent to the patient. Two of the arms
hold special laparoscopic instruments with an actuated wrist. These instruments mirror the
motion of the surgeon’s hand in a manner that feels natural to the surgeon looking at them
through the stereoscopic view, thereby avoiding the problems associated with tip motion
reversal [29]. A third arm holds the stereo-thoracoscope and is also controlled by the sur-
geon. An optional fourth arm holds a third instrument if required for the procedure. Since
only two arms can be controlled at a time, the surgeon switches the arms under control by
using foot pedals at the master console. The da Vinci system provides no automation, i.e.,
the surgeon is completely in control. However, the system provides motion enhancements
such as tremor cancellation, motion scaling, better dexterity due to the actuated wrist on
the instruments, greater accuracy and intuitive hand-eye coordination. This arrangement
makes performing MITS much easier for the surgeon and safer for the patients.

The da Vinci system also comes with several limitations apart from the high cost of the
system itself. The main limitation, which is also shared by other commercially available
robotic systems, is the absence of kinaesthetic and tactile feedback. The level of feedback
is even lower than manual MITS since the surgeon does not even physically hold the instru-
1.4 Feedback for the Surgeons

The forms of feedback that can be made available to the surgeon in order to prevent unnecessary tissue damage, to decrease the duration of MITS, and to help improve the accuracy and reliability of localizing a lung tumour are described in this section.

1.4.1 Kinaesthetic

Kinaesthetic feedback, or force feedback, is information about the overall magnitude and direction of the force being applied by the instrument on the tissue, but not about local variations in tissue stiffness and texture. The inclusion of kinaesthetic feedback makes the surgeon aware of the true force being applied on the tissue [30]. This information helps the surgeon to avoid permanent tissue damage. Kinaesthetic feedback can be provided to the surgeon via a visual display of the applied force, and/or via haptic feedback from the controllers that the surgeon uses to manipulate the instrument in the da Vinci system. Two common techniques of obtaining kinaesthetic information are incorporating some form of direct or indirect force/torque/strain sensing in the instrument, and measuring the current consumption in the motors of the surgical robot driving the instrument [31]. The latter is a much less accurate technique due to friction and non-linearities in a surgical instrument, but provides some useful information if the former is not possible. Section 2.2 describes the current and emerging force sensing technologies used in minimally invasive instruments.
1.4.2 Tactile

Tactile feedback, or touch feedback, is the information about the local variations in tissue stiffness and texture that one would feel by using one’s finger to palpate tissue [32]. Tactile information can be obtained by creating a map of the pressure distribution over the contact area when a sensing surface is pressed flat against the tissue. This pressure distribution directly correlates with the local variations in tissue stiffness since stiffer underlying structures will cause a greater contact pressure than the surrounding softer tissue [33]. The information about texture, although sometimes useful, is not crucial for tumour localization, and hence is typically ignored due to the difficulty of sensing such information. The contact pressure distribution is typically measured using a tactile sensor array, which is essentially a grid of discrete contact pressure sensing elements, making it more complex when compared to kinaesthetic sensors [33]. Tactile sensors typically do not need to be calibrated to obtain only tactile information because all that is required is a relative pressure map to identify variations in tissue stiffness. However, proper calibration to measure the true pressure enables tactile sensors to provide kinaesthetic feedback as well. Proper incorporation of tactile sensing in minimally invasive instruments and techniques for relaying tactile information to surgeons is an area of active research [33]. Section 2.1 describes the current and emerging tactile sensing technologies used in minimally invasive instruments.

1.4.3 Ultrasound

As mentioned earlier, intraoperative ultrasound is a popular modality for tumour localization. Even though the images obtained are of poor quality and require the assistance of a radiologist for proper interpretation, ultrasound provides valuable structural information that can be used for tumour localization. Furthermore, there is ongoing research to use image processing techniques to extract more useful information from ultrasound images.
and reduce the dependence on a radiologist. These reasons justify the inclusion of ultrasound as a desired form of feedback for tumour localization. Combining ultrasound with tactile feedback can increase the localization accuracy, and serve as a secondary means of confirming the presence of a tumour to increase reliability.

1.5 Project Objectives

This section describes the purpose and the scope of this thesis, and the extent of work that has been performed so far. It also briefly mentions the contributions that this work has made to the active field of improving minimally invasive lung tumour localization.

1.5.1 Purpose and Scope

The purpose of this project was the development of novel sensors and instruments that build on existing technologies to potentially provide better lung tumour localization in MITS. The basis of the work is the existing knowledge that by using a combination of the three forms of feedback described in the previous section, and by relaying the information to the surgeon through haptic and visual pathways, the benefits of the surgeon’s natural perceptions in an open surgery can be realized to a great extent in MITS.

To accomplish this goal, the development of low-cost disposable tactile sensors, a miniature force/torque sensor, a robotic palpation instrument and a wireless hand-held palpation instrument was undertaken. The two sensors are designed to be compatible with the robotic instrument, but they can also be employed in other instruments. The robotic instrument allows the use of both tactile and ultrasound data to localize the tumour, and the force/torque sensor data to control the force applied during palpation. The hand-held instrument is intended to allow the use of the tactile sensor when robotic assistance is not being used. The ultimate objective is to develop instruments that enable surgeons to
quickly, easily and reliably locate lung tumours intraoperatively, with or without robotic assistance, by fusing the data from multiple sensors. This data fusion is expected to decrease the occurrence of false positives and false negatives, and improve localization accuracy.

This thesis mainly focuses on the electrical and mechanical design and analysis of the sensors and the instruments, along with the construction of early version prototypes to demonstrate the functionality of the designs. Very limited visualization software development and structured experiments have been performed thus far, which will be a part of the future work necessary to eventually take these new devices to clinical trials.

1.5.2 Contributions

The work presented in this thesis makes four major contributions to the field of minimally invasive lung tumour localization:

1. **Novel Disposable Tactile Sensors:** The novel tactile sensors presented in this thesis are unique because of their simple construction, low-cost, disposability, on-board signal processing and 4-wire digital interface. Even though there are several existing tactile sensor designs, none of them feature these desirable characteristics. These tactile sensors use proven existing piezoresistive and capacitive sensing technologies, but implement them in a novel physical design. The design allows the tactile sensors to be easily installed in and removed from the palpation instruments, enabling the instruments to be autoclaved.

2. **A Novel Optical 6-DOF Force/Torque Sensor:** The novel Fibre Bragg Grating (FBG) based optical force/torque sensor presented in this thesis features a robust construction that enables 6-DOF force/torque measurements with temperature compensation in a very small form-factor with acceptable accuracy. The sensor uses a single optical fibre to make all of the measurements, thereby avoiding sensitive electronics
and making the sensor autoclavable. The sensor has a large force and torque measurement range implemented in a unique easy-to-manufacture design.

3. **A Novel Robotic Palpation Instrument:**  The robotic palpation instrument presented in this thesis is the first da Vinci robot compatible palpation instrument that combines three different sensing modalities to improve tumour localization accuracy. The instrument features three degrees of freedom to allow the end effector to follow the surgeon’s natural hand motion, making it very easy to use on the da Vinci robot. The instrument features a one-sided palpation design with force feedback to minimize the risk of tissue damage. The instrument has a tactile sensor and an ultrasound transducer back-to-back that allows the surgeon to quickly switch between the two modalities and confirm the presence of a tumour with minimal hassle.

4. **A Novel Wireless Hand-Held Palpation Instrument:**  The wireless hand-held instrument presented in this thesis enables the use of a tactile sensor to palpate tissue when robotic assistance is not being used in surgery. The wireless design improves the ergonomics and dexterity of the instrument. The instrument features a very simple and robust construction. It has a unique end effector design with a single passive internal degree of freedom that allows the tactile sensor to self-orient along the tissue surface, simplifying the use of the instrument. The instrument also uses an innovative design to allow for the easy removal of the sensitive electronics to enable autoclaving.

### 1.6 Thesis Organization

A brief overview of the seven chapters in this thesis is presented in this section. The first two chapters lay the groundwork for the thesis, the next four chapters present the four major components of this project, and the last chapter summarizes and concludes the thesis.
1.6 Thesis Organization

1.6.1 Chapter 1 — Introduction

This chapter presents the motivation behind this work and the necessary background information to understand how the work presented in this thesis can improve lung tumour localization. This chapter also outlines the scope of the project and provides an overview of the entire thesis.

1.6.2 Chapter 2 — Literature Review

This chapter presents a succinct literature review of the work that has been done by researchers in the past related to the work that is being presented in this thesis. It reviews existing and emerging technologies and designs in the field of tactile sensors, miniature force sensors and minimally invasive lung tumour localization instruments. It discusses the advantages and limitations of the work that has been done in the past to illustrate how this research builds upon it. Existing mechanical designs of both hand-held and robotic instruments are also described in this chapter. The different techniques used to provide feedback to surgeons, and the use of multiple modalities in a single instrument to improve tumour localization is also discussed.

1.6.3 Chapter 3 — Tactile Sensors

This chapter presents all of the work done in this project related to the design, analysis and testing of novel low-cost disposable tactile sensors. Four tactile sensor designs are presented – two different sizes (8 mm wide and 10 mm wide) with a piezoresistive sensing and a capacitive sensing based version in each size. The two different sizes are intended to allow for a choice based on patient size, preferred trocar size, tumour size, etc. The two different sensing technologies are intended to allow for a choice based on the average stiffness of the healthy tissue around the tumour. The piezoresistive version has been shown
to work better with stiffer tissue while the capacitive version works better with softer tissue. Due to this freedom in choice, the use of these tactile sensors can also be expanded to localizing tumours in the liver, the kidneys and other organs that have different healthy tissue stiffness when compared to the lungs.

This chapter begins by discussing the design requirements for these tactile sensors, followed by a detailed description of the electrical and mechanical design of the sensors. The manufacturing process of the sensors is also described in detail. Mathematical models are presented to understand the relationship between the sensor readings and the actual force applied on individual sensing elements. The visualization software developed to provide tactile information to the surgeon in an intuitive visual manner is also presented. Finally, the experiments conducted to evaluate the accuracy of the mathematical models and the performance of the sensors is described and the results obtained are discussed.

1.6.4 Chapter 4 — Miniature Force Sensor

This chapter presents all of the work done in this project related to the design and analysis of a novel miniature force sensor. The force sensor utilizes a single optical fibre with 12 Fibre Bragg Gratings (FBG) that is routed around the sensor geometry to enable six degrees of freedom force and torque measurement with temperature compensation. The sensor is designed to be a cylinder that is only 10 mm in diameter and 4 mm tall with the option of a 7 mm diameter axial hole through the centre if required to allow the passage of cables, etc. depending on the instrument it is used in.

This chapter begins by discussing the design requirements for this force sensor, followed by a brief description of the Fibre Bragg Grating sensing technology used in the sensor, highlighting why it was chosen. A detailed description of the mechanical design of the sensor is also presented along with the intended manufacturing process. A mathemati-
1.6 Thesis Organization

cal model is presented to relate the FBG readings to the forces and torques applied on the sensor. The model is also used to compute the theoretical post-calibration accuracy of the sensor based on the known accuracy of the FBG readings. Finally, the partially-complete prototype of the sensor is presented along with a discussion of the difficulties encountered that prevented the timely completion of prototype.

1.6.5 Chapter 5 — Robotic Instrument

This chapter presents the design, analysis and testing of a robotic palpation instrument that is compatible with the da Vinci surgical system. The instrument incorporates a tactile sensor, a force sensor and an ultrasound transducer to provide all of the three forms of feedback useful for tumour localization. The instrument has a wrist with three actuated degrees of freedom that provides the full range of motion for comfortably palpating tissue in any orientation. The tactile sensor and the ultrasound transducer are positioned back-to-back and the sensor in use can be switched by simply rolling the tool, a motion that can be implemented to occur on simply the press of a foot pedal at the master console. The information from the force sensor can be used to provide feedback to the surgeons as well as to partially automate the palpation procedure.

This chapter begins by discussing the design requirements for this instrument, followed by a detailed description of the mechanical design that is validated by appropriate finite element analysis. The kinematics of the instrument and control system parameters used in the da Vinci Reasearch Kit (dVRK) system to drive the instrument is also presented. The software developed for overlaying the tactile and ultrasound information on the stereoscopic video feed is also described. Finally, the experiments conducted to demonstrate the functionality of the instrument are described and the results obtained are discussed.
1.6.6 Chapter 6 — Wireless Hand-Held Instrument

This chapter presents the design, analysis and testing of a wireless hand-held palpation instrument for use in cases where robot assisted surgery is not used. The instrument uses only a tactile sensor to provide both tactile and kinaesthetic feedback. The incorporation of an ultrasound transducer was avoided to simplify the design and due to the fact that the instrument can be swapped for a laparoscopic ultrasound probe very quickly if required. The instrument has a single passive joint at the end effector that allows the tactile sensor to self align with the tissue surface. This joint can be locked during the insertion and removal of the instrument.

This chapter begins by discussing the design requirements for this instrument, followed by a detailed description of the mechanical design that is validated by appropriate finite element analysis. The design of the embedded and external electronics is also presented. The visualization software developed to provide the tactile and kinaesthetic feedback to the surgeons is also discussed. Finally, the experiments conducted to demonstrate the functionality of the instrument is described and the results obtained are discussed.

1.6.7 Chapter 7 — Conclusions

This chapter briefly summarizes all the work presented in this thesis and offers recommendations for the future work necessary to eventually take the new sensors and instruments described in this thesis to clinical evaluation.

References


Chapter 2

Literature Review

2.1 Tactile Sensors

The use of a tactile sensor array has been explored in the past to enable virtual tactile feedback for palpation during MIS [1]. A tactile sensor array (or simply, tactile sensor) is a cluster of small discrete contact pressure sensing elements that can be used to detect variations in underlying stiffness when pressed against a surface. Since tumours have a different stiffness when compared to the surrounding soft tissue, a tactile sensor can be used to locate them [2]. There are several research groups and commercial organizations that have developed different tactile sensors for MIS, however they have some major drawbacks, including lack of sterilizability, using a large number of wires (preventing their use in robotic MIS instruments with a wrist), and high cost due to expensive and highly specialized manufacturing processes [3]. Hence, developing a low-cost, re-sterilizable/disposable tactile sensor with a limited number of wires is paramount in enabling the clinical use of this technology. A thorough review of existing tactile sensor designs was conducted to determine the state-of-the-art. The three major design choices that can be used to classify tactile sensors are the sensing technology employed, the configuration used and the fabrication process.
2.1 Tactile Sensors

2.1.1 Sensing Technologies

Several different pressure sensing technologies have been explored by researchers to construct tactile sensors. The key requirement for a pressure sensing technology to be feasible for use in tactile sensors is the ability to make the individual sensing elements small enough to provide reasonable spatial resolution. It is generally accepted that a spatial resolution of $3 \times 3$ mm is a recommended minimum for obtaining any useful tactile information [4]. Depending on the application, the spatial resolution can be as fine as $1 \times 1$ mm in some existing tactile sensor designs. Almost all of the existing tactile sensors use one of the following sensing technologies.

**Piezoelectric:** This approach uses the measurement of either the charge created across a piezoelectric material, or the change in the electrical impedance of the material to determine the applied pressure [5–8]. Piezoelectric materials typically used in tactile sensors are Lead Zirconate Titanate (PZT, rigid) crystals and Polyvinylidene Fluoride (PVDF, flexible) films. PVDF is typically preferred due to its flexibility and low-cost, but PZT provides much better sensitivity. The measurement of the generated charge is simpler than the measurement of the change in electrical impedance. However, the latter typically provides more accurate measurement under quasi-static loads, which is the normal mode of operation for tactile sensors since they are pressed and held against the target tissue for a few seconds. The main drawback of piezoelectric sensing is the complexity of the circuitry required to make accurate measurements.

**Piezoresistive:** This approach uses the phenomenon that some materials experience a decrease in their electrical resistance on the application of pressure [9–13]. Some piezoresistive materials are fabricated by evenly dispersing fine particles of a conductive material such as silver, nickel, carbon black or graphite in a polymer or elastomer matrix [14]. The concentration of the conductive particles in the composite is adjusted to obtain a volume
resistivity of a few hundred $\Omega$-cm. The relationship between the resistance of these composites and the applied pressure depends on the size, material and concentration of the conductive particles, and the material of the matrix. Depending on the mechanical properties of the matrix material, the measurement can have significant hysteresis and a long response time. Another piezoresistive material that is used in some MEMS piezoresistive tactile sensors is Indium Tin Oxide (ITO), a heavily-doped n-type semiconductor that is typically deposited directly on the sensor using vapour deposition or layer etching [15–18]. ITO is a stiffer material and is used in thinner layers when compared to piezoresistive polymers and elastomers. The change in resistance can be measured using a Wheatstone bridge or a simple voltage divider.

**Capacitive:** This approach uses the change in the electrical capacitance of an electro-mechanical structure in response to applied pressure [19, 20]. The basic principle is that the capacitance between two electrodes changes as the properties or dimensions of the dielectric material between the electrodes varies as a function of applied pressure. The dielectric material can be air, polymers, elastomers or even oils [21]. The capacitance typically increases with an increase in applied pressure in most existing capacitive sensor designs. The measurement of the capacitance is usually performed by measuring the time taken to charge and/or discharge the electro-mechanical capacitor through an R-C circuit with a known resistance. Some of the notable commercial tactile sensors, such as the TactArray sensor from Pressure Profile Systems (PPS) [22] and the SureTouch sensor from Medical Tactile Inc. [23], also use capacitive sensing with excellent results, suggesting that capacitive tactile sensors can be very sensitive. The drawbacks of these commercial sensors are a large number of signal wires and high cost. These sensors are marketed for aiding in breast tumour localization by external palpation, but some researchers have adapted them to be used for MIS.

**Optical:** This approach uses miniature optical structures for the measurement of the
applied pressure. Most designs use optical fibres to carry the light to and from the sensing elements so that the electronics can be placed away from the sensor [24]. The main advantage offered by optical sensors is added the safety and robustness that results from avoiding the use of electricity in the sensor. The challenges, however, are the complexity of manufacturing the optical structures and the management of the optical fibre bundles that can become bulky with a large number of sensing elements. Some optical techniques employed in tactile sensors include using fibre Bragg grating arrays for measuring pressure through strain, the measurement of bending losses in optical fibres due to applied pressure, and using structures that alter the amount of reflected light based on the applied pressure.

2.1.2 Configuration

The configuration of a tactile sensor dictates how the discrete sensing elements are linked together to form an array. The majority of tactile sensors either use independent elements or a mechanical multiplexing configuration. There are only a few designs that have employed electrical multiplexing.

Independent Elements: This is the simplest configuration in which the individual sensing elements are completely independent pressure sensing units. Most optical tactile sensors [24], and some piezoelectric [5–8], resistive [9, 10, 14–18] and capacitive sensors [19] use this configuration. This configuration has negligible crosstalk, but results in the highest number of interface lines, since each element requires one or two optical fibres or wires. The large number of interface lines does not allow these sensors to be used on instruments with an actuated wrist.

Mechanical Multiplexing: This is a widely used configuration for electrical tactile sensors in which the sensing elements share electrodes as a feature of the mechanical construction in order to greatly decrease the number of interface lines [11, 12, 20–23]. Two
popular configurations are common electrode and grid electrode. In the common electrode configuration, one of the electrodes of the sensing elements is shared by all of the elements, which is typically the ground electrode. This reduces the number of interface lines for \( N \) sensing elements to \( N + 1 \) instead of \( 2N \). In the grid electrode configuration, one of the electrodes is shared by the elements in the same row while the other is shared by elements in the same column. By selecting the correct row–column pair, each element can be individually accessed. This reduces the number of interface lines for \( M \times N \) sensing elements to \( M + N \) instead of \( 2MN \). This form of multiplexing is used by the commercial TactArray and SureTouch tactile sensors.

**Electrical Multiplexing:** This configuration employs the use of some electrical multiplexing circuitry on board the tactile sensor to significantly reduce the number of interface lines. This approach decreases the number of measurement lines to only a few by adding some digital selection lines depending on the extent of multiplexing. This novel approach is employed by very few existing designs. The most notable design that uses electrical multiplexing is the one by Schostek et al. [13] that uses an on board 32:1 analog multiplexer on the rigid LTCC (low temperature co-fired ceramic) sensor to reduce the number of output lines to only five digital and one analog for 32 sensing elements. This design sacrifices the flexibility of the sensor in exchange for a compact interface. In MIS applications, the tactile sensor is typically mounted on a rigid tool, hence the flexibility of the sensor is not an essential feature. The design also uses distinct multiplexed sensing elements instead of the grid design, thereby reducing crosstalk and simplifying the construction. The drawback of the design is that there are still six output lines for only 32 elements, with one of the lines being analog. It is not preferable to have long analog lines due to noise considerations. Also, the LTCC technology is not a common manufacturing process and is relatively expensive for small production quantities.
2.1.3 Fabrication

There are three common fabrication processes found in the literature for the manufacture of tactile sensors. The selection of the fabrication process depends on the mechanical design, and dictates the majority of the cost of the sensor. A tactile sensor may utilize one or more of the processes listed below.

**MEMS:** Micro-Electro-Mechanical Systems is a relatively new class of devices that involve electro-mechanical structures made of semiconductors, polymers, metals and ceramics at the scale of 20 µm to 1 mm. MEMS has become very popular in the manufacture of compact sensors such as accelerometers and gyroscopes that can be packaged in a single IC. Due to the compactness and versatility of MEMS, researchers have investigated its use in tactile sensing [7, 8, 13, 15–18]. MEMS devices use similar process technology as semiconductor devices, such as deposition of material layers, patterning by photo-lithography, chemical etching, co-fired ceramics, etc. The challenge of developing MEMS tactile sensors is that it requires access to very specialized equipment and significant custom process design, which makes the sensors very expensive for small scale production. However, MEMS has many benefits in large scale production such as high precision and repeatability. MEMS tactile sensors found in the literature are usually rigid.

**Layer Deposition:** Layer deposition is a chemical manufacturing process that can be employed to manufacture tactile sensors that use a layered design involving polymers and metals [6, 14, 19–21]. It is much simpler than MEMS but still requires the use of specialized equipment and custom developed processes. The layers are typically deposited using foil adhesion and etching for metals, and vapour deposition for polymers. This process is commonly used to manufacture flexible tactile sensors.

**Macro-scale:** Macro-scale manufacturing is the simplest of all manufacturing processes. It involves conventional macro-scale device manufacturing techniques such as ma-
chining, printed circuit boards, etc. Tactile sensors made using these processes are usually low in cost but have poor spatial resolution due to the restriction on the size of the features that can be made using these techniques [5, 9–12, 24]. The advantage of using macro-scale fabrication is that the processes are very well developed to make custom components and prototypes with high reliability and minimal process design. The equipment necessary for these processes is widely available as well.

2.2 Miniature Force/Torque Sensors

Several force/torque sensors, developed specifically for minimally invasive surgical applications, have been proposed in the literature. This literature review is specifically focussed on multi-axial sensors, and therefore several uni-axial force sensors are not discussed here. Multi-axial force sensors are usually designed to either measure 3D force only (three degrees of freedom) or both 3D force and 3D torque (six degrees of freedom). The designs being reviewed are also limited to macro-scale force sensors that have force and torque measurement ranges greater than \( \pm 10 \) N and \( \pm 100 \) N·mm respectively, or designs that can be scaled up to achieve this. This criterion eliminates MEMS and other micro-scale force sensors from this review. Force/torque sensors typically have two design characteristics that can be used to classify them — the mechanical structure that translates the applied forces and torques to a measurable quantity such as strain, pressure or displacement, and the sensing technology employed to measure the aforementioned quantity.

2.2.1 Sensing Technologies

The sensing technologies used in force sensors are very similar to the ones described for pressure measurement in tactile sensors. Force/torque sensors employ these sensing technologies in a different geometry to measure forces and torques applied at a point of
interest rather than measuring contact pressure.

**Piezoresistive Strain Gauges:** This approach uses the change in the electrical resistance of a sensing structure attached to a compliant section of the mechanical structure to measure the strain at that location. The sensing structure can be traditional metal foil strain gauges [25–30] or silicon semiconductor strain gauges [31–34]. If possible, silicon semiconductor strain gauges are typically preferred over metal foil strain gauges in miniature force/torque sensors because they are more compact and have much better strain sensitivity. The change in resistance is typically very small and is measured using a Wheatstone bridge. Strain gauges can be designed to measure normal or shear strain, and several strain gauges are required for multi-axis force/torque measurements. The major drawback of this technique is the relatively large size of the sensing structures, which restrict the size of the sensor.

**Capacitive:** This approach is employed in force sensing in a similar manner as in tactile sensing. The mechanical structure is designed to include electro-mechanical capacitors that change their capacitance based on the applied forces and torques [35,36]. A drawback of capacitive force sensors is that the electro-mechanical capacitors have to be sufficiently large to measure the force with good accuracy. Due to manufacturing limitations and the size of the electro-mechanical capacitors with sufficient force capacity, only three DOF force sensors are able to use capacitive sensing. Most of the force sensors that use capacitive sensing have the measurement electronics located very close to the sensor because long wires can add significant errors to capacitance measurements.

**Optical:** This is a very popular approach for MIS force/torque sensors since optics based sensors can be easily made compact, biocompatible and sterilizable. The two broad classes of optical force sensors are reflection intensity based and fibre Bragg grating based. Reflection based sensors such as [37] and [38] use a reflective surface and a pair of optical fibres (transmitting and receiving) incorporated in a mechanical structure that causes the reflecting surface to displace under applied load and change the intensity of the reflected
light. On the other hand, fibre Bragg grating based sensors measure the strain in the mechanical structure just like piezoresistive strain gauges [39–41].

2.2.2 Mechanical Structures

The mechanical structures for the force/torque sensors that are capable of taking the specified loads are typically fabricated by micro-machining and conventional machining processes. The mechanical structures for MIS force/torque sensors found in the literature can be broadly classified into 3-DOF and 6-DOF designs. This excludes some MIS instrument designs that incorporate sets of sensors at different locations to measure additional DOFs such as grasping. The designs being discussed here are only the ones that form a force/torque sensing module that can function independently of the instruments that they are incorporated in. The function of the mechanical structures is to convert the applied loads to either pressures or strains that can then be measured using one of the aforementioned sensing technologies.

Three Degrees of Freedom: The majority of the existing force sensors designed for MIS instruments are 3-DOF. Quite frequently, additional DOFs are measured using independent sensors placed at various locations, such as the grasper, the shaft and the handle, rather than using a single sensing module with more than three DOFs, due to space restrictions [42]. Three-DOF force sensors typically have three or four sensing elements placed 120° or 90° apart, respectively, and use very simple mechanical structures. Some sensors may have additional sensing elements for temperature compensation. 3-DOF force sensors usually measure the axial force and two orthogonal bending moments since the force is usually applied to the tip of the instrument at a finite distance from the sensor. Some miniature 3-DOF sensors are as simple as using three axially-oriented piezoresistive strain gauges or fibre Bragg gratings at 120° spacing around a tube-like structure, which may just
be a section of the instrument shaft, at a few millimetres from the tip [35,37,39]. Some designs opt for four sensing elements at 90° spacing in order to decouple the two orthogonal bending moments and improve accuracy [28, 30, 36, 40]. The drawback of these designs is that they typically have low axial force measurement accuracy since the structure does not deform significantly under axial load. This problem can be avoided by making the structure more deformable, i.e., by making the wall of the tube thinner or by making cuts/holes in the tube, but the sensor has to then be moved closer to the tip so that the bending moment loads do not cause failure [29, 31, 38].

**Six Degrees of Freedom:** There are only a few miniature 6-DOF force/torque sensor designs found in the literature due to the complexity of measuring 6-DOFs within the space restriction of an MIS instrument. Three types of design are commonly seen — Stewart platform, Maltese cross and square tube. Of these, the first two classes require the measurement of only normal strain, whereas the third requires the measurement of normal as well as shear strain. The Stewart platform design has the best volume-to-strength ratio, and uses only six sensing elements. A notable monolithic design is proposed in [33] where piezoresistive strain gauges are used as the sensing elements. The drawback of this design is that the manufacture of these sensors is very challenging and expensive due to the complex monolithic geometry. An innovative solution to simplify manufacturing is proposed in [41], by making the geometry in multiple pieces rather than as a single monolithic unit. This design uses fibre Bragg gratings, but it has a much lower force/torque capacity than the monolithic design due to the limited strength of the joints in the design. The Maltese cross design is a very common and robust force/torque sensor design that is also used in larger industrial force/torque sensors [25–27, 34]. At the expense of compactness, this design typically uses of multiple sensing elements for each DOF, thereby adding redundancy and increasing accuracy. The design also allows for easy manufacturing via conventional machining processes. Nevertheless, due to the relatively large size of this design, it has not
found much acceptance in the field of minimally invasive surgery. The square tube design is a novel and simple design proposed in [32]. It simply uses four custom semiconductor strain gauges on the four faces of a titanium square tube that has thin walls to allow for the maximum possible elastic deformation. The custom strain gauges are designed to measure both the normal and the shear strain on the four faces. These eight measurements are then used to compute the 6-DOF forces and torques. A drawback of this design is that it requires custom strain gauges that can be expensive to manufacture.

### 2.3 Tumour Localization Instruments

Several instruments to aid in tumour localization by providing some form of haptic feedback to the surgeon have been developed by researchers in the past. This section describes the different types of existing tumour localization instruments, and the different techniques used by these instruments to relay haptic information to the surgeon. In general, the instruments can be divided into hand-held and robot-manipulated instruments. The hand-held instruments can be either completely passive with no actuators, or have some actuators to provide better control to the surgeon.

#### 2.3.1 Hand-held Palpating Instruments

Palpating instruments interact with tissue from only one direction by poking, pressing, etc., rather than grasping the tissue. Palpating instruments can either provide only kinaesthetic feedback by using a sensor integrated somewhere along the instrument shaft to measure the interaction force [43], or provide tactile feedback as well by incorporating a tactile sensor at the end [44–46]. Some palpating instruments are designed for just axial or lateral palpation with no internal degrees of freedom [43–45], whereas some other instruments have a manually controlled articulated wrist for added dexterity [46]. Instruments
without internal degrees of freedom are simpler to use for the surgeon, but they allow for very limited orientations at which tissue can be palpated, which can become inconvenient. Axially palpating instruments typically require a longer time to palpate the tissue surface since the palpation area on the tip of the instrument is small. Besides being faster to use, laterally palpating instruments also have an added benefit that the tactile sensors on these instruments can be large enough to cover a tumour along with some healthy tissue, which allows these instruments to use the relative tissue stiffness identify a tumour. In contrast, axially palpating instruments have to measure the true tissue stiffness accurately to detect tumours, which can be challenging without robotic assistance. However, axially palpating instruments are easier to manipulate for a surgeon because motion reversal and scaling has less effect on their performance.

Some palpating instruments use unconventional approaches for characterising tissue stiffness. The instrument described in [47] uses pulsed air jets and video feedback of the tissue response to estimate the tissue stiffness. A related approach is described in [48], where an elastic attachment on the end of an endoscope is used to palpate the tissue and the video feedback is again used to estimate the tissue stiffness. A novel instrument with a 3-DOF force sensor that has a wheel as the end effector to allow the instrument to be rolled on the tissue surface and form a mechanical image of the tissue stiffness has been proposed in [49]. The drawback of these instruments is that they do not provide direct kinaesthetic feedback to the user. Some hand-held palpating instruments have also been designed to be wireless to provide better freedom of motion and avoid entanglement issues [45,50].

2.3.2 Hand-held Grasping Instruments

Grasping instruments provide an advantage over one-sided palpating instruments since they provide two well-constrained rigid surfaces between which the tissue can be grabbed,
and this allows the contact pressure and stiffness measurements to be more repeatable [51, 52]. Another advantage is that they can be used to manipulate tissue as well. But grasping instruments also come with drawbacks, such as the inability to reach areas of the organ away from the periphery due to fixed jaw length, the risk of accidentally shearing, pinching or otherwise damaging tissue when manipulating the organ between the jaws, etc.

Some implementations of grasping instruments, such as the one developed by Bicchi et al. [53], are as simple as applying strain gauges on a standard Babcock grasper to measure grasping forces. The knowledge of the grasping force along with the extent of tissue deformation can be used to determine tissue stiffness. Their design used strain gauges near the handle to measure grasping force, which is not ideal, since the force at the handle is distorted by backlash and friction.

Some other implementations involve developing a custom instrument that incorporates the sensors right at the jaws or very close to them in order to improve force measurement accuracy. An example is the instrument developed by Tholey et al. [54] that incorporates four piezoresistive force sensors and a thin film pressure sensor in one of the jaws to directly measure the 3D grasping forces. The benefit of measuring 3D grasping force is that the instrument can also be used in lateral sliding and axial probing mode to locate tumours if required. This instrument is one of the very few existing instruments that utilize an automatic grasping algorithm to achieve consistent results.

Another notable example of a grasping instrument is the one developed by Kurowski et al. [55, 56] that has two 60 mm long jaws. It was intended to have a tactile sensor on one of the jaws and an ultrasound transducer on the other to obtain additional data for better tumour localization. The instrument was also designed to include strain gauges on the jaw linkages to measure the grasping force. This instrument has two coupled DOFs at the jaws, one to control the distance and another to control the angle between the jaws in order to allow tissue grasping with even pressure. The jaws are actuated by cables that are driven
by two motors located in the instrument handle. However, the advantage of the two DOFs to avoid pinching can only be accomplished with very good control that compensates for cable stretch and friction, which had not been developed yet.

2.3.3 Robot-Manipulated Instruments

Some researchers have explored the use of robotic assistance to localize tumours in order to achieve better accuracy, repeatability and safety than hand-held instruments. This is because a master–slave robotic system can compensate for hand motion reversal, force magnification and poor dexterity, and make the instrument easier to manipulate. A master–slave system can also relay physical kinaesthetic information to the surgeon in an intuitive manner. The use of a robotic system allows the palpation process to be automated to achieve consistent results. It should be noted that almost all of the robotic tumour localization systems found in the literature have been developed to be used with a custom or non-surgical robotic platform rather than utilizing an existing standard platform such as the da Vinci surgical system.

Some robotic systems use components from existing MIS instruments and add actuators and sensors (force and position) on them to convert them into a master–slave system. An example is the system proposed in [57] that uses a modified Babcock grasper to perform an automatic palpation routine consisting of three sinusoidal cycles at 1 Hz at every location to characterize the tissue stiffness. Experimental results indicate that this instrument can distinguish different mechanical properties of tissues. In [58], several strain gauges and a load cell integrated into a custom endoscopic instrument is used to characterize tissue. A different approach is proposed in [59], in which the magnitude of current applied to the actuator of a motorized grasper is used to determine tissue stiffness.

Several custom master–slave systems have also been developed that employ a tactile
sensor on the end of a custom instrument that is manipulated by a robotic arm. A notable example is [60] in which a Pressure Profile Systems TactArray tactile sensor is placed on the end of a rigid instrument mounted on a Mitsubishi PA-10 industrial robot. An industrial force/torque sensor is positioned at the interface between the instrument and the end effector of the robot to enable palpation force control. The system was used to automatically palpate porcine lung tissue samples and it was able to locate 5 mm agar tumours embedded within. Under automatic palpation, the maximum pressure of palpation decreased by 35% and tumour localization success rate increased by 50%, as compared to manual palpation. Another example is [61] in which an optical tactile sensor is mounted on the tip of a rigid shaft attached to an industrial robot. This system was able to successfully find 5 mm tumours hidden in a silicone phantom and a lamb kidney.

2.3.4 Information Relaying Techniques

Some researchers, such as [59], have conducted studies to try and identify the best technique for relaying the tactile and kinaesthetic information to the surgeon. The results show that the performance with only visual feedback is comparable to physical feedback. However, a combination of both, such as physical kinaesthetic feedback combined with visual tactile feedback, outperforms either single feedback method. Providing proper physical kinaesthetic feedback is usually possible only if a robotic master–slave system, such as the da Vinci surgical system, is used in which the master controller has actuators to apply force to the user. In this scenario, it is very common to have physical kinaesthetic feedback because it is easy to implement. In cases where a hand-held instrument is used, only visual feedback is typically possible.

Some researchers have created physical tactile feedback systems as well to assess if they provide any benefits over visual tactile feedback. Ottermo et al. [62] have attempted
to create a physical tactile feedback system using a grid of pins with controllable height that can be used on a hand-held instrument. Li et al. [63] have created an alternative system that uses small pneumatic actuators to provide tactile feedback. The latter is a less expensive system, but it has a much poorer spatial resolution that limits the amount of detail that can be relayed to the user. These systems do provide an improvement over visual feedback alone, but the complexity of incorporating physical tactile feedback outweighs the benefits.

2.3.5 Multiple Modalities

A few researchers have explored the benefits of combining multiple modalities to improve tumour localization in the past. Tactile sensing can only reliably provide the 2D location of the tumour on the tissue surface. It cannot precisely identify the size and the depth of a tumour because the obtained tactile information represents a combination of these two factors. A solution to overcome this limitation is to incorporate an ultrasound transducer because 2D ultrasound images can display the size and the depth of the tumour within the scanning plane.

A multi-modal teleoperated device for breast tumour localization was constructed in [64]. The device provided haptic feedback to the physician and included both tactile sensing and ultrasound. It was claimed that the system may outperform the physician’s own hand. Although this specific design is not suitable for MIS, it demonstrated that including multiple modalities is an attractive option to pursue due to the possibility of increasing tumour detection performance. A similar, but hand-held, instrument used ultrasound and tactile pressure sensing for breast examination was proposed in [65]. The two sensors were mounted at different locations on the device, and thus the data fusion process required that the respective positions of both sensors be tracked. The sensor information was combined to determine the position and the depth of a tumour. Due to the inclusion of tracking,
the device had the capability to repeatedly palpate in the same orientation with the same amount of force for subsequent examinations. This feature is desirable for lung tumour localization since a surgeon may need to re-examine a previously identified area of interest. Another hand-held grasping instrument developed specifically for lung tumour localization also used both ultrasound and tactile sensing [55, 56]. This instrument had the ultrasound transducer on one jaw and the tactile sensor on the other. This setup allowed the sensors to always be aligned with each other and individual sensor tracking was no longer required. Both sensors could be used to collect data simultaneously for the area being palpated which simplifies the data fusion process.

2.4 Summary and Conclusions

This chapter presented a concise, but comprehensive, review of the existing tactile sensors, miniature force/torque sensors and MIS instruments that are related to tumour localization. The gathered information was used to determine the direction of research so that the novel sensors and instruments presented in this thesis solved some of the problems identified in the existing designs.

It was recognized that tactile sensing is very useful because it transfers the most amount of information about the palpated region when compared to one-dimensional (point-by-point) palpation techniques and techniques involving only kinaesthetic feedback. With good spatial resolution, a tactile sensor can make it easier to visualize the underlying tissue and provide the ability to distinguish between tumours and bronchial tubes. It was noted that most existing designs suffered from one or more of the following problems – lack of sterilizability, large number of wires, low spatial resolution, small palpation area, and high cost. Therefore, the development of a novel tactile sensor that can address these issues is a worthwhile endeavour.
There are also several compact force sensor designs that have been proposed for use in MIS tools, but they are either very difficult and expensive to manufacture, or they do not provide the required measurement degrees of freedom for palpation using a robotic instrument with an articulated wrist. Some existing designs meet the size and DOF requirements, but do not have sufficient measurement range. Therefore, the development of an inexpensive and compact 6-DOF force/torque sensor, that is biocompatible, sterilizable and has an adequate measurement range for palpation is also warranted.

In terms of minimally invasive palpation instruments, a few robot-manipulated and hand-held designs have been proposed in the past. However, the robot-manipulated instruments utilize custom or non-surgical robotic platforms, and are therefore not able to benefit from the established use of the acclaimed da Vinci surgical robot platform. It has also been seen that combining ultrasound with tactile sensing can result in more accurate tumour localization. A da Vinci surgical robot compatible MIS instrument that combines tactile sensing, force sensing and ultrasound imaging capability in one tool with an articulated wrist has never been developed in the past, and hence it is worthwhile to explore the benefits of using such a mechatronic system for tumour localization.

Although robot-manipulated instruments have several advantages, economic considerations do not allow for the use of robotic assistance in all procedures. Therefore, hand-held palpation instruments that provide tactile feedback for tumour localization are also required. The existing hand-held instruments are either rigid and do not allow palpation in a wide range of orientations, or have actively controlled DOFs that make them cumbersome to use. Also, no existing wireless hand-held palpation instrument has been developed to be sterilizable. Therefore, the development of a wireless hand-held palpation instrument that is sterilizable and has passive DOFs to provide the necessary orientation flexibility is advantageous.
References


Chapter 3

Tactile Sensors

3.1 Introduction

This chapter presents the design, analysis and evaluation of novel disposable tactile sensors developed for minimally invasive tumour localization, with specific focus towards detecting 5–20 mm sub-surface Non-Small Cell Lung Cancer nodules. Tactile sensor arrays have been proposed as an alternative for tumour localization in several publications, but current designs have drawbacks including a large number of wires, high cost and lack of sterilizability. The proposed sensors have been designed with all of the analog signal processing electronics on board and have a purely digital four-wire interface to avoid these problems. The low-cost and simple interface allow the sensors to be made disposable, which circumvents the challenges of resterilizing the sensor at clinical centres. The sensors utilize well-established piezoresistive and capacitive sensing technologies in an innovative, easy to manufacture design. This chapter reports the major design specifications of the sensors, describes their mechanical and electrical designs, presents a mathematical model for their operation and discusses the evaluation of the prototypes.
3.2 Design Specifications

The basic design requirements that were developed to guide the design of the tactile sensors are listed below:

1. Must fit through a 12 mm diameter or smaller trocar
2. Must provide a spatial resolution of at least $2 \text{ mm} \times 2 \text{ mm}$
3. Must be able to measure pressures up to 150 kPa without saturating
4. Must have the minimum possible number of interface lines to allow for easy integration in robotic instruments with an articulating wrist and other minimally invasive surgical instruments
5. Must either be disposable or be able to withstand autoclaving so that it may be sterilized and re-used.

The spatial resolution requirement was established based on the minimum size of a lung tumour nodule that is typically excised surgically. Due to the limitations of diagnostic imaging, a pulmonary nodule is usually identified for excision only when it is at least 5 mm in diameter. Nodules smaller than this rarely present any symptoms and are rarely discovered, unless the nodule eventually does grow to 5 mm or larger in diameter. The $2 \text{ mm} \times 2 \text{ mm}$ spatial resolution was chosen such that at least four sensing elements would detect the presence of the stiff nodule at a time for greater reliability and localization accuracy. This spatial resolution will also allow tumours to be distinguished from bronchial tubes.

The length of the active sensing area was chosen such that the sensor can completely cover a 20 mm tumour along with some healthy tissue, thereby allowing the difference in stiffness to be registered. The 150 kPa measurement range was established based on past studies that indicate average palpation pressure beyond 37 kPa causes permanent lung tissue damage [1]. The maximum average pressure was increased by about four times to obtain the measurement range so that the sensor can accommodate an increase in local
3.2 Design Specifications

pressure due to the presence of a tumour. Since each sensing element is $2 \times 2$ mm, this translates to a force range of 0 N to 0.6 N that each element must be calibrated for.

It was determined early in the design process that it is preferable to make the sensor disposable since it is extremely difficult to design sensors that can withstand repeated autoclaving (the most widely used sterilizing technique in clinical centres) and still perform reliably. Designing a disposable sensor also added two more design requirements: low cost and easy replacement. Based on the cost of various types of disposable medical equipment used in surgeries such as ablation catheters, IV infusion supplies, etc., it was determined that about US$100 is a reasonable price limit for a single-use sensor. Detailed market analysis is required to establish the ideal price point for these sensors. The requirement for easy replacement imposed a condition that the sensor must attach to the instrument using some form of a quick connect mechanism. The limited space available in a minimally invasive instrument for such a mechanism limited the sensor to be electrical rather than optical since even the smallest optical connectors are at least a few millimetres in diameter. This also further emphasized the need to have the minimum possible number of interface lines.

To meet the aforementioned requirements, it was decided that the best approach is to use electrical multiplexing by placing some electronics on-board the sensor. Piezoelectric sensing was avoided due to the complex electronics required for its operation, which are too bulky for the available space. Piezoresistive and capacitive sensing were both considered feasible and therefore designs using each of them were constructed to investigate which one provides better performance. The tests conducted to date with the sensors show that both sensing technologies have comparable performance. Further tests with various phantoms and ex-vivo tissue samples are required to determine if one performs better than the other, given variations in the average stiffness of the healthy tissue, and the size, depth and stiffness of the tumour. With the freedom to choose the sensing technology, the use of these tactile sensors may be expanded to localizing tumours in the liver, the kidneys
3.2 Design Specifications

and other internal organs. Over several iterations of the circuit design, it was observed that it is preferable to have all of the analog signal processing and analog-to-digital conversion circuitry on board. This approach resulted in the sensor requiring a minimum of only four interface lines, two for power and two for digital communication. This approach also has an added benefit of not having long analog signal lines that may be susceptible to Electro-Magnetic Interference (EMI) from the operating room equipment.

Two different size options have been developed for both the piezoresistive and the capacitive tactile sensors, with the same spatial resolution of $2 \times 2$ mm for all of the designs. The two sizes are intended to allow for a choice based on patient size, preferred trocar size, tumour size, etc. The smaller size option may be used if a small trocar must be used to reduce trauma to the patient, while the larger size option may be used to palpate a large area quickly. The piezoresistive and capacitive versions of the same size look identical on the exterior, and have the same pin-out to allow for interchangeability. The larger sensor is $49 \times 10 \times 2$ mm in size with 90 elements covering a $36 \times 10$ mm sensing area, while the smaller one is $30 \times 8 \times 2$ mm in size with 48 elements covering a $24 \times 8$ mm sensing area.

Ease of manufacture is an important design factor for reducing cost and increasing commercializability. The sensors are designed to use standard materials and processes that have been well established in industry to allow for high manufacturing reliability. The specifications of the final tactile sensor designs are listed in Table 3.1. It must be noted that the quoted accuracy and repeatability are for force measurement per element after individual element calibration. These values hold for quasi-static forces in the range of 0.1 N to 0.6 N during 0.2–0.5 Hz palpation. The accuracy is low for forces below 0.1 N due to mechanical separation of the sensing electrodes from the sandwiched piezoresistive or dielectric material. This, however, can be desirable since the drastic change in the readings when the force increases from 0 N to 0.1 N can be used to detect tissue contact. If accurate force measurement is not required, then individual element calibration is not necessary.
Table 3.1: Major design specifications of the novel tactile sensors.

<table>
<thead>
<tr>
<th>Design Specifications</th>
<th>Overall Size</th>
<th>Sensing Area</th>
<th>Number of Sensing Elements</th>
<th>Spatial Resolution</th>
<th>Force Measurement per Element</th>
<th>Scanning Frequency</th>
<th>Materials Cost (US$)</th>
<th>Sterilizability</th>
</tr>
</thead>
<tbody>
<tr>
<td>Overall Size</td>
<td>Large size: 10 mm × 49 mm × 2.5 mm &lt;br&gt;Small size: 8 mm × 30 mm × 2.5 mm</td>
<td>Large size: 10 mm × 36 mm &lt;br&gt;Small size: 8 mm × 24 mm</td>
<td>Large size: 5 × 18 = 90 &lt;br&gt;Small size: 4 × 12 = 48</td>
<td>2 mm × 2 mm</td>
<td>Range: 0.1–0.6 N (25–150 kPa pressure) &lt;br&gt;Accuracy: 84% &lt;br&gt;Repeatability (including hysteresis): 86%</td>
<td>30 Hz</td>
<td>Large size: $40 for electronics + $10 for the rest &lt;br&gt;Small size: $25 for electronics + $5 for the rest</td>
<td>Once during manufacture &lt;br&gt;Ozone or gamma irradiation, no heat</td>
</tr>
</tbody>
</table>

3.3 Mechanical Design

3.3.1 Printed Circuit Board

A printed circuit board (PCB) forms the core mechanical structure of the sensor. The finished PCBs for the all of the proposed sensor designs have all of the electronic components and the four connection terminals located on the top side, and all of the sensing
elements on the bottom side. The electrical designs require the use of at least a three-layer PCB to make all of the connections. Since the bottom layers are made up of sensing pads and do not allow any other PCB traces or through-hole vias (electrical connection between different PCB layers) to be routed there, traditional PCB manufacturing techniques are not sufficient. A special feature called “blind vias” is required that significantly drives up the cost. This is not favorable, especially for low quantity production.

Figure 3.1: 2D CAD renderings of the Printed Circuit Boards (PCBs): (i) top side of upper PCB, (ii) bottom side of upper PCB, (iii) top side of lower PCB, (iv) bottom side of lower PCB; (a) large piezoresistive sensor, (b) large capacitive sensor, (c) small piezoresistive sensor, and (d) small capacitive sensor.
3.3 Mechanical Design

As an alternative, a novel technique to make the three-layer PCB is proposed. In this technique, the middle layer is broken into two parts, one that has all of the connections to the top layer and the other that has all of the connections to the bottom layer. Now the PCB is manufactured as two regular 0.6 mm thick two-layer PCBs, as shown in Fig. 3.1. The PCBs have an immersion gold finish to keep the sensing pads smooth and flat. The bottom layer of the upper PCB and the top layer of the lower PCB have exposed connection points where the top and bottom halves of the middle layer must connect. The exposed points are coated with lead-free solder paste (a mixture of powdered solder and flux) using a stencil. The upper and lower PCBs are then placed in a mould that keeps the connection points aligned, and it is heated up to 280°C using a hot-plate to allow the solder paste to melt and make strong electrical connections that hold the two PCBs together. This results in an inexpensive pseudo-three-layer PCB that is only 1.2 mm thick.

3.3.2 Components Side

Fig. 3.2 shows CAD renderings of the component (top) side of the PCBs with the components in place. All of the components used are of surface-mount type and the maximum height is only 0.9 mm. After soldering the components, they are encapsulated in a 1 mm thick layer of sterilizable Loctite E-60NC epoxy to protect and seal the electronics. The epoxy has been specifically developed for encapsulation and has low viscosity to allow it to cover the electronics evenly. The encapsulation is done by placing the sensor in a Teflon mould to protect the sensing pads and the connection terminals from the epoxy. After the epoxy cures, it is sanded down to bring it to the correct thickness, and make it flat and smooth. In a commercial production process, the fusing of the two PCBs with solder paste, and the soldering of the components can be combined into one streamlined step, and several sensors can be manufactured simultaneously by using a PCB panel and a larger mould.
Figure 3.2: Dimensioned 2D CAD renderings of the printed circuit boards with the electronic components placed: (a) large piezoresistive sensor, (b) small piezoresistive sensor, (c) small capacitive sensor, and (d) large capacitive sensor.

3.3.3 Sensing Side

For all of the proposed designs, the sensing side of the PCB is comprised of a sheet of sandwiched piezoresistive or dielectric material that covers the sensing pads, which in turn is covered and held in place by a metal foil, as can be seen in Fig. 3.3. The foil is glued down to the two ground strips using a sliver-based conductive epoxy such as MasterBond EP77M-F. This foil is the common ground electrode for all of the sensing elements. Since the pressure is being sensed through this foil, it must be only 20-30 µm in thickness to avoid coupling among the sensing elements. It was found that 25 µm (1 mil) thick copper foil worked the best for this application.
The specific piezoresistive material used in the piezoresistive version of the sensor is known as Velostat and is manufactured by 3M Company to make electrostatic discharge (ESD) protection bags for electronics. It is a carbon black impregnated polyolefin that is about 0.2 mm in thickness, and has excellent piezoresistive characteristics. It is a low-cost (only US$4 per square foot) standardized commercial product that can be easily sourced, making it preferable over specialized difficult to obtain materials.

The specific compressible dielectric material used in the capacitive version of the sensor is known as Elastosil Film and is manufactured by Wacker Chemie AG for Electro-Active Polymer (EAP) applications. Elastosil Film is a line of ultra-thin cross-linked platinum-cured 100% silicone elastomer films. The dielectric material must be thin and have a low Young’s modulus $E$ for the best sensitivity, while still resisting tear and puncture. The specific Elastosil film chosen is only 20 µm in thickness and has a relatively low Shore-A hardness $S_A$ of 30A, while still being strong enough for this sensor. Based on the approx-
imate relation between $S_A$ and $E$ in MPa given by Eq. (3.1), the Young’s modulus of this material is about 1.2 MPa [2]. The relative permittivity $\varepsilon_r$ of this material is 2.1. Elastosil Film is has not been released to the general market yet, hence its market price is unknown, but it is not expected to be very high. The films used in the prototypes were generously provided by Wacker as free samples.

$$E = 10^{(0.0235S_A−0.6403)}$$

(3.1)

A CAD rendering of the tactile sensors with the components and sensing sides completed is shown in Fig. 3.4.

Figure 3.4: 3D CAD renderings of the bottom (left) and top (right) of the tactile sensors: (a) large size version, and (b) small size version.

3.3.4 Protective Covering and Sterilization

The last steps in the manufacturing of the sensor are to cover it in a protective layer so that contaminants do not get in between the sensing electrodes, and to sterilize the sensor.
The best option for covering the sensor that has been discovered so far is a single layer of medical grade polyurethane adhesive tape. This tape is thin (30 µm), waterproof, long lasting and has a strong adhesion. The effectiveness of this sealing technique when the sensor is used in a very wet environment has not yet been tested, however it allowed the sensor to be used on a moist porcine liver tissue sample without any damage.

Since the sensors have been designed to be disposable, they have to be sterilized only once at the manufacturing site. This allows the use of a no-heat sterilization process such as ozone sterilization or gamma irradiation to prevent damage to the sensor. The equipment to use these techniques on the tactile sensors was not accessible, and therefore their effects on the sensor have not yet been determined. But based on the fact that these sterilization techniques are used on plastic and rubber, there is a very low likelihood that they would damage the sensor [3].

3.4 Electrical Design

In all the sensor versions, one of the electrodes that sandwich the piezoresistive or the dielectric material is simply connected to the electrical ground of the sensing circuit. This electrode is hence shared between all of the sensing elements. The other electrode, referred to as the “sensing pad”, is distinct for each sensing element and all of the pads are individually connected to the sensing circuit. The final output of all of the sensing circuits is a pair of Inter-Integrated Circuit (I²C) protocol synchronous digital communication lines (SDA: Data and SCL: Clock) that are used to obtain measurements from the sensor.

3.4.1 Piezoresistive Tactile Sensor

Fig. 3.5 illustrates the sensing circuits used in the large and small versions of the piezoresistive tactile sensor. In the large version, 30 sensing pads are connected to a single
ADG732 32:1 analog multiplexer (mux). Three such multiplexers are used to cover the 90 pads. The outputs of the multiplexers have pull-up resistors that form the high-side resistor $R_s$ of the voltage divider. A PIC16F1508 microcontroller is used to simultaneously switch the multiplexers via five shared address lines (A0-4) using digital output pins (RD0-4). The outputs of the three voltage dividers feed into three 10-bit analog-to-digital converter (ADC) input pins (RA0-2) of the microcontroller through RC low pass filters. The microcontroller scans the 30 channels of the multiplexer while taking 8 ADC samples from each analog input per channel. The entire sensor is scanned at 30 Hz. The low pass filter was designed to have a 1600 Hz cut-off frequency to obtain the best compromise between noise reduction and settling time. The small version has a very similar circuit that uses one ADG732 32:1 and one ADG1606 16:1 analog multiplexer to cover the 48 pads. A smaller PIC16F1503 microcontroller is used instead of the PIC16F1508 to save space. The sampling and scanning rate of the small sensor is the same as the large sensor.

### 3.4.2 Capacitive Tactile Sensor

Fig. 3.6 illustrates the sensing circuits used in the large and small versions of the capacitive tactile sensor. In the large version, 10 sensing pads are connected to a single AD7147-1 13-channel capacitance-to-digital converter (CDC), while in the small version 12 sensing pads are connected to a single CDC. This CDC is designed to be used for touch user interfaces and can measure capacitances up to 16 pF with a resolution of 0.24 fF. It is used in this sensor because the capacitance of the sensing elements falls in its measurement range. It also conveniently uses the $I^2C$ communication protocol and allows for up to four of them to exist on the same $I^2C$ line. The small version uses only four chips to cover the 48 pads, so the circuit is really simple. On the other hand, the large version uses nine chips to cover 90 pads, hence a PCA9546A 4:1 $I^2C$ multiplexer is used. This $I^2C$ multiplexer allows for
the channels to be switched by simply using \(I^2C\) commands, which eliminates the use of extra address lines. Three channels of the \(I^2C\) multiplexer are used to connect two groups of four and a single AD7147-1 CDC to a single output \(I^2C\) line. The CDC is programmed to automatically perform 256 sample averaging for each connected pad, and it allows for the entire sensor to be scanned at 30 Hz.

Figure 3.5: Schematic diagrams of the piezoresistive tactile sensors: (a) large size version (90 pads), (b) small size version (48 pads).
Figure 3.6: Schematic diagrams of the capacitive tactile sensors: (a) large size version (90 pads), (b) small size version (48 pads).

3.4.3 Interface Circuit

Fig. 3.7 illustrates the interface circuit used for connecting the tactile sensors to a computer via USB. The core of the circuit is a Parallax Propeller V1 microcontroller that communicates with the tactile sensor using the \( \text{I}^2\text{C} \) protocol, and with the computer using the Universal Asynchronous Receive and Transmit (UART) protocol. This microcontroller
was chosen because it has eight cores that can run eight different processes simultaneously at 20 million assembly instructions per second, while allowing the processes to share data. This feature was considered advantageous since the microcontroller can use two of the cores to communicate with the tactile sensor and the computer simultaneously while also repackaging the data packets in another core, thereby supporting a large data throughput. This breakdown of tasks among three cores allowed for simpler programming and faster software development. The microcontroller is programmed to automatically detect which of the four different sensor designs is connected to it and establish communications accordingly. The I₂C communication is set to run at a relatively low speed of 100 kHz to ensure minimal data corruption while still maintaining the required data throughput. The UART protocol is translated into the USB protocol with the help of an FTDI FT230X UART-to-USB transceiver. The UART communication is set to run at 115.2 kHz, which is a standard UART speed that is close to the I₂C speed used on the sensor side. The software running on the computer uses the FTDI D2XX library to interface with the FT230X over USB and receive the data being transmitted by the microcontroller. The circuit also includes an ADP3335 Low-Dropout Linear Voltage Regulator to obtain 3.3 V to power the circuit from the 5 V USB power supply. The net current requirement for this circuit is about 70 mA, which is much lower than the 500 mA current limit for USB powered devices.

3.5 Mathematical Modelling

3.5.1 Piezoresistive Tactile Sensor

The piezoresistive versions of the tactile sensor use a carbon black impregnated piezoresistive material sandwiched between two flat electrodes. The resistance between the electrodes decreases with an increase in applied pressure. As described in [4], based on
the Quantum tunnel effect theory, the relationship between the resistance $R$ and the applied force $F$ over a specified contact area $A_c$, is described by Eq. (3.2) where $A_1$, $B_1$ and $R_0$ are constants that depend on the material. Using Eq. (3.3), Eq. (3.2) can be normalized with respect to the contact area $A_c$ used for characterization and the nominal thickness $d_0$ (assumed constant) of the piezoresistive material to obtain Eq. (3.5), which gives the relationship between the resistivity $\rho$ and the applied pressure $P$.

$$R = B_1 F^2 + A_1 F + R_0$$  \hspace{1cm} (3.2)

$$R = \frac{\rho d_0}{A_c}; \quad F = P A_c$$ \hspace{1cm} (3.3)

$$\Rightarrow \rho = \left( \frac{B_1 A_3}{d_0} \right) P^2 + \left( \frac{A_1 A_c^2}{d_0} \right) P + \left( \frac{R_0 A_c}{d_0} \right)$$ \hspace{1cm} (3.4)

$$\Rightarrow \rho = C_1 P^2 + C_2 P + \rho_0$$ \hspace{1cm} (3.5)
3.5 Mathematical Modelling

$C_1$, $C_2$ and $\rho_0$ are constants to be determined via experimental characterization. Using Eq. (3.5), Eq. (3.2) can be rewritten as Eq. (3.6), where $A$ is now the contact area of a single sensing element in the sensor. These steps are taken since the contact area that is used for the characterization procedure can be different from the contact area in the actual sensor, especially if the sensor has a very small contact area. The resistance $R$ can be measured by simply using a voltage divider with a series resistor $R_s$ on the high-side and a known excitation voltage $V_0$. The value of $R_s$ is determined during calibration to account for the resistance introduced by the measurement circuit. The measured voltage $V$ is then related to $R$ by Eq. (3.7).

\[
R = \left( \frac{C_1 d_0}{A^3} \right) F^2 + \left( \frac{C_2 d_0}{A^2} \right) F + \left( \frac{\rho_0 d_0}{A} \right)
\]

\[
V = \frac{R}{R + R_s} V_0
\]

Rearranging Eq. (3.7), substituting it in Eq. (3.6) and solving for $F$ as a function of the measured voltage $V$ yields Eq. (3.8), where $K_1$, $K_2$ and $K_3$ are constants given by Eq. (3.9).

\[
F = K_1 + \sqrt{K_2 + \frac{K_3}{V_0 \left( \frac{V}{V - 1} \right)}}
\]

\[
K_1 = \frac{C_2 A}{2C_1} ; \quad K_2 = K_1^2 - \frac{A^2 \rho_0}{C_1} ; \quad K_3 = \frac{A^3 R_s}{C_1 d_0}
\]
3.5.2 Capacitive Tactile Sensor

The capacitive versions of the tactile sensor use a compressible dielectric material sandwiched between two flat electrodes to make a parallel plate capacitor. The capacitance between the electrodes increases with an increase in applied pressure. This happens because the applied pressure reduces the distance $d$ between the electrodes, which is inversely related to the measured capacitance $C$ as shown in Eq. (3.10). $A$ is the electrode area, $\varepsilon_0$ is permittivity of free-space, and $\varepsilon_r$ is the relative permittivity of the dielectric. $C_p$ is the parasitic capacitance that is introduced by the circuit that is used to measure the true capacitance, and it is determined during calibration. The change in the distance $d$ as a function of the applied force $F$ is given by Eqs. (3.11) and (3.12), where $e$ is the engineering strain, $d_0$ is the nominal uncompressed dielectric thickness, $\sigma$ is the applied stress, and $E$ is the Young’s modulus of the dielectric material.

\[
C = \frac{\varepsilon_0 \varepsilon_r A}{d} + C_p \quad (3.10)
\]
\[
e = \frac{\Delta d}{d_0} = \frac{d_0 - d}{d_0} = 1 - \frac{d}{d_0} \quad (3.11)
\]
\[
e = \frac{\sigma}{E} = \frac{F}{EA} \quad (3.12)
\]

Using Eqs. (3.10), (3.11) and (3.12), and solving for $F$ as a function of the measured capacitance $C$ yields Eq. (3.13), where $J_1$ and $J_2$ are constants given by Eq. (3.14).

\[
F = J_1 + \frac{J_2}{C - C_p} \quad (3.13)
\]
\[
J_1 = EA \; ; \; J_2 = \frac{\varepsilon_0 \varepsilon_r EA^2}{d_0} \quad (3.14)
\]
3.6 Visualization Software

Fig. 3.8 shows the visualization software developed for the tactile sensor. The figure presents the user interface in three different states displaying the controls that are enabled or disabled in each. The user interface can be divided into three sections from top to bottom: the connection management section, the display section, and the calibration section. The connection management section has a pair of radio buttons to indicate whether a piezoresistive or a capacitive tactile sensor has been detected by the software, and a button to connect or disconnect from the sensor. The display section has a colour-contour tactile pressure map, and a vertical bar indicating the applied force as a simple form of visual kinaesthetic feedback. A slider bar to adjust the sensitivity of the tactile pressure map, and check-boxes to enable or disable compensation using the calibration data and filtering are also included. The calibration section has three buttons and a progress bar for performing the bulk calibration of the sensor (discussed later in this section).

3.6.1 Display

The tactile pressure map in the visualization software displays the measured contact pressure using a continuous spectrum colour scale where blue, green and red correspond 20% or less, 60% and 100% of the maximum pressure respectively, and the rest of the colour spectrum in between is scaled linearly. The maximum pressure is approximately 150 kPa (0.6 N per element) when the display sensitivity is set to the minimum, and approximately 25 kPa (0.1 N per element) when the display sensitivity is set to the maximum. If the bulk calibration and compensation feature is disabled, the best performance was obtained when the sensitivity is set to about 75%. The display resolution of the sensor is tripled along both axes using bicubic interpolation to make a smoother force distribution map that has a better visual appeal and makes it easier to visualize the size of the tumour.
3.6 Visualization Software

Figure 3.8: The tactile sensor visualization software in three different states: (a) not connected, (b) connected but not calibrated, and (c) connected and calibrated.

The vertical force bar displays the approximate total force on the tactile sensor as a percentage of the maximum allowable force established during the calibration process. Ideally, the applied force should be the same for each palpation so that the results are consistent and repeatable. To allow the surgeon to accomplish this, the force bar changes
its colour to communicate if the applied force is too high, too low or just right. Based on empirical experimentation, it was established that the ideal force range is 70% to 90% of the maximum force, which is indicated by a green force bar. If the force is too low or too high, the bar turns yellow or red respectively. If the force crosses the maximum allowable force, the tactile pressure map turns grey and a warning message is shown. It must be noted that this force bar is activated only after the min–max calibration steps of the bulk calibration process has been completed.

3.6.2 Bulk Calibration and Filtering

When the tactile sensors were initially tested without any form of individual element calibration, it was seen that some elements had higher sensitivity than others due to inconsistencies in the PCB surface finish and the sandwiched material. This resulted in green patches appearing on the tactile pressure map even when there is no tumour underneath. This can be seen in the results presented in the evaluation section. The tumours were still identifiable since they appeared as distinct yellow-red spots on the pressure map, but the green patches can be distracting and confusing for the user, making it important to eliminate them. These patches could be eliminated by calibrating each individual sensor element to measure the true pressure, thereby cancelling the effects of the variation. However, this is a very tedious process, and it is preferable to avoid doing it. An alternative bulk calibration approach was developed as a solution that proved to work very well.

The bulk calibration process involves three quick steps that can be performed via the user interface right before using the sensor. The button for executing the next step is enabled only after completing the previous step as a means of ensuring that the steps are completed in the correct order. The progress of each step is displayed via the small progress bar in the calibration section. These steps are outlined below:
1. *Establish minimum force:* This step is initiated by clicking on the “Set Min” button on the user interface. Before clicking the button, the tactile sensor must be pressed very lightly (approximately 0.5 N) against the surface of the tissue in an area with no tumour. This step records the readings from all the elements for one second to establish the baseline reading for detecting tissue contact.

2. *Establish maximum force:* This step is initiated by clicking on the “Set Max” button on the user interface. Before clicking the button, the tactile sensor must be pressed with the maximum allowable safe force (14 N with the large tactile sensors and 7 N with the small ones that translates to a 30 kPa pressure leaving a 7 kPa margin before tissue damage occurs [1]) against the surface of the tissue in an area with no tumour. This step records the readings from all the elements for one second to establish the reading for tissue contact with the maximum allowable safe force.

A separate force sensor must be used in the above steps to accurately establish the contact force. In a practical use scenario, this may be possible only when using a tissue phantom rather than the actual tissue being palpated, unless the instrument with the tactile sensor has a force sensor built into it. All of the element readings collected in each of these two steps is averaged to get a single value that represents the minimum and maximum contact force for the entire sensor.

A simple experiment was conducted to assess the accuracy of the average reading of all of the elements in representing the true total contact force after the two-step min–max calibration. The setup involved a commercial ATI Gamma force sensor underneath a block of Shore 00-10 silicone with no tumour, and the tactile sensor was pressed on top of the block. Comparing the average tactile sensor reading with the reading from the force sensor showed that the average reading can be used to predict the total contact force with an accuracy of at least 89% for both piezoresistive and capacitive sensors by using the mathematical model presented earlier, even though the individual elements have slightly different
response. This experiment was also repeated with a porcine liver sample instead of the silicone block. It was found that the min–max calibration data obtained with the silicone block could be used to predict the total contact force on the liver sample with an accuracy of at least 82%. This demonstrates that the approach of using the average reading as an indicator of the total force is sufficiently accurate to provide useful kinaesthetic feedback and prevent tissue damage.

Since these first two steps may be cumbersome and/or inaccurate to be performed in the operating room, it can be performed when testing the sensor during the manufacturing process, before the sterilization step. Since the experiment above showed that the calibration performed on a silicone phantom still applies to real tissue, it is not necessary that these two steps be performed on the actual tissue that will be palpated. The two calibration values can be supplied with each sensor, and the visualization software can be easily modified to accept these predetermined values.

At this point, the force bar gets activated. The total contact force is displayed as a percentage, where 0% represents no contact, 1% represents the minimum contact force, and 100% represents the maximum allowable contact force. The mathematical model presented earlier is used along with the min–max calibration data to convert the average reading of the sensor to actual contact force. The force is displayed as a percentage instead of the actual force measurement since it is more intuitive to understand.

3. Live calibration: This step is initiated by clicking on the “Calibrate” button on the user interface. This step must be performed on the actual tissue being palpated to obtain the best results. Before clicking the button, the tactile sensor must be gently resting against the surface of the tissue in an area with no tumour, such that the force bar still reads 0%. Once the button is pressed, the process lasts for 15 seconds, and the small progress bar is used to indicate the progress through the step. During this time, the user is expected to cycle the contact force two to three times between about 5% to about 95% as indicated by the force
bar, and the software records the readings from each individual element separately.

The 450 samples of data collected during the calibration step are used to create a map of the change in the reading of each individual element as a function of the average reading of all the elements. To create this map, the difference between the minimum and maximum average readings established in the first two bulk calibration steps is divided into 50 equal intervals. The 450 samples for each element are then divided into 50 sets based on the interval that the corresponding average reading falls in. All of the samples in each of the sets is averaged to produce 50 equally-spaced calibration data points for each element as a function of the average reading. A three-point moving average filter is used on the 50 data points to smoothen any sharp variations. At this point, the entire calibration process is complete, and the check-boxes for activating compensation and filtering are enabled.

If the compensation check-box is checked, then the live calibration data is used to compensate for the variations in the sensitivity of the individual elements. The compensation process simply involves determining in which of the 50 intervals the current average reading falls in, and then subtracting the calibration data points corresponding to the next higher interval from the readings of each element. If this process results in an element’s pressure reading being negative, it is simply changed to zero. Performing this compensation was found to enable an underlying tumour to be seen very clearly since, for the same average reading, the readings from the elements on top of the tumour are significantly higher than what they would be if there was no tumour. When the compensation is enabled, the colour of the tactile pressure map no longer corresponds to the true element pressure. Instead, it represents how much higher the pressure on an element is than the average pressure on the entire sensor. Experiments show that since the elements on top of a tumour will experience about 15 kPa to 40 kPa higher pressure than the average, depending on the stiffness and depth of the tumour, the display sensitivity must be set to almost the maximum to enable the tumour to be seen as a green-red spot on the tactile map.
3.7 Evaluation

The filtering check-box can be checked to apply an empirically created 2D low-pass filter on the tactile pressure map to make a tumour appear even more distinct by smoothening the map and suppressing noise that results in isolated elements measuring a higher than average pressure. The filtering is performed by convolving the tactile map data with the convolution kernel $F_{LP}$ presented in Eq. (3.15). To allow the filter to be applied on the corner and edge elements, the size of the tactile map is expanded by one in all four directions using bicubic extrapolation. This filter is applied as the final signal processing step before bicubic interpolation is used to increase the display resolution of the tactile map.

$$F_{LP} = \begin{bmatrix} 0.08 & 0.08 & 0.08 \\ 0.08 & 0.36 & 0.08 \\ 0.08 & 0.08 & 0.08 \end{bmatrix}$$ (3.15)

3.7 Evaluation

3.7.1 Prototype

So far, only prototypes of the large size piezoresistive and capacitive tactile sensors have been constructed and evaluated. The small sensors use similar circuits as the large ones and hence they are expected to have identical performance. Figs. 3.9 and 3.10 present the prototype sensors at different stages of the manufacturing process. The prototyped sensors are identical to the proposed design described in this chapter, however the manufacturing process should be significantly improved, optimized and streamlined for commercial production. The prototypes are made using the novel pseudo-three-layer PCB manufacturing technique and demonstrate that the technique is feasible. It may be noted that the prototypes have 12 signal/power connection points rather than 4 and are 56 mm long rather than 49 mm. The 12 connection points are actually 3 sets of the same 4 and therefore allow the
3.7 Evaluation

Sensor to be cut down to 52.5 mm or 49 mm as desired. This design was adopted to allow the sensor to be used in instruments that require the sensor to be longer than 49 mm. An example would be the robotic palpation instrument (Chapter 5) that requires the sensor to be 56 mm long because of the ultrasound module that is used with it. On the other hand, the wireless hand-held instrument (Chapter 6) uses the sensor in its 49 mm long form.

3.7.2 Mathematical Model Calibration

The objective of this calibration process is to determine the constants in the mathematical model of the piezoresistive and capacitive sensors described by Eqs. (3.8) and (3.13) respectively. This is an important step since the mathematical model is used for both determining individual element forces to create the tactile pressure map and for determining the total force based on the average reading.

**Calibration Apparatus:** To be able to properly calibrate the sensors and evaluate the characteristics of individual sensing elements, a specialized calibration device was developed (Fig. 3.11). This device is essentially a small 3D Cartesian positioning system that can move a high-precision unidirectional Futek LSM-300 load cell in any direction with an accuracy of 2.5 $\mu$m. The load cell has a replaceable probe attached to it that has a $2 \times 2$ mm tip contact area, the same size as an individual sensing element. The device can apply a desired downward force with an accuracy of 0.05 N anywhere within the $60 \times 30$ mm workspace, and measure the actual applied force with an accuracy of 0.01 N.

**Piezoresistive Tactile Sensor Calibration:** The first step in calibrating the piezoresistive tactile sensor is to characterize the piezoresistive material by determining the constants $C_1$, $C_2$ and $\rho_0$. To accomplish this, the probe on the calibration device was replaced with a $5 \times 5$ mm contact area probe. The device was then used to apply predetermined pressures on a piece of the piezoresistive material and the resistance across it was measured and recorded.
3.7 Evaluation

Figure 3.9: PCB fusing and component side assembly steps in the manufacture of the prototype tactile sensors: (a) the bare PCBs as obtained from the PCB manufacturer, (b) the stencil used to apply solder paste onto the PCBs, (c) a PCB with the solder paste applied ready for fusing, (d) mould used to align the two PCBs before fusing, (e) the fusing process on a hot plate with the PCBs between two steel bars to ensure proper fusing, (f) the fused PCB with the components soldered, (g) the PCBs placed in the Teflon encapsulation mould, (h) the mould filled with epoxy to encapsulate the electronic components.

This process was repeated for ten samples, and the collected data points are presented in Fig. 3.12. The least-squares best-fit quadratic curve has an excellent fit with an $R^2$ value of 0.98. The calibration was performed up to 750 kPa even though the desired maximum
3.7 Evaluation

Figure 3.10: Sensing side assembly steps in the manufacture of the prototype tactile sensors: (a) the sensing side as seen after the encapsulation process, (b) the components side as seen after the epoxy is sanded down to the correct height, (c) the sensing side with the layers placed on top, (d) the sensing side after applying the protective covering, (e) the components side of the finished sensors.

pressure is 150 kPa just to obtain the characteristics of the material over the entire pressure range that it can measure. The data collected from the ten samples fall within ±8% error margins indicating good repeatability.

Based on this calibration curve, the constants $C_1$, $C_2$ and $\rho_0$ are 9e-5, 0.125 and 55 respectively. Also, based on the calibration curve, the resistance of a single sensing element varies in the range of 0.5 kΩ to 2.7 kΩ. The series resistor $R_s$ was chosen to be 1 kΩ to obtain good voltage sensitivity since it lies within this range. It was observed that the value of $R_s$ used in the model had to be adjusted slightly for every element to achieve good data fit due to imperfections on the PCB surface and in the piezoresistive material. Based on the measured response of several individual sensing elements, the best data fit was obtained on average when $R_s$ was taken to be 1.07 kΩ in the model. Using this data, the constants $K_1$, $K_2$ and $K_3$ were determined to be 2.78 N, -2.06 N² and 3.80 N², respectively, to be used in the model for generating the tactile pressure map from the element readings. For the min–max total force calibration performed as a part of the bulk calibration process described earlier, $R_s$ is kept as a free variable that is determined based on the calibration values to
obtain the best fit for relating the total contact force to the average sensor reading.

**Capacitive Tactile Sensor Calibration:** The model developed for the capacitive sensor assumes that the dielectric material has a linear stress–strain response. The chosen dielectric material is a silicone elastomer, and it has an approximately linear stress–strain relationship for only under about 20% strain [5]. Since its Young’s modulus is 1.2 MPa, the response is almost linear for stresses up to 240 kPa, which is 60% higher than the maximum pressure of 150 kPa that the sensor has been designed for. The only unknown to be determined via calibration is the parasitic capacitance $C_p$. This value was found to vary slightly for different sensing elements due to different lengths of PCB traces, and imperfections on the PCB surface and in the dielectric material. Based on the measured response
3.7 Evaluation

of several individual sensing elements, the average value of $C_p$ that results in the best data fit was found to be 2.7 pF. Using this data, the sensor calibration constants $J_1$ and $J_2$ were determined to be 4.80 N and 17.8 N-pF, respectively, to be used in the model for generating the tactile pressure map from the element readings. For the min–max total force calibration performed as a part of the bulk calibration process described earlier, $C_p$ is kept as a free variable that is determined based on the calibration values to obtain the best fit for relating the total contact force to the average sensor reading.

### 3.7.3 Stress Relaxation Response Test

Both the piezoresistive and the capacitive tactile sensors suffer from stress relaxation, which is seen in Figs. 3.13(a) and 3.13(b), respectively. When a constant force is applied on the sensor for a long duration, stress relaxation results in the measured force gradually
Figure 3.13: Individual sensing element stress relaxation response: (a) piezoresistive sensor, (b) capacitive sensor.

creeping up beyond the applied force over time. Piezoresistive stress relaxation (gradual increase in conductivity under pressure) in the piezoresistive sensor is much more significant than the mechanical stress relaxation in the capacitive sensor. Fortunately, these tactile sensors will be typically used in palpation procedures where the sensor is pressed down on the tissue surface in a cyclic manner and not continuously. Hence, the stress relaxation can be ignored safely for the most part without significant deterioration in the sensor performance during palpation.

### 3.7.4 Palpation Test

A palpation response test was conducted to assess the response of the tactile sensor to cyclic loading and observe the errors due to the stress relaxation effect. A 0.3 Hz cyclic load pattern with the peak force increasing from 0.2 N to 0.6 N was applied on an individual element. The response is presented in Fig. 3.14. Observing the peaks and troughs of
3.7 Evaluation

Figure 3.14: Individual sensing element palpation response: (a) piezoresistive sensor, (b) capacitive sensor.

The plot shows the stress relaxation effect, but as expected, it is not very significant during palpation. The force measurement has a large error at close-to-zero force since the electrodes are not making proper contact with the sandwiched material. Overall, the accuracy and repeatability (including hysteresis) of the force measurement in the range of 0.1 N to 0.6 N during palpation (0.2 Hz to 0.5 Hz) are better than 84% and 86% respectively, which is sufficient. The response time of both the sensors was determined to be approximately 0.3 s which is acceptable for the slow palpation process.

3.7.5 Phantom Test

The last evaluation step was to assess the performance of the sensors in determining the presence of tumours using a phantom. The sensors were also tested on a porcine liver sample with embedded agar tumours using the robotic palpation instrument, however, that experiment is discussed in Chapter 5. For the phantom test, the sensors were mounted on a
simple rigid test tool with no internal articulation. The entire setup used for the test is shown in Fig. 3.15. The phantom (Fig. 3.16), is made of Shore 00-10 silicone rubber simulating healthy tissue, with 5 mm Shore 25A silicone spheres simulating tumours embedded at depths of 1 mm, 4 mm and 10 mm. Fig. 3.17 presents tactile data displayed by the software in various scenarios when not using individual element calibration or the bulk calibration and compensation feature.

![Setup used for the palpation test with the tactile sensor mounted on a simple test tool.](image)

Figure 3.15: Setup used for the palpation test with the tactile sensor mounted on a simple test tool.

When there is no tumour, the display does not have any yellow or red on it (Figs. 3.17(a) and 3.17(i)), but there are irregular patches of green due to the slight variation in the sensitivity of the different sensing elements. The results demonstrate that both of the sensors are able to easily detect the 10 mm deep tumour (red patches on Figs. 3.17(d) and 3.17(l)) without the need for individual element calibration. The test also demonstrates
3.7 Evaluation

Figure 3.16: Silicone phantom used for the palpation test, with three embedded 5 mm diameter spherical silicone tumours.

Figure 3.17: Tactile data visualization results of the phantom test without bulk calibration and compensation: (a)–(h) using resistive tactile sensor, (i)–(p) using capacitive tactile sensor; (a) and (i) no tumour, (b) and (j) 1 mm deep tumour, (c) and (k) 4 mm deep tumour, (d) and (l) 10 mm deep tumour, (e)–(h) and (m)–(p) tumours at different locations along the sensor.

that the sensors can effectively locate a tumour anywhere along the sensor (red patches on Figs. 3.17(e)–(h) and 3.17(m)–(p)). The test was also repeated after performing bulk cal-
3.8 Conclusions

Tactile sensors have the potential to overcome a major limitation of minimally invasive surgery, and improve the performance of MIS in tumour resection. Four novel inexpensive tactile sensor designs have been proposed in this chapter. Piezoresistive and capacitive versions have both been shown to effectively locate tumours in a phantom model. The simple four-wire interface and disposability makes these sensors attractive for clinical use. These sensors can be used in a “snap-in” configuration where it can be simply locked into a tool with contacts that mate with the four connection points on the sensor and a quick-

Figure 3.18: Tactile data visualization results showing the effect of using bulk calibration and compensation: (a)–(c) without compensation, (d)–(f) with compensation; (a) and (d) no contact, (b) and (e) no tumour, (c) and (f) 10 mm deep tumour.
release mechanism. This allows the sensor to be easily installed before use and disposed of afterwards. When the sensors are mounted on such a tool, they should be encased in a short silicone rubber sleeve to protect the sensor and the electrical contacts from body fluids. The disposability allows the sensors to be designed for a one-time chemical sterilization process during manufacturing, and they do not suffer from deterioration due to autoclaving in hospitals.

References


Chapter 4

Miniature 6-DOF Force/Torque Sensor

4.1 Introduction

This chapter presents the design and analysis of a novel six degree-of-freedom force and torque sensor that was developed specifically for use with the robotic palpation instrument described in Chapter 5. However, the unique compact monolithic design makes the sensor very versatile and attractive for use in other minimally invasive surgical instruments as well. Even though the sensor is not predicted to have a very good accuracy based on theoretical estimates, the simplicity and robustness of the sensor still make it a worthwhile design. The predicted accuracy of the force and torque measurements is good enough to allow the sensor to provide useful information about tool–tissue interaction, and it can be used to provide basic haptic feedback to the surgeons. This chapter reports the major design specifications of the sensor, describes its mechanical design, validates the design with finite element analysis, presents a mathematical model and discusses the progress of the prototype.
4.2 Design Specifications

The basic design requirements that were developed to guide the design of the force sensor are listed below:

1. Must provide three DOF force and three DOF torque measurements
2. Must fit through a 12 mm diameter or smaller trocar
3. Must be able to withstand the maximum forces and torques that can be applied on it by the da Vinci robot when it is a part of the robotic instrument
4. Must be easy to manufacture
5. Must be biocompatible, cleanable and sterilizable

The design of the force sensor was developed for compatibility with the robotic instrument presented in the next chapter. The reason for choosing a six DOF sensor over a three DOF one used in most existing sensorized MIS instruments designs is because palpation does not restrict the application of interaction forces to the tip of the instrument. It is essential to know both the direction and the location of the applied force to ensure that the palpating surface is indeed flat against the tissue rather than at an angle. This requirement was established to retain the possibility of making the palpation process semi-automatic in the future.

Based on the analysis of the palpation instrument (Chapter 5), it was determined that the force sensor must be able to withstand a transverse force of 20 N at 70 mm (distance to the instrument tip) from the sensor. This translates to a maximum bending moment of 1400 N·mm on the sensor, which is the limiting loading condition that will cause failure.

From the onset of the design process, it was decided that the force sensor will utilize Fibre Bragg Gratings (FBGs) rather than alternatives such as piezoresistive strain gauges due to the significant advantages of FBGs such as compactness, robustness, linearity, no electrical current, noise immunity, high sensitivity, ability to have multiple sensors on a sin-
4.2 Design Specifications

gle optical fibre, etc. The geometry was designed to be such that it can be easily machined using wire EDM by cutting a 2D contour on two perpendicular faces (discussed in more detail later in this chapter). It was also decided that it is essential to include temperature compensation in the sensor, since surgical instruments operate within the human body that is about 10°C warmer than room temperature. The specifications of the final instrument design are listed in Table 4.1.

Table 4.1: Major design specifications of the novel six DOF force/torque sensor.

<table>
<thead>
<tr>
<th>Design Specifications</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Overall Size</strong></td>
</tr>
<tr>
<td>Square Profile Option: 8.5 mm × 8.5 mm × 4 mm</td>
</tr>
<tr>
<td>Circular Profile Option: 10 mm diameter × 4 mm</td>
</tr>
<tr>
<td><strong>Axial Hole Option</strong></td>
</tr>
<tr>
<td>Maximum 7 mm diameter</td>
</tr>
<tr>
<td><strong>Sensing Technology</strong></td>
</tr>
<tr>
<td>12 FBGs on a single optical fibre</td>
</tr>
<tr>
<td><strong>Measurement Range</strong></td>
</tr>
<tr>
<td>3D Force: ±230 N</td>
</tr>
<tr>
<td>3D Torque: ±550 N·mm</td>
</tr>
<tr>
<td><strong>Failure Loading</strong></td>
</tr>
<tr>
<td>3D Force: ±230 N</td>
</tr>
<tr>
<td>3D Torque: ±1400 N·mm</td>
</tr>
<tr>
<td><strong>Theoretical Worst Accuracy</strong></td>
</tr>
<tr>
<td>3D Force: 0.5 N (0.22% full scale)</td>
</tr>
<tr>
<td>3D Torque: 0.7 N·mm (0.13% full scale)</td>
</tr>
<tr>
<td><strong>Sampling Rate</strong></td>
</tr>
<tr>
<td>1 kHz (using Micro Optics sm130 interrogator)</td>
</tr>
<tr>
<td><strong>Bandwidth</strong></td>
</tr>
<tr>
<td>15 Hz (with 30-sample moving average filter)</td>
</tr>
<tr>
<td><strong>Material</strong></td>
</tr>
<tr>
<td>Titanium Grade 5</td>
</tr>
<tr>
<td><strong>Sterilizability</strong></td>
</tr>
<tr>
<td>Autoclavable</td>
</tr>
</tbody>
</table>
4.3 Fibre Bragg Gratings

4.3.1 Principle

A Fibre Bragg Grating is a Distributed Bragg Reflector (DBR) that is constructed within a short segment (down to 1 mm) of optical fibre. It is essentially a periodic variation in the refractive index of the fibre core, which acts as a wavelength specific dielectric mirror [1]. Hence, it can be designed to reflect a particular wavelength of light, while transmitting the rest. Therefore, an FBG can be used as a filter to block certain wavelengths, or as a reflector for a specific wavelength. The first Bragg grating in optical fibres was demonstrated by Ken Hill in 1978. An FBG works based on the principle of Fresnel reflection, which states that light travelling from one medium to another with a different refractive index has a partial reflection and partial transmission. Equal periodic variation in the refractive index of the medium reflects light with a wavelength equal to twice the grating period due to constructive interference of several reflections.

The only known process to inscribe a fine structure like a FBG on a fibre is by using focused UV laser. Hence, it is necessary to make the fibre photosensitive by either doping the glass with germanium during manufacture, or by hydrogenating the optical fibre using hydrogen gas at low temperature and high pressure (>100 atm) [2]. The exposure to high intensity UV radiation increases the refractive index of the photosensitive optical fibre core. The FBG pattern is inscribed by the UV laser on the photosensitive optical fibre using one of three patterning techniques: sequential (point-by-point), phase-mask or laser interference [2]. Standard FBGs are constructed to reflect infrared light in the 1510 nm to 1590 nm wavelength range. The full width at half maximum (FWHM) of the reflected spectrum from a standard FBG can be as low as 0.1 nm [3]. The intensity of the reflected spectrum depends on the length of the grating and the intensity of UV light used for inscribing it.
4.3 Fibre Bragg Gratings

When the optical fibre with a Bragg grating elongates or shortens due to strain from mechanical stress or temperature variations, the reflected spectrum peak wavelength of the grating shifts proportionally [1]. Mechanical strain changes the grating period due to a change in physical length and hence changes reflected light wavelength. A temperature change also causes expansion or contraction, changing the grating period, as well as changes the refractive index of the fibre, and hence changes reflected light wavelength. Hence, by characterizing the mechanical properties of the optical fibre, an FBG can be used as a robust strain or temperature sensor.

4.3.2 Benefits and Limitations

Optical fibres are now available in sizes down to 120 µm in diameter, which makes FBG sensors a viable option where traditional sensors are too large to fit [3]. FBG sensors have near-perfect strain and temperature transfer with minimal drift and no hysteresis while being compact and flexible [1]. FBG sensors are also immune to electromagnetic noise and are safe in environments where the use of electrical sensors can be hazardous. Structurally, FBG sensors are fatigue resistant, and allow for easy and reliable handling. Due to being optical in nature, the signal can be sent over long distances with minimal distortion. The sensors are also easy to embed, and do not suffer from self-heating, aging or corrosion. Another interesting feature of FBGs is that multiple gratings of the same or different centre wavelengths can be inscribed at different locations along a single fibre, and the strain and/or temperature at the different grating locations can be sensed simultaneously using the appropriate interrogation (spectrum analysis) of the reflected light [4, 5]. This allows for the use of FBG sensors in applications such as shape sensing, sensing deformation of composite structures, etc.

FBG sensors also have some drawbacks, the most significant one being the high cost
of the interrogation device required to precisely measure the wavelength of light reflected from the gratings. The interrogation equipment also has limited wavelength resolution, which can be worse than other strain and temperature sensing options [6]. However, depending on the application, FBG sensors might still be the more economical, if not the only, option. Some other limitations of FBGs arise due to inherent problems with optical fibres such as bend losses, minimum bend radius limitations and difficulty in fixing breaks as they require the use of optical connectors. Optical systems can also be hard to debug since special equipment is required for locating and repairing damaged optical fibres.

4.3.3 Measurement

There are a few different measurement techniques employed by commercial FBG interrogators. The three state-of-the-art techniques are swept-wavelength scanning, spectrometry, and Optical Time Domain Reflectometry (OTDR) [4, 5]. Swept-wavelength scanning is the simplest, slowest (10 Hz maximum scan rate) and cheapest technique. It uses an adjustable wavelength laser to scan a preset wavelength window, and a photodiode to create a spectrum of the received light intensity as a function of the wavelength. Spectrometry is the most popular technique used today. It uses a broadband laser source along with a refracting surface to separate the different wavelength components of the reflected light, and the intensity as a function of position on a light sensor is used to measure the spectrum. Common spectrometry interrogators, such as the Micron Optics sm130, cost around US$20,000 and provide a 1–2 kHz maximum scan rate [6]. OTDR is the newest, most complex and most expensive interrogation technique. It uses a spectrometer with pulsed laser source and high-speed electronics to display the time varying spectrum of the reflected pulse. OTDR interrogators have poorer accuracy due to high-speed electronics and provide about 2 kHz maximum scan rate. The advantage of OTDR, however, is that it allows for
4.3 Fibre Bragg Gratings

multiple gratings of the same centre frequency to be used on a single fibre, whereas the other two techniques require the spectrum of the gratings on a single fibre to not overlap at all. The catch is that OTDR requires the gratings to be spaced far enough apart physically to allow sufficient time to differentiate between reflected light pulses.

4.3.4 Design Choices

For the force/torque sensor described herein, 12 FBGs, each 1 mm long with an 8.5 mm centre-to-centre spacing between the gratings is required on a single optical fibre. The use of a single optical fibre is considered an important advantage of this sensor design since it requires less space in the instrument for optical fibres, and requires the use of only a single channel on the interrogator, which increases measurement speed and allows multiple sensors to be connected to a single interrogator.

The close spacing between the gratings prevents the use of same centre wavelength gratings, thereby negating the advantage of OFDR. Hence, it was decided that the force sensor must use a spectrometry-based interrogator. The specific interrogator selected was a Micron Optics sm130 4-channel 1 kHz interrogator. This is a mid-range interrogator that has sufficient sampling rate and accuracy. It costs around US$20,000 and forms a capital equipment investment required to use the force/torque sensor. The 1 mm length for the grating was selected by assessing the trade-off between the low reflectance of shorter gratings and the strain averaging effect of longer gratings. Shorter gratings reflect less light, thereby making it harder to accurately detect the centre wavelength of the reflected spectrum. Longer gratings will measure the average strain over the entire length, deviating more from the true strain measurement at the point of interest. It was decided that 1 mm is an appropriate length to get a good strain measurement since it is the shortest length that has sufficient reflectance to be detected with good accuracy by the interrogator.
The specific optical fibre selected was a reduced-cladding low bend-loss fibre from FBGS Technologies International GmbH, Germany. This fibre with 12 gratings costs US$600 for small volume production, and forms the single most expensive component of this sensor. This fibre is only 120 $\mu$m in diameter and is the thinnest commercial optical fibre available with inscribed FBGs, which is desirable to decrease the size of grooves required for the fibre. FBGS produces the FBGs using Draw Tower Grating (DTG) technology that retains the full strength of the optical fibre after inscribing the gratings. These fibres also allow for a minimum bend radius of 1.7 mm, a requirement to be able to route the fibre in the sensor geometry. Due to its low bend-loss design, there is no noticeable attenuation in the received signal when the fibre undergoes sharp bends.

One drawback of using these DTG fibres is that there is a restriction on the maximum difference between the centre wavelengths of the adjacent gratings. This is a limitation of the laser interference technique used to inscribe the gratings and can be overcome by using the more expensive phase-mask approach. However, for the design of this sensor, the limitation was taken into account. In this case, based on the 7.5 mm spacing between the gratings, the maximum possible difference between the centre wavelengths is 1.5 nm. This means that the maximum acceptable shift in the wavelength of any FBG in the sensor is ±0.7 nm to avoid spectrum overlap while accounting for the FWHM of 0.1 nm for the reflected spectral peak.

It was decided that the bandwidth of the sensor measurements be set at 15 Hz to provide a balance between the real-time reaction speed and the accuracy of the measurements. The sampling rate of the Micron Optics sm130 interrogator is fixed at 1 kHz. A 15 Hz bandwidth allows for the use of 30-sample moving average filter implemented in hardware on the interrogator. The optical fibre to be used in the force sensor was tested with the interrogator to assess the measurement noise. It was observed that with a 30-sample moving average filter, the wavelength measurement noise was limited to within ±1 pm.
4.4 Mechanical Design

4.4.1 Sensor Geometry

Fig. 4.1 shows the four proposed geometries of the force sensor along with a cross section through the plane where the strain measurements are made. The geometries shown provide for a choice between a 8.5 mm square or a 10 mm circular outer profile depending on the application. There is also a choice to have an axial hole through the sensor that can be up to 7 mm in diameter. The specific geometry used in the robotic palpation tool (Chapter 5) is square with no axial hole. All of the proposed geometries have an identical cross section and hence have the same measurement characteristics.

The material selected for the sensor body is annealed Grade 5 Titanium alloy, also known as Ti6Al4V. This alloy has a combination of high tensile yield strength (880 MPa) and relatively low Young’s Modulus (115 GPa), along with being corrosion resistant. The high strength and low modulus together allow the construction of geometries that maximize the induced strain for a given loading, thereby increasing measurement accuracy.

4.4.2 Fibre Placement

Fig. 4.2(a) shows how the single fibre is wrapped around the entire sensor three times. Each face has three grooves, two for strain sensing FBGs and one for a temperature sensing FBG. The FBGs are labelled 1 to 12 in Fig. 4.2(b), where FBGs 1 to 8 are for strain measurement and 9 to 12 are for temperature measurement. The FBGs are labelled in the order in which they are located on the fibre, with 1 being the first FBG closest to the interrogator and 12 being the last FBG farthest from the interrogator. One of the strain sensing FBG grooves is deeper than the other on each face to allow for the strain sensing FBGs to cross over each other at the centre of the face. The temperature sensing FBGs are placed at
locations where they will experience no mechanical strain. It was decided to use a temperature sensing FBG on each face due to the simplicity of the design and additional accuracy afforded by measuring the exact temperature of each face. The temperature sensing FBG
from a face is used to compensate the measurements from the two strain sensing FBGs on the same face only.

Figure 4.2: Routing of the optical fibre in the sensor geometry: (a) 3D CAD rendering of the entire sensor with the optical fibre, the FBGs are marked as pink on the purple fibre, (b) transparent view of the sensor with the FBGs and the coordinate system labelled.

For gluing down the fibre in the grooves, Loctite 3102 UV-cure adhesive was chosen due to its high strength, high stiffness, short curing time, low viscosity and autoclave resistance. UV-cure adhesives are better than self-cure adhesives in this application because they allow ample time for the fibre to be repositioned but cure within 30 seconds when exposed to UV light once the fibre is ready to be glued down.

4.4.3 Manufacturing

The manufacturing of the sensor geometry is a straightforward process that can be broken down into the following simple steps:

1. Preparation of the outer profile — The first step is to create the outer profile (8.5 mm square or 10 mm circle with four flat faces) of the geometry using the suitable man-
ufacturing process based on the design of the component in which the force sensor geometry is being included. Typically, the process will be conventional milling, wire EDM or injection moulding.

2. *Drilling of the axial hole* — The next step is to drill or mill the optional axial hole that can be of any diameter less than 7 mm. Depending on the manufacturing process, this step may be combined with the previous one.

3. *Drilling of holes for EDM wire* — The next step is to drill four holes, two per face on two adjacent faces, within the interior contour indicated in Fig. 4.3(a). These holes will be where the EDM wire is fed through to cut the interior contour.

4. *Wire EDM cutting of the interior contours* — The next step is cutting the four two-dimensional interior contours shown in Fig. 4.3(a), two per face on two adjacent faces, using wire EDM. Four EDM passes per contour is recommended to obtain a good interior surface finish and high precision. The EDM wire diameter must be less than 0.25 mm to achieve the required minimum radius in the contour.

5. *Wire EDM cutting of the remaining supports* — If a square profile or an axial hole smaller than 2 mm diameter is used, then the geometry will still have some additional load bearing material left at the force sensing plane. This material can be cut using wire EDM by holding the part at a 45° angle from any face as shown in Fig. 4.3(b).

6. *Machining of the optical fibre grooves* — The next step is the cutting of the grooves for the optical fibre on the four faces of the sensor. These grooves are cut by a conventional milling machine using a 30° carbide engraver with a 0.005" tip radius since it is more reliable to use than a tiny endmill.

7. *Gluing the fibre* — The next and the most challenging step of all is gluing the fibre in the grooves using Loctite 3102 UV-cure adhesive, while ensuring that 12 FBGs are positioned correctly. This step of the manufacturing process has not yet been completely developed, and requires some work to make it reliable and repeatable.
4.4 Mechanical Design

The successful completion of this step with good placement accuracy (±0.1 mm) requires the use of a jig to guide the fibre and soft plastic tools to manipulate the fibre without damaging it.

8. Encapsulation — The final step is encapsulating the exposed optical fibre with high-temperature resistant autoclavable silicone rubber, such as Smooth-On Mold Max XLS II, to protect it from mechanical damage. The rubber can be used to fill the entire space between the outer profile and the axial hole. A custom mould that conforms to the shape of the part with the incorporated sensor will have to be manufactured to allow for proper encapsulation. Since the stiffness of the silicone rubber is about 100 times less than titanium, it will have a negligible effect on the measurements.

![Figure 4.3: Wire EDM steps in manufacturing the sensor geometry: (a) cutting of the interior contours, (b) cutting of the remaining supports.](image)

4.4.4 Finite Element Analysis and Failure Loading

Finite element analysis (FEA) was performed on the sensor geometry to ensure that the stress concentrations caused by the grooves for the fibre do not result in mechanical failure of the sensor up to the desired failure loading. FEA was also used in the design process to iteratively optimize the design for obtaining the maximum possible strain measurement, while ensuring that the sensor can withstand the loads subjected to it in the robotic palpation instrument. The analysis was performed using Solidworks and the mesh at the grooves was set to be 0.2 mm using mesh control to ensure that the stress concentrations are properly
represented in the analysis. The stress concentration in the grooves was the limiting factor that prevented the further decrease of the cross sectional area at the strain sensing plane. The grooves are a necessary part of the design since they allow for the proper placement of the fibre and better strain transfer to the FBG due to more adhesion surface area.

To conduct the FEA, the force sensor geometry was transferred onto a $8.5 \times 8.5 \times 100$ mm titanium structure with holes and split lines to serve as locations to apply supports and loads for the analysis. The final design has a factor of safety (FOS) of only 1.05 when a 20 N load is applied at 70 mm from the sensor, which corresponds to the $\pm 1400$ N·mm maximum torque loading mentioned earlier in this chapter. The maximum safe torque loading results are presented in Fig. 4.4(a). This FOS was deemed sufficient since the load criteria already include an FOS of almost 2. The maximum force loading that the sensor can withstand was determined by applying a lateral force at 2 mm from the centre of the sensor, which is the closest location where a force can be applied since the sensor itself is 4 mm tall. The maximum safe force loading was determined to be $\pm 230$ N with an FOS of 1.03. The maximum safe force loading results are presented in Fig. 4.4(b). The maximum force loading for this sensor is very high and therefore the sensor is expected to fail by bending rather than shear.

### 4.5 Mathematical Modelling

#### 4.5.1 Force–Strain Relationship

The approximate geometric parameters of the cross section through the force sensor geometry at the strain measurement plane are given by Eqs. (4.1) to (4.7). The parameters $l$, $t$ and $d$ are physical measurements shown in Fig. 4.5. The actual value of the dimension $l$ is 1.5 mm in the design, but its value is decreased to 1.35 mm in the calculations to account
Figure 4.4: FEA von Mises stress results for the maximum safe loading on the force/torque sensor: (a) maximum torque loading of 1400 N·mm (20 N at 70 mm), (b) maximum force loading of 230 N (at 2 mm).

for the groove cut out for the optical fibre. The parameter $A$ is the total cross sectional area. The parameter $Q$ is the static moment of area of the top half of the cross section about the neutral axis. The parameter $I$ is the second moment of inertia of the total cross sectional area about the neutral axis. The parameter $J$ is the polar second moment of inertia of the total cross sectional area about the axial direction.

\[
l = 1.35 \text{ mm} \quad (4.1)
\]
\[
t = 0.40 \text{ mm} \quad (4.2)
\]
\[
d = 4.0 \text{ mm} \quad (4.3)
\]
\[ A \approx 4lt = 2.2 \text{ mm}^2 \quad (4.4) \]
\[ Q \approx ltd + \frac{l}{2}t^2 = 2.4 \text{ mm}^3 \quad (4.5) \]
\[ I \approx 2ltd^2 + \frac{l^3t}{12} = 17.5 \text{ mm}^4 \quad (4.6) \]
\[ J \approx 4ltd^2 = 34.6 \text{ mm}^4 \quad (4.7) \]

Figure 4.5: Cross section through the force/torque sensor with the important dimensions labelled.

Basic mechanics equations can be used to relate the applied loads to the stresses at the measurement points [7]. The shear stress \( \tau_L \) along the vertical direction, caused by the lateral shear force \( V \) at locations P and Q marked on the sensor in Fig. 4.6(a), is given by Eq. (4.8). This shear stress \( \tau_L \) is downwards at both P and Q which is considered positive at P and negative at Q. The normal stress \( \sigma_B \) along the horizontal direction, caused by the bending moment \( M \) at locations R and S marked on the sensor in Fig. 4.6(b), is given by Eq. (4.9). This normal stress \( \sigma_B \) is positive at R and negative at S. The normal stress \( \sigma_N \)
along the axial direction, caused by the axial force $N$ at locations P, Q, R and S marked on the sensor in Fig. 4.6, is given by Eq. (4.10). This normal stress $\sigma_N$ is positive at P, Q, R and S. The shear stress $\tau_T$ along the counter-clockwise direction, caused by the torsional moment $T$ at locations P, Q, R and S marked on the sensor in Fig. 4.6(a), is given by Eq. (4.11). This shear stress $\tau_N$ is positive at P, Q, R and S.

$$\tau_L = \frac{VQ}{I_b} = \frac{VQ}{I_{2t}} = \frac{V}{5.8 \text{ mm}^2}$$

(4.8)

$$\sigma_B = \frac{My}{I} = \frac{M(d + t/2)}{I} = \frac{M}{4.2 \text{ mm}^3}$$

(4.9)

$$\sigma_N = \frac{N}{A} = \frac{2.2 \text{ mm}^2}{2.2 \text{ mm}^2}$$

(4.10)

$$\tau_T = \frac{Tr}{J} = \frac{T(d + t/2)}{J} = \frac{T}{8.2 \text{ mm}^3}$$

(4.11)

Figure 4.6: Cross section of the sensor through the sensing plane indicating the internal forces and torques, and the strain measurement points: (a) front view, (b) side view.
At each of the locations, P, Q, R and S, there are two strain sensing FBGs positioned 45° from the axial direction. FBGs 1 to 4 are tilted by 45° in the counter-clockwise direction and correlate positively with positive shear stress, whereas FBGs 5 to 8 are tilted by 45° in the clockwise direction and correlate positively with negative shear stress (Fig. 4.2(b)). Based on the plane stress transformation equation, the normal stress $\sigma$ experienced by FBGs 1 to 8 are given by Eqs. (4.12) to (4.14), where $\sigma_{\text{net}}$ is the net normal stress (combination of $\sigma_B$ and $\sigma_N$ with appropriate signs) and $\tau_{\text{net}}$ is the net shear stress (combination of $\tau_L$ and $\tau_T$ with appropriate signs).

$$\sigma = \sigma_{\text{net}} \cos^2 \theta + 2\tau_{\text{net}} \cos \theta \sin \theta$$

(4.12)

$$\Rightarrow \sigma_{(1-4)} = \frac{\sigma_{\text{net}}}{2} + \tau_{\text{net}} \quad \text{because} \quad \theta = +45^\circ$$

(4.13)

$$\Rightarrow \sigma_{(5-8)} = \frac{\sigma_{\text{net}}}{2} - \tau_{\text{net}} \quad \text{because} \quad \theta = -45^\circ$$

(4.14)

Based on the above equations, the matrix Eq. (4.15) can be constructed to relate the normal stresses $\sigma_1$ to $\sigma_8$ along the 8 strain sensing FBGs to the normal and shear stresses from the different loading conditions. The subscripts X, Y or Z indicates the axis along which the force, or about which the moment, is applied (Fig. 4.2(b)). This equation can be rewritten as Eq. (4.16) in which $\sigma_1$ to $\sigma_8$ is replaced by the strain at the FBGs, $\epsilon_1$ to $\epsilon_8$, multiplied by 115 GPa, the Young’s modulus of titanium. The equation also has the normal and shear stresses expressed in terms of the forces and moments that cause them. This equation can then be rewritten as Eq. (4.17) that establishes the final relationship between the measured strains and the applied 6-DOF loads. The 8-by-6 numerical matrix that relates the strains and the loads is known as the Compliance Matrix ($C$), which is specific to this force sensor, where the strains are in microstrain [$\mu$], the forces are in Newtons [N] and the moments/torques are in Newton-millimetres [N·mm]. Eq. (4.17) can be rewritten in
symbolic form as Eq. (4.18). This equation can be rearranged as Eq. (4.19) to obtain the loads \( f \) as a function of the measured strains \( \epsilon \), where \( C^+ \) is the left pseudo-inverse of \( C \), since \( C \) is not a square matrix. Eq. (4.19) gives a least-squares solution for \( f \) since this is an overdetermined system where 8 strain measurements are used to calculate only 6-DOF loads. For this force sensor, the compliance matrix and its left pseudo-inverse are full-rank constant matrices, and the sensor has a closed-form force–strain relationship.

\[
\begin{bmatrix}
\sigma_1 \\
\sigma_2 \\
\sigma_3 \\
\sigma_4 \\
\sigma_5 \\
\sigma_6 \\
\sigma_7 \\
\sigma_8
\end{bmatrix} =
\begin{bmatrix}
0 & -1 & 0.5 & 0 & 0.5 & 1 \\
1 & 0 & 0.5 & -0.5 & 0 & 1 \\
0 & 1 & 0.5 & 0 & -0.5 & 1 \\
-1 & 0 & 0.5 & 0.5 & 0 & 1 \\
0 & 1 & 0.5 & 0 & 0.5 & -1 \\
-1 & 0 & 0.5 & -0.5 & 0 & -1 \\
0 & -1 & 0.5 & 0 & -0.5 & -1 \\
1 & 0 & 0.5 & 0.5 & 0 & -1
\end{bmatrix}
\begin{bmatrix}
\tau_{L-X} \\
\tau_{L-Y} \\
\sigma_{N-Z} \\
\sigma_{B-X} \\
\sigma_{B-Y} \\
\tau_{T-Z}
\end{bmatrix}
\]  
(4.15)

\[
\begin{bmatrix}
\epsilon_1 \\
\epsilon_2 \\
\epsilon_3 \\
\epsilon_4 \\
\epsilon_5 \\
\epsilon_6 \\
\epsilon_7 \\
\epsilon_8
\end{bmatrix} =
\begin{bmatrix}
0 & -1 & 0.5 & 0 & 0.5 & 1 \\
1 & 0 & 0.5 & -0.5 & 0 & 1 \\
0 & 1 & 0.5 & 0 & -0.5 & 1 \\
-1 & 0 & 0.5 & 0.5 & 0 & 1 \\
0 & 1 & 0.5 & 0 & 0.5 & -1 \\
-1 & 0 & 0.5 & -0.5 & 0 & -1 \\
0 & -1 & 0.5 & 0 & -0.5 & -1 \\
1 & 0 & 0.5 & 0.5 & 0 & -1
\end{bmatrix}
\begin{bmatrix}
V_x/5.8 \text{ mm}^2 \\
V_y/5.8 \text{ mm}^2 \\
N_z/2.2 \text{ mm}^2 \\
M_x/4.2 \text{ mm}^3 \\
M_y/4.2 \text{ mm}^3 \\
T_z/8.2 \text{ mm}^3
\end{bmatrix}
\]  
(4.16)

(115 GPa)
4.5 Mathematical Modelling

\[
\begin{bmatrix}
\varepsilon_1 [\mu] \\
\varepsilon_2 [\mu] \\
\varepsilon_3 [\mu] \\
\varepsilon_4 [\mu] \\
\varepsilon_5 [\mu] \\
\varepsilon_6 [\mu] \\
\varepsilon_7 [\mu] \\
\varepsilon_8 [\mu]
\end{bmatrix}
= 
\begin{bmatrix}
0 & -1.5 & 2.0 & 0 & 1.04 & 1.06 \\
1.5 & 0 & 2.0 & -1.04 & 0 & 1.06 \\
0 & 1.5 & 2.0 & 0 & -1.04 & 1.06 \\
-1.5 & 0 & 2.0 & 1.04 & 0 & 1.06 \\
0 & 1.5 & 2.0 & 0 & 1.04 & -1.06 \\
-1.5 & 0 & 2.0 & -1.04 & 0 & -1.06 \\
0 & -1.5 & 2.0 & 0 & -1.04 & -1.06 \\
1.5 & 0 & 2.0 & 1.04 & 0 & -1.06 \\
\end{bmatrix}
\begin{bmatrix}
F_X [N] \\
F_Y [N] \\
F_Z [N] \\
T_X [N \cdot \text{mm}] \\
T_Y [N \cdot \text{mm}] \\
T_Z [N \cdot \text{mm}]
\end{bmatrix}
\tag{4.17}
\]

\[
\Rightarrow \varepsilon = Cf
\tag{4.18}
\]

\[
f \approx (C^T C)^{-1}C^T \varepsilon = C^+ \varepsilon
\tag{4.19}
\]

4.5.2 Calibration and Temperature Compensation

The relationship between the peak reflected wavelength \( \lambda \) and the strain \( \varepsilon \) experienced by the FBG is given by Eq. (4.20) where \( \lambda_0 \) is the nominal peak wavelength under no strain, and \( K \) is the strain sensitivity of the fibre which is quoted as 0.78 in the datasheet from FBGS Technologies GmbH. The value of \( \lambda_0 \) for each FBG is measured during calibration after the fibre is glued onto the sensor. The actual value of the compliance matrix \( C \) will also have to be determined experimentally during calibration to account for any errors in positioning the FBG and obtain best accuracy. The theoretical compliance matrix can be used as a starting point for the calibration algorithm to ensure that the solution converges.

\[
\varepsilon = \frac{\lambda - \lambda_0}{\lambda_0 K}
\tag{4.20}
\]

The source of the strain \( \varepsilon \) can be the mechanical or thermal expansion of the material.
that the FBG is glued onto. For the temperature sensing FBGs 9 to 12, the only source of strain is thermal. It is assumed that the two strain sensing FBGs on the same face as a temperature sensing FBG will experience a sum of the thermal strain and the mechanical strain. Therefore, to isolate the mechanical strain on FBGs 1 to 8, the strain measurement of the temperature sensing FBG on the same face is simply subtracted from the strain measurements of the strain sensing FBGs.

Based on Eq. (4.20), the accuracy and range of strain measurements can also be calculated. It was reported earlier in this chapter that the experimental peak wavelength measurement accuracy of the Micron Optics sm130 interrogator with the chosen FBGS fibre is ±1 pm and the maximum allowable shift in the peak wavelength of an FBG in the sensor is ±0.7 nm. Assuming a nominal value of 1550 nm for $\lambda_0$ and 0.78 for $K$, the accuracy and range values correspond to ±0.83 $\mu$ and ±580 $\mu$ strain, respectively.

### 4.5.3 Theoretical Worst Accuracy and Measurement Range

The condition numbers of the compliance matrix $C$ and its left pseudo-inverse $C^+$ were both calculated to be 2.7. Ideally, the compliance matrix should have a condition number of 1, which would mean that any noise in the strain measurements will have the minimum possible effect on the calculated loads. Since the condition number is larger than 1, a Simulink model was used to determine the effects of measurement noise on this force/torque sensor. The model computed the strains for several different combinations of forces and torques applied on the sensor spanning the entire safe-loading range. The model then applied random white noise in the range of ±0.83 $\mu$ to the strain measurements and recalculated the forces and torques based on the noisy strain measurements. Using this approach, it was determined that the maximum inaccuracy in force measurements was limited to within ±0.5 N and the maximum inaccuracy in torque measurements was limited to within ±0.7 N·mm.
This accuracy rating can be significantly improved by using a more sensitive interrogator, better FBGs, better digital filters and lower measurement bandwidth.

The force and torque measurement range of the sensor can be determined by using the ±580 μ measurable strain range with Eq. (4.17). From the compliance matrix it can be clearly seen that the largest factors relating the 3D force in [N] and the 3D torque in [N-mm] to the strain in [μ] is 2.0 and 1.06 respectively. This implies that the theoretical maximum 3D force and 3D torque measurement range of this sensor is ±290 N and ±550 N-mm respectively. However, based on FEA, the maximum safe force loading is only ±230 N, which is the limiting case for the 3D force range. For the robotic palpation instrument discussed in the next chapter, the ±550 N-mm torque measurement range translates to a maximum measurable palpation force of 14 N, since the centroid of the palpation surface of the tactile sensor is 40 mm from the centre of the force/torque sensor. Since the maximum allowable safe palpation force discussed in Chapter 3 for the large tactile sensor is 14 N and since the maximum palpation force that the robotic palpation instrument can apply is also 14 N, the measurement range of the force/torque sensor is just right for this application. In this case, the sensor has an overload factor of safety of 2.5, which adds an adequate safety buffer to avoid mechanical failure under normal use.

4.6 Prototype

To allow for easy mounting and easy application of known weights for calibration, validation and evaluation, it was decided that the structure with holes created for FEA should be manufactured as the first prototype. The prototype was manufactured out of 17-4PH stainless steel instead of Grade 5 Titanium since the former is much less expensive but has similar hardness as titanium and hence can be used to fine-tune the manufacturing process, especially cutting the narrow grooves for the fibre. The prototype of the sensor geometry
4.7 Conclusions

along with the grooves is shown in Fig. 4.7, and this validates that the sensor geometry can be manufactured by using the manufacturing process described in this chapter. The figure also shows the carbide engraving tool used to cut the grooves. Unfortunately, due to time constraints, the necessary jigs and tools for accurate fibre placement have not yet been designed and manufactured, which resulted in the sensor prototype being incomplete.

Figure 4.7: The force/torque sensor prototype constructed so far, and the carbide engraving tool used to cut the grooves (insets show close-up views of the fibre grooves and the engraving cutter).

4.7 Conclusions

This chapter has presented the design and analysis of a novel 6-DOF force/torque sensor with temperature compensation developed specifically for minimally invasive surgical instruments. The design specifications of the sensor and the design choices made during its development have been discussed in detail. The mechanical structure of the sensor has been designed to be easily manufactured which significantly reduces the cost of this sensor. A complete mathematical model has also been developed and has been used to determine the theoretical accuracy and measurement range of the sensor. Experimental validation of the mathematical model was not performed since the prototype of the sensor is incomplete. Unanticipated difficulties were encountered in accurately positioning and gluing the optical fibre on the sensor prototype by hand with tweezers and other standard tools. It was established that special jigs and tools are required to accomplish this, which requires
further design and manufacturing. Nevertheless, the theoretical analysis and the partially completed prototype demonstrate that this novel sensor has several advantages over other designs in the literature, and hence it is worthwhile to pursue the further development and evaluation of this sensor.

References


Chapter 5

Robotic Palpation Instrument

5.1 Introduction

This chapter presents the design, analysis and evaluation of a novel robotic palpation instrument that is designed to be compatible with the da Vinci surgical robot. The instrument incorporates the tactile sensor presented in Chapter 3 and the force/torque sensor presented in Chapter 4 along with an ultrasound transducer in its end effector. The motivation behind the development of this instrument is to enable the use of multiple modalities to achieve higher tumour localization accuracy in minimally invasive robotic surgery. The data from the three sensors can be fused to provide tactile, kinaesthetic and ultrasound feedback to the surgeon. The design of the instrument is inspired by the commercial da Vinci surgical instruments that have been proven to work well. The instrument has three serial cable-driven internal DOFs that have been designed to be decoupled to improve control by minimizing the effects of cable stretch. This chapter reports the major design specifications of the instrument, describes its mechanical and control system design, validates the design with appropriate analysis, and discusses the evaluation of the prototype.
5.2 Design Specifications

The basic design requirements that were developed to guide the design of the robotic palpation instrument are listed below:

1. Must be compatible with the da Vinci surgical robot
2. Must fit through a 12 mm diameter or smaller trocar
3. Must be able to provide tactile, kinaesthetic and ultrasound feedback
4. Must be able to withstand the maximum forces that the da Vinci robot can apply
5. Must have sufficient degrees of freedom and range of motion to comfortably palpate tissue in any orientation
6. Must allow for the disposable tactile sensor to be easily replaced
7. Must be biocompatible, cleanable and sterilizable

The instrument was designed to be compatible with the da Vinci surgical system because it is currently the only surgical robot that is sufficiently advanced and robust enough to be able to obtain the clearance to operate on humans. Developing an instrument that works with the da Vinci will have the best possibility of being used in clinical trials in the shortest time span. Furthermore, Intuitive Surgical and John Hopkins University have collaborated to develop the da Vinci Research Kit (dVRK) platform with the specific purpose of allowing researchers to develop and test custom instruments and algorithms with the da Vinci system. The dVRK is more user-friendly and stable in terms of control software, as compared to other surgical robot platforms, thereby making it simpler to test new instruments with it. The actuators, position sensors, control electronics and software are all a part of the da Vinci system, reducing the majority of the design process to mechanical design and analysis.

The da Vinci surgical robot allows for up to four actuated DOFs on the tool, along with the three external DOFs provided by the robot arm: two for pivoting about the re-
mote centre of motion (RCM) along two orthogonal axes, and one for linear motion of the instrument in and out of the body. Based on the typical construction of existing da Vinci EndoWrist instruments, it was determined that three DOFs on the tool (six DOFs in total) will provide the necessary orientation flexibility for the end effector. For simplicity, the roll DOF is implemented on the part of the instrument outside the body, and the pitch and yaw DOFs are implemented at the instrument wrist within the body. Inspired by the EndoWrist instruments, it was established that a range of motion of $\pm 90^\circ$ is desired for the pitch and yaw motion, while $\pm 270^\circ$ is desired for the roll motion [1]. This range allows the instrument to move through the entire range of motion that the surgeon’s hand can move through in the da Vinci master console. Cables were selected as the mode of transmitting motion because this technique is implemented on the majority of Endowrist da Vinci tools and have been proven to work reliably. Using cables allows for isotropic motion and much simpler construction than using other techniques such as push/pull rods.

Another major decision was the choice of a one-sided palpation design rather than a grasping design for the instrument. The major drawback of using a one-sided palpation design is that there is no fixed rigid surface to support the tissue from the bottom when palpating it from the top. This, however, is a problem that can be solved easily by either palpating the deflated lung against the chest wall or by using another instrument to provide a rigid backing. The reason for choosing a palpation design is to avoid the excessive problems and risks associated with a grasping design. The first problem is that the da Vinci system allows for a maximum of four actuated DOFs on the tool, which is not sufficient to provide two DOFs (the distance and the angle between the jaws) for grasping varying tissue geometries along with three DOFs (roll, pitch and yaw) for the articulated wrist. The three DOF articulated wrist is a key feature of a robotic instrument that cannot be avoided since it is essential to allow the instrument to mimic the natural hand motion of the surgeon. The second problem is that the design and manufacture of a grasping instrument with an
articulated wrist that can withstand the forces involved and fit through a 12 mm port is too complicated to be desirable in a medical device. Lastly, there is a significant risk of damaging the lung tissue when using a grasping design due to the requirement of lifting and placing the lung in the jaws, and due to the chance that the instrument can accidentally pinch or pull on the tissue while grasping it. Also, the length of the jaws required to be able to palpate the entire width of the lung (about 9 cm) will make the end effector too long to safely articulate within the chest cavity. These various reasons justify the selection of a one-sided palpating design rather than a grasping design for the instrument.

It was determined that the instrument would require at least a 12 mm trocar to be able to accommodate a tactile sensor and an ultrasound transducer back-to-back. The design of the end effector of the instrument was constrained partly by the tactile sensor and mostly by the ultrasound module that was available to construct the prototype of the instrument with. A standard 12 mm trocar actually has a 12.7 mm internal diameter, and the final design of the instrument barely fits through it. The da Vinci system, however, uses its own special 8 mm trocar that attaches to the robot arm. Therefore, a custom trocar was designed to use the instrument with the da Vinci robot. The prototype of the custom trocar uses a standard stainless steel tube to simplify manufacturing, which has an internal diameter of 13.2 mm.

The da Vinci robot is able to apply a maximum torque of 3 N·m to pivot about the RCM. This limit is enforced by setting a current limit on the drive motors for the two pivot DOFs. It is assumed that the instrument tip is typically about 15 cm from the RCM when palpating the lung. This implies that the instrument must be able to withstand a 20 N transverse force at the tip. Also, the maximum cable drive torque is 0.52 N·m, and because the minimum radius of the cable drive pulleys in the instrument is 2.5 mm, the drive cables must withstand a maximum tension force of 210 N in addition to the pre-loading added to ensure that the cables are taut at all times. It is estimated that 20 N preload is sufficient, bringing the maximum cable tension to 230 N. In the final design, the distance to the centroid of the
palpation surface from the pitch DOF is 60 mm, and since the pitch DOF pulley has a 4 mm radius, the maximum palpation force that can be achieved is 14 N, which matches the maximum safe palpation force for the large size tactile sensor. The specifications of the final instrument design are listed in Table 5.1.

Table 5.1: Major design specifications of the novel robotic palpation instrument.

<table>
<thead>
<tr>
<th>Design Specifications</th>
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<tbody>
<tr>
<td>Compatibility</td>
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<tr>
<td>End Effector Design</td>
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<tr>
<td>End Effector Outer Diameter</td>
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<tr>
<td>End Effector DOFs (in sequence)</td>
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<tr>
<td>End Effector Actuation</td>
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<tr>
<td>Maximum Transverse Load</td>
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<tr>
<td>Maximum Cable Tension</td>
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<td>Maximum Palpation Force</td>
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<td>Sterilizability</td>
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5.3 Mechanical Design

5.3.1 Overview

Fig. 5.1 shows a CAD model of the entire instrument. To make it easier to understand the design of the instrument, the complete assembly is broken into four subassemblies (Fig. 5.2): the End Effector, the Wrist, the Instrument Shaft and the da Vinci Attachment. Each of these subassemblies is discussed in detail in the next four subsections. The three internal DOFs of the instrument are shown in Fig. 5.3. The pitch DOF is perpendicular to the sensor surface, while the yaw DOF is parallel to the sensor surface. The instrument has been designed with simplicity, robustness, easy manufacturing and low cost in mind. The intricate parts of the instrument, particularly in the wrist, have been designed for simple manufacture using wire EDM followed by the drilling of a few holes. The pulleys in the tool can be easily manufactured using a rotating spindle with wire EDM cutting, followed by drilling the central hole. There are only a few large components that require the use of conventional milling to cut some pockets. The design of the tool also allows for easy assembly and disassembly, a feature required for an instrument meant for testing and evaluation rather than production. If the instrument is eventually modified for production, several fasteners can be easily replaced by snap fits and press fits, and the aluminium da Vinci attachment can be made out of plastic instead, just like the commercial da Vinci instruments.

Figure 5.1: CAD model of the entire robotic palpation instrument.
5.3 Mechanical Design

(a) (b) (c) (d)

Figure 5.2: The four subassemblies of the robotic palpation instrument: (a) end effector, (b) wrist, (c) instrument shaft, (d) da Vinci attachment.

(a) (b)

Roll: ±270º
Pitch: ±90º
Yaw: ±90º

Figure 5.3: The three degrees of freedom in the robotic palpation instrument: (a) da Vinci attachment, (b) wrist.

5.3.2 End Effector

The end effector subassembly (Fig. 5.4) consists of the ultrasound transducer module and the tactile sensor mounted on a sensor holder frame that attaches to the sensor holder
5.3 Mechanical Design

bracket in the wrist subassembly using four screws. The tactile sensor interfaces to the tool via four spring contacts, and is held in place by a securing clip. This setup allows the tactile sensor to be easily replaced between uses. The geometry of the ultrasound transducer module was predetermined since the transducer available for use with the instrument was already mounted in a stainless steel frame. The ultrasound transducer is sterilizable along with the entire instrument using low-temperature sterilization processes such as ethylene oxide gas and hydrogen peroxide gas sterilization used in the healthcare industry for laparoscopic ultrasound probes [2].

Figure 5.4: Labelled exploded view of the end effector subassembly.

5.3.3 Wrist

The wrist subassembly (Fig. 5.5) is comprised of three brackets and the associated shafts, pulleys and cables that provide the pitch and yaw degrees of freedom. The sensor holder bracket is the most distally located and includes the force/torque sensor. This location was chosen for the force/torque sensor because it is not affected by drive cable tension,
allowing it to measure the tip and palpation forces with the greatest degree of accuracy. Three sets of guide pulleys are used to route the four drive cables to the centre of the instrument shaft and make space for the sensor wiring. The arrangement of these guide pulleys were iterated upon several times to optimize the drive cable paths. The final configuration of pulleys provides decoupled motion between the pitch and yaw DOFs. This configuration makes the control simpler and more robust, and also minimizes the orientation error due to cable stretch. The arrangement of the Yaw Cable Guide 1 pulleys ensures that there is minimal variation in the length of the yaw cables as the pitch DOF moves, but since the pulleys...
have a finite radius of 2 mm, there is 0.2 mm of slack that develops when the pitch DOF is at $\pm 90^\circ$. This slack is absorbed by the pre-tension in the cables, but the reduced pre-tension results in a greater cable stretch error under load. Fortunately, due to the design, the yaw DOF is parallel to the sensor surface and does not take any loading during normal use, and hence this reduced pre-tension does not result in any significant orientation error.

![Figure 5.6: Orthogonal cross sections of the wrist subassembly showing the drive cables and wiring.](image)

There are two bundles of wiring, one for the ultrasound module and one for the tactile sensor. The latter also includes the fibre optic cable for the force/torque sensor. Fig. 5.6 shows the cable and wiring layout in orthogonal cross section views. A big challenge was routing the wiring in a manner that allows for free pitch and yaw rotation. The chosen approach was to secure the wiring with a tie down between the pitch and yaw axes. Sufficient slack is maintained before the tie down to allow the wiring to bend sideways for the yaw motion. To enable the pitch motion, the wiring is kept taut using springs in the da Vinci attachment subassembly, that allow the wiring to be pulled and released as the pitch rotation angle changes. The use of springs ensures that loose wiring does not cause entanglement, and allows the instrument to be easily inserted into the trocar.
5.3.4 Instrument Shaft

The instrument shaft subassembly consists of a standard stainless steel seamless tube with an outer diameter of 12.7 mm and inner diameter of 10.9 mm, and some small components to attach the shaft to the wrist and the da Vinci attachment subassemblies (Fig. 5.7). On the distal end, a pitch bracket clamp that is split into four sections, and a pitch bracket locking pin are used to secure the wrist subassembly in place, as can be seen in Fig. 5.8. The two larger clamp sections form a tight fit inside the instrument shaft, while the two smaller sections form a stop to prevent the pitch bracket from sliding further into the instrument shaft due to tension in the drive cables. The clamp sections have cut-outs for the yaw drive cables and the sensor wiring. The clamp is made in four sections to enable easy manufacturing using EDM, and the sections are glued on the pitch bracket using medical grade epoxy such as Loctite M31-CL. The 1.6 mm diameter locking pin is used to prevent the rotation of the pitch bracket with respect to the instrument shaft.

On the proximal end, a roll pulley that is split into two sections, and a retaining washer are used to secure the instrument shaft to the da Vinci attachment subassembly while still allowing for the roll motion. The retaining washer is clamped between the roll pulley and instrument shaft in this subassembly, and between the two instrument shaft bearings in the da Vinci attachment subassembly, thereby preventing the shaft from sliding. The roll pulley is split longitudinally so that it can be assembled around the ultrasound transducer wiring, since the available transducer has components on either end of the wiring that are too large to fit through the axial hole in the pulley without splitting it. The split pieces are held together by threading into the instrument shaft and are secured in place using two screws.

It was noted that due to the nature of the roll DOF, its rotation would result in the drive cables and the wiring twisting together, which raised the concern of damage to the wiring due to abrasion from the drive cables. The twisting happens in almost all da Vinci tools.
but it is not a problem since they do not have electrical wiring among the drive cables. However, it was discovered in the prototype that there is no visible damage caused to the wiring due to this twisting since the motion of the drive cables is very short and slow during normal use, and since the majority of the length of the drive cables in contact with the wiring is reinforced with smooth stainless steel tubes as described earlier.

Figure 5.7: Labelled exploded view of the instrument shaft subassembly.

Figure 5.8: Close-up of the end effector and wrist showing the force/torque sensor and attachment to the instrument shaft.
5.3.5 da Vinci Attachment

The da Vinci attachment subassembly (Fig. 5.9) has the necessary geometry and components to interface with the da Vinci robot. The actuation motion for the three internal DOFs are transmitted through three of four drive interface plates. The two instrument shaft bearings enable the roll DOF while supporting the instrument shaft. The bearings are held in place by a two-piece clamp that is secured to the base frame using four screws. The two cable guides and associated pulleys route the drive cables to the drive pulleys, while redirecting the wiring away from moving components (Fig. 5.10). The drive pulleys and shafts are supported by bearings on the base frame and the top frame. The top frame and cable guides are also secured to base frame using four screws. A plastic PCB holder is attached to the rear end of the frame using two screws. The PCB holder has the instrument identification PCB on the bottom and the tactile sensor to USB interface PCB on the top. The instrument identification PCB has spring contacts that mate with contacts on the da Vinci robot to let the software know that an instrument has been inserted. These PCBs can be encapsulated in epoxy to protect the electronics during sterilization, as it is done in the existing da Vinci instruments. The PCB holder also has two holes to serve as attachment points for the sensor wiring tension springs (not shown in CAD) that were described earlier.

5.3.6 Material and Components Selection

The selection of different materials used in the instrument was made based on the trade-off between required strength and cost in terms of both the materials and manufacturing methods. All of the materials chosen are corrosion resistant and sterilizable. The custom-manufactured components in the end effector and the wrist, and the instrument shaft retaining washer, are made of annealed 17-4PH stainless steel (750 MPa yield strength) with the exception of the sensor holder bracket that is made of Grade 5 titanium (880 MPa yield
strength) since it incorporates the force/torque sensor. The custom-manufactured components in the da Vinci attachment are made of 6061-T6 aluminium (270 MPa yield strength) due to its easy availability and its optimal strength-to-weight ratio for this application. The exception is the PCB holder that is made of Delrin (63 MPa yield strength), an autoclavable plastic also known as acetal homopolymer, because this component is not structural and is not subjected to significant loads. All of the pins used in the instrument are standard sizes that are made of 416 stainless steel (580 MPa yield strength). This alloy was chosen since it is used to make commercial high-tolerance stainless steel pins in the required sizes while meeting the strength requirement. The bearings chosen are made of hardened 440C stainless steel (1600 MPa yield strength) and are unshielded. These bearings can be sterilized and used without any lubrication since they do not experience continuous high speed mo-
Figure 5.10: Cross section of the da Vinci attachment subassembly showing the drive cables and wiring.

The bearings have been chosen such that there is at least an FOS of 2 between the rated dynamic load of the bearing and the absolute maximum expected loading that they will experience. The instrument shaft tube, the roll pulley and all of the fasteners used in the instrument are made of 304 Stainless Steel (290 MPa yield strength). This alloy was chosen because it is the least expensive option for standard corrosion resistant tubes and fasteners that still meets the strength requirements. The electrical wires have a chemical-resistant FEP insulation that can withstand gas sterilization. The drive cables are 0.6 mm diameter $8 \times 19$ format stranded tungsten alloy cables that have an ultimate tensile strength of 490 N, giving a factor of safety of 2.1. These cables were selected due to their low stretch and small bending radius allowance that enables them to bend around the 3 mm diameter pulleys in the instrument. To minimize the tip deflection due to cable stretch, the straight section of the cables within the instrument shaft are reinforced with crimped stainless steel tubes, as is done in most da Vinci instruments.
5.4 Analysis

5.4.1 Finite Element Analysis

To ensure a robust mechanical design, Solidworks FEA was performed on all of the critical components of the instrument under the worst case loading scenario, which is 20 N applied laterally at the tip and 230 N tension on both pitch and yaw drive cables, with no support from the trocar at the RCM. The pulleys, pins, and non-structural components were removed from the analysis because they over-complicate the model and yield incorrect results. The pins have been analysed independently to ensure that they meet the required shear strength. To perform the FEA, the instrument was divided into two parts, one consisting of the da Vinci interface and the instrument shaft, and the other consisting of the wrist and the end effector. Since both of the parts have components made out of different materials, factor of safety (FOS) results have also been included along with the von Mises stress results, since the former accounts for the yield strength of the actual material that each component is made out of. To analyze the da Vinci interface and the instrument shaft, a simplified version of the end effector was used, as can be seen in Figs. 5.11 and 5.12. The component contacts and connections in the FEA model have been selected to represent the real scenario as closely as possible.

Figs. 5.11 and 5.12 show the stress and FOS results, respectively, for the worst case loading scenario on the instrument shaft and the da Vinci interface. The 460 N axial force represents the maximum combined tension of the drive cables. The results indicate an FOS of 2.6 which is considered adequate. Figs. 5.13 and 5.14 show the stress and FOS results for two worst case loading scenarios on the wrist and the end effector. High stresses were seen at the holes for the pins in the wrist subassembly, and on the thin walls of the sensor holder in the end effector subassembly. The results indicate an FOS of 2 in both cases which is sufficient since the loading is about twice what is expected under normal use.
5.4 Analysis

Figure 5.11: FEA von Mises stress results for the worst case loading on the instrument shaft and da Vinci attachment.

5.4.2 Pin Analysis

There are five sets of critically loaded pins used in wrist subassembly of this design that are analyzed to ensure sufficient strength. The loading conditions and the FOS associated with each set of pins is discussed below in detail. The pins used in the da Vinci attachment subassembly are well over-designed with respect to the pins being discussed in this section. All of the pins are made of 416 stainless steel with the exception of the tool bracket attachment screws, which are included here since they are loaded in shear like the pins. A minimum FOS of only 2 was established as a requirement for these pins since the loading criteria themselves include an FOS of about 2.

Pitch and Yaw Shafts: These pins are 2.4 mm in diameter with a quoted double-shear yield strength of 3300 N. The pitch shaft has to withstand a greater loading than the yaw
5.4 Analysis

Figure 5.12: FEA factor of safety results for the worst case loading on the instrument shaft and da Vinci attachment.

shaft since it has to support the drive cable tension of both the joints and is farther away from the instrument tip resulting in a greater bending moment load. The maximum loading condition is when the drive cables are under the maximum tension of 230 N, while there is a 20 N lateral force being applied on the instrument tip, 83 mm from the pitch joint. The lateral force creates a moment of 1660 N·mm at the pin joint. Since the distance between the two shear interfaces is 5 mm, the interfaces experience a shear force of 330 N in opposite directions. Including the cable tension, the maximum shear force that the pin must withstand is 790 N. This translates to a double-shear load of 1580 N in the worst case scenario, which results in an adequate factor of safety of 2.1.

Pitch Cable Guide Shafts: These pins are 1.6 mm in diameter with a quoted double-shear yield strength of 1500 N. These pins support the Pitch Cable Guide pulleys, each
5.4 Analysis

Figure 5.13: FEA von Mises stress and factor of safety results for the worst case loading on the end effector and wrist in the extended position.

of which changes the direction of a pitch drive cable by 90°. Since the maximum cable tension is 230 N, the maximum force exerted on the pins is 325 N. Since the cable is routed much closer to one shear interface than the other, a conservative simplification is made that the entire load is subjected to a single shear interface. This translates to a double-shear load of 650 N in the worst case scenario, which results in an adequate factor of safety of 2.3.

_Yaw Cable Guide 1 Shafts:_ These pins are 1.6 mm in diameter with a quoted double-shear yield strength of 1500 N. These pins support the Yaw Cable Guide 1 pulleys, each of which changes the direction of a yaw drive cable by a maximum of 90° when the pitch joint is at the extremes of its range of motion. Since the maximum cable tension is 230 N, the maximum force exerted on the pins is 325 N. These pins have a fixed support on only
5.4 Analysis

Figure 5.14: FEA von Mises stress and factor of safety results for the worst case loading on the end effector and wrist in a fully bent position.

one side, thus the entire load is subjected to a single shear interface. This translates to a double-shear load of 650 N in the worst case scenario, which results in an adequate factor of safety of 2.3.

**Yaw Cable Guide 2 Shafts:** These pins are 1.2 mm in diameter with a quoted double-shear yield strength of 830 N. These pins support the Yaw Cable Guide 2 pulleys, each of which changes the direction of a yaw drive cable by 22°. Since the maximum cable tension is 230 N, the maximum force exerted on the pins is 125 N. These pins have a fixed support on both sides, thus the pins are loaded in double-shear. This results in a factor of safety of 6.6, which is considered more than adequate.
5.5 Control System

**Tool Bracket Attachment Screws:** These M1.2 × 0.25 mm screws are made of 304 stainless steel with a quoted single-shear yield strength of 250 N. The worst case loading scenario for these screws is 20 N applied laterally at the tip of the instrument, 60 mm from the screws. This lateral force creates a moment of 1200 N·mm at the screws. Since the distance between the two shear interfaces is 5.6 mm, the interfaces experience a shear force of 214 N in opposite directions. However, since there are two screws per shear interface, this translates to a single-shear load of 107 N per screw in the worst case scenario, which results in an adequate factor of safety of 2.3.

5.5 Control System

The control system used to drive the instrument is integrated into the da Vinci Research Kit (dVRK) system software package. The software for this system has been developed to allow for easy modification of the control system parameters to enable the use of custom-developed instruments, such as the one discussed in this chapter. There are three sets of parameters that must be provided to the dVRK software for a custom instrument, each of which is described in the subsections below.

5.5.1 Modified DH Parameters (Kinematics)

The dVRK software requires the kinematics of the instrument to be provided using modified DH parameters. These parameters follow the Modified Denavit–Hartenberg (DH) convention [3] to describe the kinematics of the instrument. This instrument, combined with the da Vinci robot, forms a 6-DOF serial-link manipulator with five revolute joints and one prismatic joint. In the modified DH convention, each link \( i \) has an associated reference coordinate frame \( (O_iX_iY_iZ_i) \). Each joint \( i \) between link \( i-1 \) and link \( i \) is described mathematically by four parameters: \( \alpha_i \), which is the angle between \( Z_{i-1} \) and \( Z_i \) measured
about $X_{i-1}$; $a_i$, which is the distance between $Z_{i-1}$ and $Z_i$ measured along $X_{i-1}$; $\theta_i$, which is the angle between $X_{i-1}$ and $X_i$ measured about $Z_i$; and $d_i$ which is the distance between $X_{i-1}$ and $X_i$ measured along $Z_i$ [4]. Link zero ($i = 0$) is the fixed link that has the base frame attached to it, and the frame index $i$ increments with each joint when approaching the instrument tip from the base frame. The frame assignment has to follow two rules: the $Z_i$ axis must be aligned with the axis of joint $i$, and the $X_i$ axis must be perpendicular to and intersect the $Z_i+1$ axis. The $Y$ axes are chosen to make the coordinate frames right-handed.

Figure 5.15: The robotic instrument with seven modified DH convention coordinate frames attached.

Fig. 5.15 shows the entire instrument with seven coordinate frames that follow the DH convention. The convention allows for flexibility in the placement and orientation of the base frame, the placement of the origin $O_i$ if $Z_i$ and $Z_{i+1}$ are parallel, and the orientation
of the $X$ axis for the final frame. To comply with the dVRK software, the origin $O_0$ must be positioned at the RCM with the $Z$ axis pointing up and the $X$ axis pointing along the outer pitch axis. Also, the origin $O_3$ must be positioned on the plane that is shared by the axes of the two instrument drive pulleys that are closer to the tip in the da Vinci attachment.

Finally, the $X_6$ axis must point along the instrument pitch axis. Following the modified DH convention, the origins $O_1$ and $O_2$ are also positioned at the RCM, $O_4$ and $O_5$ are positioned on the instrument pitch axis, and $O_6$ is positioned on the instrument yaw axis. The DH parameters are presented in Table 5.2. The joint variables are represented by $\theta^*_i$ for revolute joints and $d^*_i$ for prismatic joints. The table includes an offset parameter that indicates the value of the modified DH convention joint variable when the position of the physical joint as measured by the encoders in the robot is zero. The configuration where all the joint positions are zero is shown in Fig. 5.15. The table also includes a mode parameter that is used to set whether a joint is actively controlled by the dVRK software or is simply left to move passively. For this instrument, all joints are required to be active.

Table 5.2: Modified DH parameters used with the dVRK for the robotic instrument.
5.5 Control System

5.5.2 Drive Coupling Parameters

The drive coupling parameters are used to define how the motion of the four instrument drive motors controls the DOFs in the instrument described by the DH parameters. The setup of these parameters allows the motion of multiple motors to actuate a single DOF, but this feature is not used in this instrument since all of the DOFs are decoupled. For this instrument, each row and each column of the coupling matrix has only one non-zero value since each DOF is actuated by a single motor. The values describe the ratio between the angular motion of the interface disks (Fig. 5.16) and the angular motion of the DOF it is controlling. Since the instrument is completely cable driven, this essentially translates into the ratio of the driving pulley pitch diameter to the driven pulley pitch diameter. For the pitch and yaw DOFs, the driving pulleys are 5 mm and the driven pulleys in the wrist are 8.6 mm, making the ratio 0.58, while for the roll DOF, the driving pulley is 16 mm and the driven pulley is 10 mm, making the ratio 1.6. A negative sign on the values indicates that the positive (counter-clockwise) rotation of the drive motor actuates the respective DOF in the negative direction based on the modified DH parameter joint variable. For this instrument, only the roll DOF has a negative coupling value. Disk 2 is unused in this instrument because of the absence of a grasping DOF. The entire coupling matrix is presented in Table 5.3.

![Diagram](image)

Figure 5.16: Numbering convention of the drive interface disks in the dVRK.
Table 5.3: Drive coupling parameters used with the dVRK for the robotic instrument.

<table>
<thead>
<tr>
<th>dVRK Drive Coupling Parameters</th>
<th>Disk 1</th>
<th>Disk 2 (Unused)</th>
<th>Disk 3</th>
<th>Disk 4</th>
</tr>
</thead>
<tbody>
<tr>
<td>Instrument Roll</td>
<td>−1.60</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>Instrument Pitch</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0.58</td>
</tr>
<tr>
<td>Instrument Yaw</td>
<td>0</td>
<td>0</td>
<td>0.58</td>
<td>0</td>
</tr>
<tr>
<td>Instrument Grasping (Unused)</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
</tbody>
</table>

5.5.3 PID Control Loop Parameters

The PID control loop parameters are the gains and the output range of the PID control loops used for position control of the drive motors. These parameters were determined by employing a trial-and-error approach until the response and the stiffness of the instrument resulted in the best palpation performance on silicone phantoms with a maximum palpation force of up to about 14 N. The final PID gains are listed in Table 5.4. The integral limits, the dead band and the position limits are also specified along with the PID gains. The integral limit specifies the maximum value accumulated by the integrator component of the control loop to reduce oscillations and settling time by limiting integral wind-up. The dead band specifies the accepted range of error within which the control loop does not try to correct the error anymore. The dead band value was set to zero for all DOFs to ensure most accurate control. A non-zero dead band is usually specified to prevent oscillations, but it was not required in this case. The position limits specify the extremes of the range of motion of the DOFs about the offset specified in the DH parameters.
Table 5.4: PID control loop parameters used with the dVRK for the robotic instrument.

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Outer Yaw</th>
<th>Outer Pitch</th>
<th>Insertion</th>
<th>Instrument Roll</th>
<th>Instrument Pitch</th>
<th>Instrument Yaw</th>
</tr>
</thead>
<tbody>
<tr>
<td>P Gain</td>
<td>50</td>
<td>50</td>
<td>4000</td>
<td>1.0</td>
<td>2.0</td>
<td>2.0</td>
</tr>
<tr>
<td>D Gain</td>
<td>5.0</td>
<td>5.0</td>
<td>100</td>
<td>0.1</td>
<td>0.2</td>
<td>0.2</td>
</tr>
<tr>
<td>I Gain</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0.002</td>
<td>0.005</td>
<td>0.005</td>
</tr>
<tr>
<td>I Limit</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>±10</td>
<td>±5</td>
<td>±5</td>
</tr>
<tr>
<td>Dead Band</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>Position Limits</td>
<td>±70°</td>
<td>±50°</td>
<td>0 mm – 150 mm</td>
<td>±270°</td>
<td>±90°</td>
<td>±90°</td>
</tr>
</tbody>
</table>

5.6 Visualization Software

Currently, only visual feedback of the tactile and ultrasound information is provided to the surgeon. The sensor feedback is provided to the surgeon as a visual overlay through the da Vinci stereo viewer in order to make it convenient for the surgeon to view the feedback. Fig. 5.17 shows a schematic diagram of the visualization setup. The left channel output of the stereo endoscope system is rerouted to a computer through an Epiphan VGA/DVI-2-USB frame grabber. This computer has access to both the ultrasound and the tactile information. The ultrasound transducer used in the instrument is designed to interface with an Ultrasonix RP ultrasound console. The ultrasound information is retrieved from the console using another Epiphan frame grabber. The tactile information is easily accessible via USB by using the USB interface circuit discussed in Chapter 3.
5.6 Visualization Software

The computer runs a simple application written in C++ that overlays the feedback from the currently active sensor on the endoscope video. This application combines the visualization software developed for the tactile sensor with the libraries provided by Epiphan to interact with the frame grabbers. In this application, the controls that were originally buttons in the tactile sensor visualization software are accessed using keys on the keyboard so that the on-screen display is not cluttered. Pressing ‘P’ or ‘U’ on the keyboard activates the tactile mode or the ultrasound mode respectively. This application runs in full-screen mode on the computer, and the computer’s VGA output is used to transfer the left channel video with the overlaid sensor feedback to the da Vinci stereo viewer. Overlaying the sensor feedback on only one of the stereo endoscope channels creates a pseudo-transparency effect due to the way the human brain combines the left and right channels, allowing the surgeon to see the area behind the overlay. The visualization software currently runs independently of the dVRK software. Once the force sensor is implemented, physical force reflection can be provided to the surgeon using the master controllers in the da Vinci mas-
ter console. This will require combining the visualization software and the dVRK software into one application.

5.7 Evaluation

5.7.1 Prototype

A prototype, that very closely represents the final design of the instrument, was constructed to evaluate the performance of the instrument in localizing tumours. Fig. 5.18 shows the prototype mounted on a da Vinci Classic system with the custom cannula. Fig. 5.19 shows close-up views of the end effector and wrist of the prototype. A major difference between the prototype and the final design is that the springs for keeping the sensor wiring under tension have not been implemented on the prototype at this stage. This is because the wiring of the existing ultrasound module has a layer of heat shrink tubing for protection that makes the wiring too thick to easily slide inside the instrument shaft. The absence of the springs does not affect the functionality of the instrument, it just results in some slack in the wiring (Fig. 5.19) that makes it slightly difficult to pass the instrument through a trocar. Fig. 5.20 shows the simple process of installing a tactile sensor on the instrument. Tests with the prototype confirm that the design allows the tactile sensor to be installed or removed in under five seconds. Fig. 5.21 shows the prototype at the extremes of its range of motion, proving that the wire management technique does allow the instrument to move through the entire range of motion that it was designed for.

A simple experiment (Fig. 5.22) was conducted to assess the amount of cable stretch under the maximum palpation force of ±14 N at the centroid of the palpation surface (simulated by ±10 N at the tip). The test showed that the maximum orientation error is within ±9.5° for both pitch and yaw DOFs. This is considered acceptable because the force/torque
5.7 Evaluation

Figure 5.18: The prototype instrument mounted on the da Vinci Classic system: (a) the complete instrument, (b) close-up of the da Vinci attachment.

Figure 5.19: Close-up views of the end effector and wrist of the prototype instrument: (a) the ultrasound transducer, (b) the tactile sensor.

Figure 5.20: The steps for installing a tactile sensor on the instrument: (a) the instrument without the sensor, showing the spring contacts, (b) inserting the sensor, (c) pushing the securing clip into place to hold the sensor.
5.7 Evaluation

(a) (b) (c) (d) (e) (f)

Figure 5.21: Range of motion of the robotic palpation instrument prototype: (a) roll -270°, (b) roll +270°, (c) yaw -90°, (d) yaw +90°, (e) pitch -90°, (f) pitch +90°.

sensor data can be used to compensate for this when semi-automated palpation is employed in the future. Currently, the instrument is used only in teleoperation mode, and the palpation experiments show that the user can easily compensate for this orientation error because it is only along the direction of palpation, which is intuitive for a human to understand and adjust for.

5.7.2 Experimental Setup

Fig. 5.23 shows the experimental setup used to test the instrument on a da Vinci Classic system retrofitted with the da Vinci Research Kit electronics and software. The user controls the instrument and receives the sensor feedback via the master console (left side of the
figure). The instrument is mounted on the right arm of the surgical robot (right side of the figure). The phantom or ex-vivo tissue sample to be palpated with the instrument is placed on a table under the surgical robot. The endoscope is pointed straight down looking at the sample being palpated. A commercial ATI Gamma force sensor is placed underneath the sample to measure the palpation force since the force sensor on the instrument has not been implemented yet. Currently, the force data is not provided to the user, it is only collected for evaluation purposes.

![Figure 5.23: The setup used for testing the instrument with the da Vinci Research Kit.](image)

5.7.3 Results

Only basic tests to verify the functionality of the instrument have been conducted to date. Fig. 5.24 presents some sample screen-captures of the endoscope view when the instrument is on top of a tumour. First, the instrument was tested on a phantom made of Shore 00-10 clear silicone rubber simulating healthy tissue. The phantom allows the location of the embedded tumours, simulated by Shore 25A silicone rubber balls, to be seen. The simulated tumours are 6 mm in diameter and 4 mm deep. In Fig. 5.24(a), the
5.7 Evaluation

ultrasound image clearly shows the embedded tumour as a speckled white ball. The tests verified that the active sensor can be easily switched by rolling the end effector on the spot. In Fig. 5.24(b), the tumour shows up as a green-red spot on the tactile pressure map.

Next, the instrument was tested on a porcine liver sample with an embedded 10 mm simulated agar tumour. A liver sample was used instead of a lung sample because obtaining a properly deflated lung requires conducting an animal lab and sacrificing an animal, which is very expensive and was considered unnecessary at this stage of evaluation. The liver sample could be easily procured from a supermarket and was considered sufficient to demonstrate the functionality of this instrument on real tissue. In Fig. 5.24(c), the tumour shows up as a dark spot on the ultrasound image that has been highlighted with a red ellipse. Here, it can be seen that it is not very easy to identify a tumour in real tissue using ultrasound alone. This is where the benefit of including a tactile sensor can be seen clearly. In Fig. 5.24(d), the tactile map clearly shows that the suspected area has a stiff lump, confirming the presence of a tumour.

![Figure 5.24: The results of the palpation experiment as seen though the da Vinci stereo viewer: (a) and (b) using a clear silicone phantom with 6 mm simulated silicone tumours at 4 mm depth, (c) and (d) using a porcine liver sample with a 10 mm simulated agar tumour; (a) and (c) ultrasound mode, (b) and (d) tactile mode.](image)

During the palpation tests, the palpation force was recorded using an external force sensor. Fig. 5.25 shows a sample force profile collected when the instrument was tested on a porcine liver sample. The first set of five peaks were obtained when using the ultrasound
mode, and the rest of the peaks were obtained when using the tactile mode. The tumour was located successfully in both modes during this test. This sample force profile shows that the instrument can apply at least 11 N of palpation force. At the conclusion of the test, there was no visible damage to the liver tissue.

![Sample palpation force profile](image)

**Figure 5.25**: Sample palpation force profile of a palpation experiment on a porcine liver sample with the robotic palpation instrument.

### 5.8 Conclusions

This chapter has presented the design, analysis and preliminary evaluation of a novel da Vinci robot compatible palpation instrument that combines tactile, ultrasound and force sensing. The design specifications of the instrument and the design choices made during its development have been discussed in detail. Finite element analysis and manual calculations
were used to ensure that the mechanical design met the strength requirements. A prototype that very closely resembles the final design was constructed successfully, verifying the manufacturability of the design. The control system specifications developed to interface the instrument with the dVRK system were verified by observing the proper functioning of the prototype. The prototype was used to demonstrate that the design does indeed allow for the tactile sensor to be replaced easily, and for movement through the entire desired range of motion. A basic visualization setup was also developed for the instrument to provide the tactile and ultrasound feedback visually, and it proved to work well in the tests conducted so far. Preliminary testing with silicone phantoms and porcine liver samples show promising results. This instrument is the first of its kind, and the positive results obtained so far, encourage pursuing this project further.

References


Chapter 6

Wireless Hand-Held Palpation Instrument

6.1 Introduction

This chapter presents the design, analysis and evaluation of a novel wireless hand-held palpation instrument that is designed to work with the tactile sensor presented in Chapter 3. The motivation behind the development of this instrument is to allow the use of tactile sensing in cases where robot assisted surgery is not used. This instrument uses a single passive degree of freedom to provide the necessary flexibility to palpate tissue in a wide range of orientations. The instrument features an extremely simple user interface that makes it easy to use. The design of the instrument was specifically developed to ensure that it can be easily cleaned and sterilized using an autoclave. The instrument provides both tactile and kinaesthetic feedback to the surgeon in visual form using only the tactile sensor. This chapter reports the major design specifications of the instrument, describes its mechanical and electrical design, validates the design with appropriate analysis, and discusses the evaluation of the prototype.
6.2 Design Specifications

The basic design requirements that were developed to guide the design of the wireless hand-held palpation instrument are listed below:

1. Must be easy and intuitive to manipulate by hand
2. Must fit through a 12 mm diameter or smaller trocar
3. Must be able to provide tactile and kinaesthetic feedback
4. Must be able to withstand the maximum forces that can be applied on it during use
5. Must have sufficient degrees of freedom and range of motion to comfortably palpate tissue in any orientation
6. Must allow for the disposable tactile sensor to be easily replaced
7. Must be biocompatible, cleanable and sterilizable

It was decided that this hand-held instrument must be designed to be wireless in order to make it safer and more convenient for the surgeon to use. Avoiding wires increases the design challenges since it requires the incorporation of a battery and substantially more electronics in the instrument, but it is a challenge worth pursuing due to the advantages of a wireless design. A wireless instrument is more ergonomic for the surgeon to use and avoids issues with cable entanglement, restricted motion, distance to the visualization console, etc. The additional dexterity offered by a wireless instrument is expected to decrease the time taken for palpation and provide better results overall.

Ideally, the instrument should have six DOFs to allow the tactile sensor on the end effector to palpate lung tissue in any position and orientation within the chest cavity. For a hand-held instrument, four of these are in direct control of the surgeon’s hand: two for pivoting at the trocar, one for linear motion of the instrument in and out of the body, and one for the axial rotation (roll) of the instrument. This leaves the pitch and yaw motion that should be implemented close to the end effector. The pitch DOF is essential since it
allows the end effector to orient itself to be parallel to the tissue surface — a requirement for the use of a tactile sensor. The yaw DOF, however, was deemed unnecessary in favour of simplicity since the orientation of the tactile sensor in the plane parallel to the tissue surface does not have a significant impact on the efficacy of the palpation procedure. Based on this justification, the instrument was designed to have only a single internal DOF.

Over a few design iterations, it was discovered that the best approach is to make the pitch DOF passive since it provides three major benefits. Firstly, it makes the use of the instrument much simpler since the surgeon does not have to think about actively controlling another DOF. Secondly, it allows the palpation surface to self-align with the tissue surface, which results in much better consistency than controlling the DOF by hand. Thirdly, it makes the design of the instrument much simpler and much more robust. The pitch DOF was made passive by positioning the joint right above the centroid of the palpation surface, thereby allowing the surface to self-align when pressed against the tissue. To prevent the passive DOF from causing problems when inserting and removing the instrument, and to offer the surgeon with some degree of control over this DOF, a single pull cable is included. Pulling on this cable aligns the end effector with the instrument and holds it in place, allowing easy insertion and removal of the instrument through the trocar. The small amount of resistance created by this cable also helps by preventing the passive DOF from randomly moving around when the instrument is not in contact with tissue.

The instrument uses only a single disposable tactile sensor to provide both tactile and kinaesthetic feedback. The incorporation of an ultrasound transducer as was done for the robotic instrument described in Chapter 5 was avoided to simplify the design and due to the fact that the instrument can be swapped for a standard hand-held laparoscopic ultrasound probe very quickly if required during manual surgery. The passive DOF, combined with the absence of an ultrasound transducer, allows the instrument to not require a force sensor for palpation. To allow the instrument to be sterilized in an autoclave, the Lithium Polymer
(LiPo) battery and the sensitive electronics must be designed to be easily removed before sterilization, and be re-installed while not compromising the sterility of the rest of the instrument. The novel design does not require the use of cumbersome sterile plastic covers that are also risky due to chances of perforation. The specifications of the final instrument design are listed in Table 6.1.

Table 6.1: Major design specifications of the novel wireless hand-held instrument.

<table>
<thead>
<tr>
<th>Design Specifications</th>
</tr>
</thead>
<tbody>
<tr>
<td>Usage</td>
</tr>
<tr>
<td>Hand-held, single hand</td>
</tr>
<tr>
<td>Ergonomics</td>
</tr>
<tr>
<td>Comfortable to hold (33 mm diameter handle)</td>
</tr>
<tr>
<td>Lightweight (≈600 g)</td>
</tr>
<tr>
<td>User Interface</td>
</tr>
<tr>
<td>One button for turning instrument on/off</td>
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<tr>
<td>One LED for wireless link status</td>
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<tr>
<td>Trocar Size</td>
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<tr>
<td>Standard 12 mm</td>
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<tr>
<td>End Effector Design</td>
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</tr>
<tr>
<td>End Effector DOFs</td>
</tr>
<tr>
<td>Pitch: 0–90° (passive)</td>
</tr>
<tr>
<td>End Effector Manipulation</td>
</tr>
<tr>
<td>Pull cable (Tungsten, low-stretch)</td>
</tr>
<tr>
<td>Maximum Transverse Load</td>
</tr>
<tr>
<td>20 N</td>
</tr>
<tr>
<td>Maximum Cable Tension</td>
</tr>
<tr>
<td>10 N</td>
</tr>
<tr>
<td>Maximum Palpation Force</td>
</tr>
<tr>
<td>20 N</td>
</tr>
<tr>
<td>Sensors</td>
</tr>
<tr>
<td>Tactile only</td>
</tr>
<tr>
<td>Provides both tactile and kinaesthetic feedback</td>
</tr>
<tr>
<td>Materials</td>
</tr>
<tr>
<td>Stainless steel 304 for the instrument body</td>
</tr>
<tr>
<td>Delrin (acetal homopolymer) for the handle</td>
</tr>
<tr>
<td>Sterilizability</td>
</tr>
<tr>
<td>Autoclavable</td>
</tr>
</tbody>
</table>
6.3 Mechanical Design

6.3.1 Overview

Fig. 6.1 shows a CAD model of the entire instrument. To make it easier to understand the design of the instrument, the complete instrument assembly is broken into three subassemblies: the End Effector, the Instrument Shaft and the Handle. Each of these subassemblies is discussed in detail in the next three subsections. The instrument has been designed with simplicity, robustness, easy manufacturing and low-cost in mind. The small components in the end effector have been designed for easy manufacture using conventional milling followed by wire EDM and the drilling of a few holes. The large components in the instrument can be easily manufactured using conventional turning and milling processes. The design of the instrument also allows for easy assembly and disassembly. The design has been specifically developed to ensure sterilizability and easy cleaning with minimal disassembly.

Figure 6.1: CAD model of the entire hand-held palpation instrument.

6.3.2 End Effector

The end effector subassembly (Fig. 6.2) has two major components, the sensor holder bracket and the sensor holder, along with a few other small components that form the quick-release mechanism for the tactile sensor. The sensor holder bracket has an axial hole through which a pull cable and the four wires for the tactile sensor exit. The sensor holder
6.3 Mechanical Design

The bracket also has a cantilever arm that extends out to a position directly above the centroid of the sensor holder, where the passive pitch pin joint is located. The underside of the cantilever arm has a slot to guide the tactile sensor wires. The sensor holder is designed to position the sensing area of the tactile sensor in the centre of the palpation area that contacts tissue to ensure an even load distribution. The sensor holder uses a fixed lip on one end and a spring-loaded sliding clip on the other to hold the tactile sensor in place, while still allowing for easy replacement. The end with the sliding clip has a small PCB with the spring contacts to mate with the tactile sensor. The sensor holder also has channels cut on its top surface to guide the wires for the tactile sensor, and a slot with a pin on its proximal end that forms the attachment point for the pull cable (Fig. 6.3). The sensor holder with the tactile sensor is encased by a custom silicone rubber sleeve that serves three purposes — 1. protecting the tactile sensor electrical contacts from body fluids, 2. providing a soft raised surface under the sensing area to ensure that the sensing area makes proper contact with tissue surface, and 3. preventing the tactile sensor from accidentally coming off the instrument. This raised surface is also slightly curved, making it thicker in the middle, which improves tissue contact and pressure distribution for the tactile sensor. The sleeve material must be very soft, such as Smooth-On Ecoflex Shore 00-10 silicone rubber, to minimize reduction in the sensitivity of the tactile sensor.

6.3.3 Instrument Shaft

The instrument shaft subassembly (Fig. 6.4) is the simplest subassembly of the three. It consists of two concentric tubes, both of which are standard 304 stainless steel seamless tubes. The outer tube has an outer diameter of 11.1 mm and an inner diameter of 8.6 mm, while the inner tube has an outer diameter of 3 mm and an inner diameter of 2.5 mm. The four electrical wires for the sensors pass between these tubes, while the pull cable for
6.3 Mechanical Design

Figure 6.2: Labelled exploded and cross section views of the end effector subassembly: (a) exploded view, (b) cross section view.

Figure 6.3: Labelled detail views of the end effector subassembly: (a) showing the sensor clip mechanism, (b) showing the wiring path.
the end effector passes through the inner tube. The space between the outer and the inner tube is sealed at both ends, completely encasing the wires. The electrical wires have a flexible silicone insulation that can withstand the heat from an autoclave. Loctite M31-CL medical grade epoxy is used to attach the sensor holder bracket to instrument tube, and the instrument tube to the corresponding end-cap. The pull cable is a 0.4 mm diameter $8 \times 19$ format stranded Tungsten alloy cable that has an ultimate tensile strength of 220 N, which is well over-designed for this application. The inner tube allows for sufficient space around the cable for easy cleaning of the area with cleaning agents during reprocessing. The inner tube has a $90^\circ$ bend close to the handle where it protrudes out for 2 mm through a hole in the outer tube. This is where the cable exits and terminates in a loop that the surgeon can use to manipulate the end effector.
6.3.4 Handle

The handle subassembly (Fig. 6.4) consists of three major components — 1. the outer plastic tube that forms the actual handle, 2. the inner plastic tube that encloses the battery and all of the electronics that comprise the electronics module, and 3. the two end-caps that seal the inner tube inside the outer tube. The outer plastic tube is a 130 mm long standard Delrin tube with a 1.25" outer diameter and a 1" inner diameter. The tube is threaded on both ends with a 1" National Pipe Tapered Thread (NPT) that the end-caps screw onto. The inner plastic tube is a 120 mm long standard Delrin tube with a 1" outer diameter and a 0.5" inner diameter. The ends of this tube are covered by two circular PCBs that each have four concentric electrical contact rings. One side of the tube has the connections for the tactile sensor, while the other side has the connections for a button, an LED and battery charging. The tube has a lip and a slot on one end that ensures the correct polarity when it is inserted in the instrument or the charging station. One of the end-caps has a stainless steel button with an integrated LED that forms the user interface, while the other is attached to the outer tube of the instrument shaft. The inside of the end-caps has PCBs with four concentric sets of spring contacts that mate with the PCBs on the inner plastic tube. These PCBs are kept at the correct location using plastic spacers that also provide a fixed surface to ensure that the outer plastic tube tightens to the exact same depth in the end-caps every time. These PCBs and the rest of the empty space on the inside of the end-caps is encapsulated with MasterBond EP42HT-2Med, a low-viscosity, medical grade, autoclavable epoxy. This ensures that the electrical connections inside of the end-caps are completely encased and water-tight, allowing the end-caps to be autoclaved. This also reinforces the attachment between the instrument shaft and the corresponding end-cap.
6.3 Mechanical Design

6.3.5 Material and Components Selection

The two major materials used in this instrument are 304 stainless steel and Delrin (acetal homopolymer) plastic. 304 stainless steel was selected since it is inexpensive, meets the strength requirements, and has good machinability. It is the material used for standard components such as the end-cap and the instrument shaft tubes. The end-cap is a standard 1" NPT high-pressure stainless steel threaded pipe fitting that is turned down in diameter and length to reduce weight before drilling the axial hole for the instrument shaft or the button. Delrin was chosen for the handle tubes because of a combination of high strength, low weight and sterilizability, besides it being a standard material for manufacturing tubes. The specific button chosen is a Bulgin MPI002/28/RD IP68-rated waterproof button made of stainless steel and autoclavable plastic, with an integrated red LED. Mill-Max 0908-0-15-20-75-14-11-0 spring contacts are used in the end-cap PCBs. These spring contacts are made of copper alloy with gold plating to make them corrosion resistant.

6.3.6 Sterilization

Before sterilizing the instrument, the electronics module is removed by unscrewing the button end-cap. After this, the instrument can be easily cleaned throughout and sterilized by autoclaving. During this time, the electronics module is re-charged followed by high-level disinfection using germicidal wipes. After the rest of the instrument is sterilized, the electronics module is re-inserted with the help of a sterile plastic drape to ensure that the external sterile surface of the instrument does not contact the non-sterile electronics module. Once the module is in place, the button end-cap is tightened thereby sealing off the non-sterile area of the instrument and leaving only the sterile area exposed.
6.4 Analysis

6.4.1 Finite Element Analysis

FEA was performed on the critical components of the instrument using Solidworks in order to ensure sufficient strength. The critical structural components were identified as the sensor holder bracket, the outer tube of the instrument shaft, and the outer tube of the handle. Other components are sufficiently over-designed in comparison to these, and therefore it is expected that one of the aforementioned components will be the first to fail. The analysis of these components was divided into two parts, one consisting of all of the stainless steel components and the other consisting of all of the plastic components. This was done for two reasons — it is difficult to accurately simulate the stresses at the threaded connection between the end-cap and the handle, and it is difficult to interpret the stress results if there are materials with vastly different yield strengths in the same study.

The pin joint in the end effector was simulated in Solidworks by using a pin connection feature. The end effector and the instrument shaft were analyzed in two worst-case loading scenarios — 20 N applied at sideways at the very tip (Fig. 6.5) and 20 N applied at the centroid of the palpation surface, right under the pin joint (Fig. 6.6). In the first case, the maximum stress was 145 MPa, and in the second case it was 98 MPa. The minimum FOS was determined to be 2, which is considered adequate since the loading is about twice what is expected under normal use.

In order to analyze the outer tube of the handle without the instrument shaft, it was extended beyond the threaded region to the same length as the distance to the tip (Fig. 6.7). This allowed the worst-case 20 N lateral load to be applied at the end of the extended tube in order to assess its effect on the threaded area, which is prone to stress concentrations. The results showed that the expected maximum stress is only 1.32 MPa leaving an FOS of 4.7, which indicates that the failure of the handle is not a concern.
6.4 Analysis

Figure 6.5: FEA von Mises stress results for the maximum lateral tip loading on the end effector and the instrument shaft.

Figure 6.6: FEA von Mises stress results for the maximum palpation loading on the end effector and the instrument shaft.
6.4 Analysis

6.4.2 Pin Analysis

There is only one critical pin in this design, the one connecting the sensor holder to the sensor bracket to form the passive pitch motion DOF. This pin is 1.2 mm in diameter and is made of solid 416 stainless steel, with a quoted double-shear yield strength of 830 N. The worst case loading scenario for this pin is 20 N applied laterally at the tip of the sensor holder, 30 mm from the pin joint. This lateral force creates a moment of 600 N·mm at the pin joint. Since the distance between the two shear interfaces is 4.5 mm, the interfaces experience a shear force of 133 N in opposite directions. This translates to a double-shear load of 266 N in the worst case scenario, which results in a factor of safety of over 3. The failure of this pin is therefore not a concern in this design.

Figure 6.7: FEA von Mises stress results for the worst case loading on the outer plastic tube of the handle.
6.5 Electrical Design

6.5.1 Embedded Electronics

The electronics embedded within the inner tube of the handle consists of a 12 × 70 mm cylindrical 1000 mAh single-cell Lithium Polymer (LiPo) battery and a 12 × 46 mm printed circuit board. The LiPo battery was selected due to its convenient form factor, low weight, low cost and high energy density. It can power the embedded electronics (100 mA) for 8 hours on a single charge, and takes only 2 hours to recharge. The LiPo battery can withstand 300 charge cycles before requiring replacement, which results in a battery life of over two years, which is reasonable for a medical device [1]. A schematic of this circuit is presented in Fig. 6.8(a). The circuit consists of a Parallax Propeller V1 microcontroller, a Panasonic PAN1322 Bluetooth module, a 3.3 V Low-Dropout Linear Voltage Regulator and a battery monitor. Bluetooth was chosen as the wireless protocol due to its low power consumption, sufficient range and noise immunity. The Bluetooth protocol allows the embedded and the external module to be programmed to automatically establish secure communications with only each other, and ignore other Bluetooth devices. Four out of the eight cores in the microcontroller are used to perform the following tasks simultaneously: 1. communicating with the tactile sensor, 2. repackaging the data packets, 3. communicating with the Bluetooth module, and 4. monitoring the battery status and the user interface.

The user interface to this circuit is a push button and an LED which is incorporated in the push button itself. When the instrument is off, pressing the push button for one second turns it on. Once on, the microcontroller tries to establish communications with the Bluetooth module via the Universal Asynchronous Receive and Transmit (UART) protocol at 115.2 kHz and with the tactile sensor via the Inter-Integrated Circuit (I²C) protocol at 100 kHz. If the embedded Bluetooth module does not link with the external Bluetooth module within 10 seconds, indicating that the external module is offline, or if a tactile
sensor is not detected on the instrument, the microcontroller turns the instrument off to save battery life. While communications is being established, the LED blinks at 2 Hz, and it turns solid when the instrument is ready for use. To turn the instrument off, the push button must be held down for two seconds. The instrument also shuts down on its own if the tactile sensor is removed or if the external Bluetooth module is turned off. A comparator is used to detect when the battery voltage drops below 3.5 V, indicating that the battery is drained to about 20% of its capacity. At this point, the microcontroller shuts off the instrument and prevents it from being turned back on to prevent the battery from becoming damaged by over-discharging.

![Schematic Diagrams](image)

Figure 6.8: Schematic diagrams of the Bluetooth interface circuits with current consumptions: (a) embedded electronics, (b) external USB-to-Bluetooth electronics.

### 6.5.2 External Electronics

There are two pieces of external electronics associated with the instrument: the USB-to-Bluetooth dongle that allows the instrument to connect to a computer, and the charging
6.6 Visualization Software

The visualization software used with this instrument is a modified version of the software used with the robotic palpation instrument discussed in Chapter 5, with essentially the ultrasound feedback removed and kinaesthetic feedback added. Both tactile and kinaesthetic feedback are provided to the surgeon simultaneously as visual overlays on the endoscope video that is typically displayed on several monitors in the operating room. This setup makes it convenient for the surgeon to view the feedback. Fig. 6.9 shows a schematic diagram of the visualization setup. The output of the endoscope video processor is rerouted to a computer through an Epiphan VGA/DVI-2-USB frame grabber. This computer has access to the tactile information via USB through the USB-to-Bluetooth dongle discussed in the last section. The computer runs a simple application written in C++ that
overlays the tactile feedback in the form of a tactile pressure map, and the kinaesthetic feedback in the form a force level bar on the endoscope video. The force level bar is set up in a similar manner as it was in the original tactile sensor visualization software discussed in Chapter 3. This application combines the visualization software developed for the tactile sensor with the libraries provided by Epiphan to interact with the frame grabber. In this application, the controls that were originally buttons in the tactile sensor visualization software are accessed using keys on the keyboard so that the on-screen display is not cluttered. This application runs in full-screen mode on the computer, and the computer’s VGA output is used to transfer the endoscope video with the overlaid sensor feedback to the monitors.

Figure 6.9: Schematic of the visualization setup for overlaying the sensor feedback on the endoscope view.

6.7 Evaluation

6.7.1 Prototype

A prototype of this instrument that very closely resembles the final design has been constructed to validate the functionality of the instrument (Figs. 6.10 and 6.11). The prototype verifies that the electronics meet the design specifications in terms of battery life and wireless range, and that the mechanical design allows for the desired range of motion. The prototype also demonstrates that the encapsulation of the end-caps proposed in the design
is a feasible approach to completely seal the cavities in the instrument while still ensuring proper functionality. Fig. 6.12 shows the simple process of installing a tactile sensor on the instrument. The mechanical structure was tested with 10 N applied at the tip in various directions to ensure that the design met the strength requirements, as shown in Fig. 6.13. No permanent deformation of the instrument was observed with a 10 N load.

Figure 6.10: The prototype wireless hand-held palpation instrument: (a) the complete instrument, (b) and (c) extremes of the range of motion of the end effector, (d) the electronics module removed from the handle.

Figure 6.11: The external electronics associated with the instrument: (a) the USB-to-Bluetooth dongle, (b) the charging station, (c) the charging station with the electronics module in place.
6.7 Evaluation

(a) (b) (c)

Figure 6.12: The steps for installing a tactile sensor on the instrument: (a) the instrument without the sensor, showing the spring contacts, (b) inserting the sensor, (c) pushing the securing clip into place to hold the sensor.

Figure 6.13: Experimental setup used to verify that the instrument can withstand 10 N applied at the tip.

6.7.2 Experimental Setup

The setup used to test the functionality of the instrument is shown in Fig. 6.14. The setup uses a minimally invasive surgery training box with a silicone phantom inside it that can be placed at various locations and orientations. The endoscopic view, displayed on a laptop placed on top of the box, is used to manipulate the instrument inside the box.

6.7.3 Results

Only basic tests to verify the functionality of the instrument have been conducted to date. The instrument was tested on a phantom made of Shore 00-10 clear silicone rubber
simulating healthy tissue. The phantom allows the location of the embedded tumours, simulated by Shore 25A silicone rubber balls, to be seen. The simulated tumours are 6 mm in diameter and 4 mm deep. The results obtained are presented in Fig. 6.15. The results show that the instrument allows the palpation of tissue with successful tumour localization in a wide range of orientations.

Figure 6.14: The setup used for evaluating the performance of the instrument: (a) front view, (b) rear view with the laptop removed.

**6.8 Conclusions**

This chapter has presented the design, analysis and preliminary evaluation of a novel wireless hand-held palpation instrument that uses a tactile sensor to provide tactile and kinesthetic feedback to the surgeon. The design specifications of the instrument and the design choices made during its development have been discussed in detail. Finite element analysis and manual calculations were used to ensure that the mechanical design met the strength requirements. A prototype that very closely resembles the final design was con-
structured successfully, verifying the manufacturability of the design. The prototype was used to demonstrate that the design does indeed allow for the tactile sensor to be replaced easily, and that the instrument can palpate tissue in various orientations from the same entry port. The effects of autoclaving on the instrument have not yet been tested. However, the simplicity of the design and the choice of materials provide a high likelihood that the instrument will be able to withstand several autoclave cycles. A basic visualization setup was also developed for the instrument to provide the tactile and kinaesthetic feedback, and it has worked well in the tests conducted so far. Preliminary testing with silicone phantoms show promising results. This instrument is the first of its kind in terms of sterilizability and ease of use, and the positive results obtained so far encourage pursuing this project further.

References

Chapter 7

Conclusions

7.1 Summary

The thesis began by presenting some background information to put the research presented herein into perspective. Chapter 1 presented the relevant clinical details related to intraoperative lung tumour localization, and Chapter 2 presented a literature review of the state-of-the-art in tactile sensing, force sensing, and tumour localization instruments. The literature review showed that there are no existing tactile sensors that have been designed to be either disposable or sterilizable, which is a major requirement to enable clinical use of tactile sensors. Some low-cost designs have been proposed, but they do not allow for easy replacement when incorporated in an instrument, and hence do not qualify as being disposable. The review also highlighted that there are no existing 6-DOF force/torque sensors that meet the high loading requirement of a palpation instrument while still being compact enough. Furthermore, no existing force sensor design supports multi-axial force measurements with temperature compensation using a single optical fibre. An optical force sensor has advantages over other designs because it can be made more robust and sterilizable. Also, to the best of the author’s knowledge, there is currently no da Vinci robot compat-
ible multi-modal palpation instrument. The existing palpation instruments are designed to be either hand-held or to be manipulated by custom surgical robots. A da Vinci robot compatible instrument is desirable since the da Vinci robot is the only robot that currently has clearance to operate on humans in North America. The review also revealed that there are no existing wireless hand-held palpation instruments that are designed to be sterilizable, exposing yet another area where a contribution can be made. The problems and gaps identified in the existing technology motivated the development of the novel sensors and instruments presented in Chapters 3 to 6 of this thesis. A summary of Chapters 3 to 6, along with the major conclusions that were drawn from the work, are presented below.

7.1.1 Chapter 3 — Tactile Sensors

This chapter presented the design, analysis and evaluation of novel low-cost tactile sensors that are designed to be easily replaceable, thereby making them disposable. The mechanical and electrical design of the sensors were discussed in detail, while justifying every design decision. Four different designs are proposed, two different sizes with a piezoresistive sensor and a capacitive sensor in each size. The sensors feature on-board analog single processing and use only four digital interface lines. The experiments showed that the individual sensing elements of the sensors can be calibrated to measure forces in the range of 0.1 N to 1 N during 0.2 Hz to 0.5 Hz palpation with an accuracy and repeatability (including hysteresis) of better than 84% and 86% respectively. The palpation tests with a silicone phantom demonstrated that the sensors can detect 5 mm tumours at a depth of up to 10 mm without calibrating individual elements. A bulk calibration technique to improve the quality of tactile feedback was also described and proven experimentally. The results of the initial tests with these tactile sensors are encouraging and suggest that they can be very useful in improving the localization of tumours in the lung and other organs.
7.1 Summary

7.1.2 Chapter 4 — Miniature 6-DOF Force/Torque Sensor

This chapter presented the design and analysis of a novel Fibre Bragg Grating based miniature 6-DOF force/torque sensor. The sensor uses 12 Fibre Brag Gratings on a single optical fibre to provide 6-DOF force/torque measurements with temperature compensation. A mathematical model of the sensor predicts that the sensor will be capable of a large 3D force and 3D torque measurement range of $\pm 230$ N and $\pm 550$ N-mm respectively, while providing a theoretical accuracy of 0.5 N and 0.7 N-mm. Finite Element Analysis was performed to ensure that the sensor does not fail within the quoted measurement range. The simple design makes the sensor easy to manufacture and integrate into other structures. The construction of the prototype was faced with several challenges that prevented its timely completion. Therefore, no experimental validation is available for the quoted specifications of the sensor. However, the advantages of the proposed design motivate further work on this sensor.

7.1.3 Chapter 5 — Robotic Palpation Instrument

This chapter presented the design, analysis and evaluation of a novel da Vinci robot compatible palpation instrument. This instrument uses the tactile sensor and the force sensor presented in this thesis, along with a custom ultrasound transducer to provide tactile, kinaesthetic and ultrasound feedback to the surgeon. It is hypothesized that fusing data from these three sensing modalities can significantly improve tumour localization accuracy. The instrument has an articulated wrist with three internal DOFs to allow the end effector to follow the natural motion of the surgeon’s hand. The instrument is designed to withstand forces of up to 20 N applied at the tip, and apply a palpation force of up to 14 N on the tissue surface. The cable-actuated joints have been designed to have decoupled motion to minimize the problems caused by cable stretch. A visualization system was also
developed to overlay the tactile and the ultrasound information on the endoscope view seen by the surgeon in the da Vinci master console. A prototype of the instrument was constructed to verify its functionality. Experiments on silicone phantoms and ex-vivo tissue samples showed that the instrument can be manipulated in a very controlled manner using the da Vinci master console, and that it makes the palpation process feel very natural and convenient. An experiment with a 10 mm agar tumour in a porcine liver sample demonstrated that a palpation force of 11 N is sufficient to locate tumours using both ultrasound and the tactile sensor. Further studies are required to establish a quantitative measure of the improvement in tumour localization provided by this instrument.

7.1.4 Chapter 6 — Wireless Hand-Held Palpation Instrument

This chapter presented the design, analysis and evaluation of a novel wireless hand-held palpation instrument that uses the tactile sensor presented in this thesis to provide both tactile and kinaesthetic feedback. There are two major innovations in the design of this instrument. The first one is the single passive DOF in the end effector that enables the tactile sensor to self-align with the tissue surface making the instrument easier to use for the surgeons. The second one is the manner in which the electronics and the battery are packaged in a single removable module enabling the rest of the instrument to be autoclaved. The primary motivation behind the development of a hand-held instrument was to enable the use of the tactile sensors in procedures where the da Vinci surgical system is not used. A prototype of the instrument was constructed to verify its functionality. Some simple experiments were conducted with a minimally invasive surgery simulator and a silicone phantom to show that the instrument works as intended. The qualitative experiments showed that the passive end effector enables the instrument to be used from different approach angles without affecting its performance or limiting its dexterity.
7.1 Summary

7.1.5 Challenges

The development of each of the four components of this project was faced with its own set of challenges. Most of the difficulties arose due to following three requirements on the instruments and sensors – they have to fit through a standard 8 mm, 10 mm or 12 mm trocar used in MITS, they must be sterilizable to Health Canada and FDA standards, and they must have a sufficient factor of safety to be considered safe for clinical use.

The tactile sensor had to be designed to minimize its cost so that the sensor could be made disposable, which was a big challenge. Disposability was important because it was discovered that it is extremely difficult to make a tactile sensor that can withstand the rigorous reprocessing of reusable medical devices. Another challenge that accompanied disposability was the requirement to make the sensor easily replaceable, restricting the number of interface lines that it can use. Optimizing the selection and placement of the on-board electronics to maximize the spatial resolution of the sensor was a very time consuming process. Perfecting the novel low-cost technique used to make pseudo-three-layer printed circuit boards by fusing two two-layer printed circuit boards required several iterations.

The 6-DOF force/torque sensor was the most challenging of all of the components of this project. The size constraint, along with the restricted space for cabling, imposed a big constraint on the sensing technologies that could be used for the sensor. It was also a requirement to make the sensor easy to construct so that a prototype could be constructed with the available manufacturing resources. The sensor required several iterations to optimize the design for the best possible sensing accuracy and ease-of-manufacture. The fabrication of the prototype presented several challenges in fine-tuning the manufacturing process to obtain the narrow grooves in the sensor for the optical fibre. Additional challenges were encountered in the accurate positioning of the optical fibre in the grooves, which prevented the timely completion of the prototype.
7.2 Concluding Remarks

The design of the robotic instrument was also challenging due to the space constraints and the requirement of having three sensing modalities in the end effector. The biggest difficulty was to make space for the sensor cabling around the actuated wrist and down the instrument shaft. Making sure that the small components forming the wrist mechanism can provide the full range of motion, while still keeping the mechanism simple, robust and easy to manufacture was also a big challenge. Ensuring that the two degrees of freedom at the wrist are decoupled to minimize problems in control due to cable stretch was also difficult task. In the manufacturing of the prototype, the most challenging aspects were making the small pulleys using spindle wire EDM and drilling the small holes in the wrist components with good accuracy.

The design of the wireless hand-held instrument was the least challenging of all of the components of this project. The biggest difficulty in the design of this instrument was to develop a means to easily remove and reinstall the electronics and the battery so that the rest of the instrument can be autoclaved. The design of the mechanism for the passive degree of freedom at the end effector also posed a challenge and required a few iterations. No significant issues were encountered in the manufacture of this instrument due to the simplicity of the design.

7.2 Concluding Remarks

The work presented in this thesis shows great potential in improving lung tumour localization in various ways such as:

1. Better localization accuracy due to the use of multiple modalities,
2. Lower risk of tissue damage due to the incorporation of kinaesthetic feedback,
3. More convenient and controlled palpation when using the robotic palpation instrument due to the inclusion of robotic assistance, and
4. Better ergonomics and ease-of-use for the surgeon when using the hand-held palpation instrument due to the passive DOF and the wireless design.

Even though the motivation for the work was lung cancer treatment, these devices can also be used on other organs, such as the liver. During the design of each of the four components, practical considerations such as ease-of-manufacture, cost, sterilizability, compactness, ease-of-use, etc. were always taken into account for each design decision. This approach ensured that the presented designs were developed to be as practical as possible. The four components have been specifically designed to work synergistically with one another and improve tumour localization as a whole.

Due to the focus on the design, analysis and prototyping of the novel components developed for this thesis, which took a considerable amount of time and work, limited experimental validation has taken place. Therefore, extensive testing, iterations and improvements are still required to allow the novel devices presented here to reach their full potential. The work presented herein also has the potential to initiate the development of other new MIS instruments. For example, the 6-DOF miniature force/torque sensor can be incorporated in a da Vinci grasper tool to provide haptic feedback to a surgeon via the master console, or the design of the robotic instrument can be adapted to make other custom da Vinci instruments for research purposes.

7.3 Recommendations and Future Work

As mentioned earlier, there is still a significant amount of work to be done to take the presented sensors and instruments to a point where they are fully operational and can undergo extensive clinical evaluation on animals and eventually humans. Some recommendations to improve the devices and suggestions for the future work to be performed are presented in this section.
7.3 Recommendations and Future Work

7.3.1 Tactile Sensors

The final tactile sensor designs have to be evaluated in various ex-vivo and in-situ tests to determine the size, depth and stiffness of tumours that can be detected reliably by the four different versions of the sensors. To further optimize the response, algorithms have to be developed to compensate for the stress relaxation behaviour seen in the sensors. The proposed technique for the bulk calibration of the sensor has to be refined to improve its performance in compensating for the differences in the sensitivity of the individual elements due to the surface irregularities on the PCB and the sandwiched material.

Better techniques to reliably seal the sensing side of the sensors from body fluids have to be developed and tested. Probable options are using epoxy to seal the edges of the metal foil or wrapping the entire sensor in water-resistant silicone or polyurethane tape. It must be ensured that the sealants and adhesives used are biocompatible and not rigid so that their effect on the sensitivity of the sensors is minimized. The effects of these sealing techniques on the sensor performance have to be investigated as well. Furthermore, for the eventual medical use of these sensors, the effectiveness of one-time sterilization during manufacturing using techniques such as gamma irradiation and ozone have to assessed, along with ensuring that they do not damage the sensor.

The visualization software should be improved based on user feedback to ensure that it communicates the ideal amount of information to the surgeon. The colour scale in the visualization software currently has an approximately linear relationship with the measured force, but a different nonlinear relationship may provide better visualization of tumours. It may also be worth investigating if there are better piezoresistive and dielectric materials that can improve the sensor performance, and are either available commercially or can be custom manufactured for a sufficiently low cost. Some modifications to the design, such as making the sensing pads domed rather than flat like it is in the design proposed by Schostek
et al. [1], may be tested to see if they are advantageous.

### 7.3.2 Miniature 6-DOF Force/Torque Sensor

For the force/torque sensor, the next step is to design plastic/rubber tools for manipulating the optical fibre without damaging it, and a jig to guide the placement of the fibre in the sensor and to hold the fibre in place while the adhesive cures. The design and manufacture of these tools and jig is expected to form the majority of the remaining work on this sensor. Once the functionality of the tools and jig is verified on the 17-4PH stainless steel prototype that has already been constructed, a prototype has to be constructed out of Grade 5 Titanium to actually evaluate the sensor performance. A mould for encapsulating the sensor in silicone after fibre placement has to also be designed and manufactured. A system to automate the calibration of this sensor may also be developed to make the process fast and repeatable. This system can also be used to test the response of the sensor after calibration.

The accuracy and repeatability of the prototype sensor must be evaluated under a variety of loading scenarios and temperatures. Another important factor to be investigated is the effect of using different cyanoacrylates and UV-cure glues to mount the fibre, in terms of both measurement accuracy and construction reliability. Next, the variation in the sensor performance as a function of the number of autoclave cycles must be investigated, and any necessary design changes must be made to ensure that the performance of the sensor is stable over a reasonable number of autoclave cycles. Optical fibres and interrogators from other manufacturers may also be tested to determine the combination that yields the best measurement accuracy.
7.3.3 Robotic Palpation Instrument

Before moving forward with further development, tests must be conducted with several subjects to ensure that the instrument design allows it to be manipulated comfortably using the da Vinci Master Console, and that it allows proper tissue palpation at various orientations. The results of these tests may point towards design modifications that can make the instrument easier and more intuitive to use. Currently, the active mode (tactile or ultrasound) is switched by manually rotating the finger grip on the da Vinci Master Tool Manipulator to rotate the end effector and then pressing a button on the PC running the visualization software. This process is cumbersome for the user, and therefore a much better alternative would be to use one of the foot pedals in the Master Console to perform the switch automatically.

Rigorous loading tests have to be conducted on the instrument to determine the loads that result in failure or excessive permanent deformation. This is necessary to establish the true factor of safety of the mechanical structure and identify the weak components. Since these tests will conclude with damaged components, copies of the wrist and the end effector of the instrument have to be manufactured. Once the force/torque sensor has been developed and tested, it has to also be incorporated into the palpation instrument prototype to enable kinaesthetic feedback. During the experiment with the liver sample, it was seen that the ultrasound gel applied on the tissue surface entered the sensor holder and disrupted the electrical connection between the tactile sensor and the four spring contacts. Therefore, some modifications must be made to the sensor holder to seal the tactile sensor contacts against contaminants. This may be as simple as filling the gaps with silicone.

Better visualization software must also be developed to fuse the data from the tactile sensor, the ultrasound transducer and the force/torque sensor, and provide intuitive feedback to the surgeon via the Master Console. The data from the force/torque sensor can
also be used to provide physical haptic feedback to the user via the Master Tool Manipulators. This is worth pursuing because past studies, such as the one conducted by Tavakoli et al. [2], have shown that physical haptic feedback reduces the time needed for palpation in comparison to visual feedback of kinaesthetic data. Some basic image processing techniques can be employed to overlay the tactile and/or ultrasound data exactly where the end effector is present on the endoscope view to make it easier for the surgeon to localize the tumours, similar to what has been proposed by Miller et al. [3].

Once the instrument is fully functional under manual control with physical haptic feedback, algorithms and software to make the palpation procedure semi-automated may be developed. The most basic semi-automation would be the automatic control of the palpation force such that it never exceeds a specified safe limit for the tissue being palpated. More advanced automation, such as automatic alignment of the sensor surface to the tissue surface, and automatic palpation of a region specified by the surgeon with an adjustable step size, can also be implemented. The automatic palpation algorithm can be made smarter by incorporating an adaptive step size feature that increases the number of palpations around an area where a tumour is suspected. The effect of drive cable stretch under both manual and semi-automated palpation must be investigated.

### 7.3.4 Wireless Hand-Held Palpation Instrument

Due to the simple design of this instrument, not a lot of work needs to be done on improving the design, but there is still a significant amount of testing to be performed. Some design modifications may be required for the silicone sleeve to improve both its sealing efficacy and the response of tactile sensor. The mould for the final sleeve design must be machined out of metal rather than rapid-prototyped to achieve better tolerances and a smoother surface finish. Similar to what was suggested for the robotic instrument,
7.3 Recommendations and Future Work

Rigorous loading tests have to be conducted on this instrument as well to determine the loads that result in failure or excessive permanent deformation.

Some of the improvements to the visualization software suggested for the robotic instrument also apply here, specifically, using augmented reality techniques to overlay the tactile data exactly where the end effector is present in the endoscope view to make it easier for the surgeon to localize the tumours. This would require tracking the instrument, either by employing image processing on the endoscope video, or by using a tracking device, preferably wireless. A suggested minor design improvement is adding a mechanism to the handle that locks the pull cable in the retracted position to make it easier and safer to insert and withdraw the instrument through a trocar. A couple of other minor improvements include making the handle more ergonomic by adding contours for the hand, and redesigning the charging station so that the electronics module can be locked into place rather than depending on gravity. The instrument prototype must also be put thorough a series of autoclaving cycles to ensure that the encapsulating epoxy inside the end caps does not fail.

7.3.5 Clinical Testing and Evaluation

After the instruments and sensors have been thoroughly tested using phantom models and bovine/porcine ex-vivo tissue samples in a lab, and all of the necessary modifications have been made, they must be taken to live animal trials in the operating room. This will require further ethics approvals and the establishment of a proper experimental protocol. Randomized control trials must be run with several users of different skill levels to assess whether this system provides statistically significant improvement in terms of the safety, time and accuracy of the palpation procedure in comparison to other approaches.

If the system still exhibits positive results after thorough evaluation on animals, further
improvements must be made such as redesigning for easier disassembly and cleaning, making the devices completely sterilizable, etc. to make it suitable for clinical use. Following this, the devices must be taken to live human trials after obtaining the necessary approvals. Several tweaks may be necessary to make sure that the devices can actually be mass produced. The final designs must obtain Health Canada and FDA approval so that they may actually be sold for use in North America. Even though there is still a really long way to go before the commercial use of these instruments is possible, they offer the potential to overcome a major limitation of minimally invasive surgery.

References


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