Robotic Actuation for Fetoscopic Interventions

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I, George Dwyer, confirm that the work presented in this thesis is my own. Where information has been derived from other sources, I confirm that this has been indicated in the work.
Abstract

Fetoscopic surgery is a form of minimally invasive surgery within the womb during pregnancy. It involves the introduction of small diameter, rigid instrumentation to fix defects in the placenta or developing fetus. Surgery within the uterine environment is technically very challenging due to the environment containing many critical structures that the instrumentation should not come into contact with. Robotics could potentially assist in these procedures through stabilising the instrumentation and introducing additional degrees of freedom at the tip of the instruments. This thesis presents the design, control and application of robotic manipulators positioned both outside and within the body to control imaging sensors. Custom endoscopes have been designed and fabricated with white light imaging sensors and all-optical ultrasound sensors. These endoscopes are held outside the body, proximal to the surgeon, by an articulated robotic manipulator. Additionally, within the body, distal to the surgeon, a continuum manipulator provides articulation at the tip of the instrument. The articulated robotic manipulator is constrained to the surgical incision point using a remote centre of motion which allows the endoscope to only pivot about the point and translate along the instrument axis. The continuum manipulator used is a form of concentric tube mechanism, which relies on the interactions between a curved tube of a super elastic material and a straight rigid tube, this can be controlled to deflect the tip of the endoscope from the main instrument axis. The developed endoscopes are then demonstrated through the execution of generated trajectories, which can be used to derive geometry of the imaging target and enhance the field of view of the imaging modalities.
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Impact Statement

This thesis has explored the integration of robotics and imaging sensors to improve fetoscopic interventions. While the work has been focused on fetal interventions, the instrumentation developed and insights gained can be applied to other endoscope-based procedures and, to some extent, the general control of imaging sensors with robotic manipulators. The instrumentation presented within this thesis has been designed to hold imaging components, specifically small chip-on-tip cameras and fibre optic based sensors; an arrangement which allows the instrumentation to be modified for other imaging modalities (as replacements or in combination). Moreover, this work has been disseminated across the academic community at five conferences: twice at the International Conference on Robotics and Automation (ICRA) (the Institute of Electrical and Electronics Engineers (IEEE) flagship conference for robotics), and three times at the Joint Workshop on New Technologies for Computer/Robot Assisted Surgery (CRAS) — a predominantly European multidisciplinary conference attended by engineers and surgeons.

In the future, this research could be used to enhance intraoperative imaging: an enhancement achieved through the improved stability and increased dexterity the instrumentation provides. This approach allows the small diameter manipulators to be closely coupled with the imaging modalities, thus providing increased diagnostic information for the surgeon. Additionally, the manipulators also have the potential to reduce invasiveness to the patient due to the reduced diameter of the instrumentation and improved control. Overall, the approach could enable the provision of clinically relevant information to the surgeon in a manner that was previously more complex and less readily comprehensible. In terms of future developments, it also
offers significant opportunities to reduce the training, cognitive and physiological burdens placed upon surgeons.

Although the pathway to clinical approval is invariably protracted, over the course of the thesis the work has been presented to the public across 3 public engagement events and 2 televised interviews. (Public engagement is perceived as a critical element in informing the public and gaining acceptance of potentially contentious emerging technologies, such as robotic surgery.) For the public engagement events, the first was at the Royal Academy of Engineering, in a networking event for their fellows in 2015; the second was in New Scientist Live 2017, with the UCL Institute of Healthcare Engineering; the third was at GITEX 2018, a technology expo in Dubai. Lastly, the work was presented as a part of a larger project on fetal surgery (GIFT-Surg) and was shown in 2 televised interviews: on Al Jazeera (2015) and BBC Arabic (2017).

Finally, the mechanism designed and implemented through this thesis is now used as a platform to underpin additional research in continuum robot control. In addition, a number of interfaces developed to control the robot arms, the imaging sensors and the tracking systems are now used for other laboratory-based surgical robotics work and for industrial manipulation and scanning within a number of UCL labs.
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Acronyms

ABS  Acrylonitrile Butadiene Styrene.

API  Application Programming Interface.

CAD  Computer Aided Design.

CMOS  Complementary Metal-Oxide-Semiconductor.

CNC  Computer Numerical Control.

CTM  Concentric Tube Manipulator.

DoF  Degrees of Freedom.

EM  Electromagnetic.

EtO  Ethylene Oxide.

FDM  Filament Deposition Modelling.

FETO  Fetoscopic Endoluminal Tracheal Occlusion.

FLP  Fetoscopic Laser Photocoagulation.

FoV  Field of View.

FPS  Frames Per Second.

FRI  Fast Robot Interface.

HHS  Helical Hollow Strand.
ICP  Iterative Closest Point.

LUTO  Lower Urinary Track Obstructions.

MIS  Minimally Invasive Surgery.

NTP  Network Time Protocol.

OpUS  Optical Ultrasound.

PAM  Pneumatic Artificial Muscles.

PLA  Polylactic Acid.

RCM  Remote Centre of Motion.

ROS  Robotic Operating System.

SMA  Sub Multi Assembly.

TCM  Tendon-driven Continuum Manipulators.

TTL  Transistor - Transistor Logic.

TTTS  Twin-Twin Transfusion Syndrome.

UART  Universal Asynchronous Receiver/Transmitter.

US  Ultrasound.
Chapter 1

Minimally Invasive Surgery and Interventions

Surgical innovation is often focused on reducing the invasiveness of the procedure while increasing or at least maintaining the efficacy of the procedure. This can be seen from the transition of surgical procedures from open to minimally invasive approaches. In open surgery, long incisions are made to provide both direct access for the instrumentation needed and a line of sight for the surgeon to perform the operation. Minimally Invasive Surgery (MIS) is a term used to define surgical procedures that reduce the invasiveness of the approach. MIS includes "keyhole" surgery, catheter-based procedures and natural orifice surgery. Figure 1.1 shows the core differences between MIS and open surgery. Keyhole surgery is one of the more common approaches especially in general surgery. The approach relies on multiple small incisions, rather than a single large one, to gain access to the surgical site. At one incision point, a long rigid endoscope is inserted for visualisation; and in the other incisions, long surgical instruments are inserted to perform the surgery. This has a number of advantages over traditional open approaches, including reduced blood loss which, in turn, reduces the chance of needing a blood transfusion. Smaller incisions also lead to reduced pain and faster recovery, while reducing external exposure to the organs and, therefore, risk of infections [12, 13]. Minimally invasive approaches can also represent lower hospital costs compared to open surgery. For example laparoscopic resection of colon cancer is £2202 cheaper
Figure 1.1: Images displaying the fundamental differences between MIS and open surgery. **Left** - an open surgical procedure where an incision is made to provide access to instrumentation, including the surgeon's hands, and a direct line of sight [CC0 1.0]. **Right** - a MIS approach known as laparoscopy, where multiple small incisions are made for each instrument including an endoscope. The image from the endoscope is then relayed to multiple displays around the operating theatre. [Samuel Bendet, 2005, Laparoscopic stomach surgery]

...after 90 days compared to the open alternative. This reduction in cost is attributed to shorter hospital stay and reduced readmission rates [14]. Lastly, there are aesthetic benefits from the smaller scars and the incision points can be placed at skin folds or the umbilicus, enabling the surgery to be almost scar-less [15].

However, MIS is technically more challenging than open surgery and require longer training times to achieve competency in the procedures [16]. Due to the use of several small incisions, the instruments must pivot about the insertion point referred to as a Remote Centre of Motion (RCM). This leads to the fulcrum effect where to move the end-effector of the instrument left the surgeon must move their hands to the right. Additionally, surgeons no longer have line of sight to the surgical site; instead the view from the endoscope is shown on a monitor to the side of the patient bed. As many procedures require two instruments, the endoscope is often held by another surgeon requiring careful communication to achieve the desired field of view. The majority of endoscopes are also 2D and this results in poor depth perception. In combination, such limitations mean that training for these procedures inevitably takes longer than for open procedures.

Robotic assistance has been proposed as a way of compensating for a number of these disadvantages. One such example is the da Vinci ® surgical robot
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Figure 1.2: **Left** - Shows the setup of the da Vinci surgical robot during surgery. **Top Right** - the view from one eye of the da Vinci endoscope showing the articulated instruments. **Bottom Right** - the manipulators used by the surgeon to control the instruments. [©2020 Intuitive Surgical, Inc.]

(Intuitive Surgical Inc., US), shown in Fig. 1.2. The da Vinci uses a teleoperation configuration, in which the surgeon sits at a console with a stereo viewer and two manipulators; the latter manipulators transfer the surgeons motion to the patient side system. This has four manipulators each constrained to a RCM, allowing three instruments and an endoscope to be controlled. In addition, the instrumentation features a small wrist at the tip that emulates the human wrist and provides increased dexterity. Overall, this approach: removes the need for the surgeon to actively compensate for the fulcrum effect; the stereo viewer allows surgeons to resolve depth; and the instrument is able to move with more stability and greater precision. However, the cost of using these systems is significantly higher than laparoscopic approaches despite having similar outcomes [17].

1.1 Fetal Interventions

Fetal interventions have not followed the same trend as more common surgical disciplines such as urology where there has been a high uptake in endoscopic procedures and robot assisted procedures. This deficiency may, in part, be due to fetal interventions being a newer field, with the first open procedure being performed in
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1981. That procedure was to release a blockage in the urinary tract, known as a congenital hydronephrosis, which causes increased pressure on the kidneys. It was corrected by introducing a catheter into the fetus bladder allowing the pressure to be relieved normally [18]. The access for these surgeries is very similar to a cesarean section, where a hysterotomy is performed to open the uterus. The fetus remains connected and dependant on the placenta but is moved to the opening of the uterus while warmed saline is introduced to maintain the volume and temperature of fluid within the uterus. Once the procedure is complete the intrauterine volume is returned to the preoperative level and the hysterotomy is closed [19]. However, this access approach is associated with an increase in fetal and maternal morbidity. This is due to the invasiveness resulting in risks of premature delivery, placental abruption and postoperative pulmonary oedema (attributed to the tocolytics given to prevent labour during surgery) [20]. Fetoscopic approaches are gaining traction due to the decreased maternal morbidity and comparable fetal morbidity [21]. However, the learning curve for fetoscopic approaches is significantly higher than open approaches. This results in hysterotomy based approaches remaining as the gold standard in terms of safety and efficacy for spina bifida operations; though fetoscopic procedures may be considered superior in terms of neuroprotection and maternal invasiveness [22]. This section will therefore introduce a number of fetoscopic procedures, discuss the benefits of performing these procedures with a minimally invasive approach and describe challenges faced in these minimally invasive procedures.

1.1.1 Fetoscopic Laser Photocoagulation

Fetoscopic Laser Photocoagulation (FLP) is a minimally invasive fetal intervention used to treat Twin-Twin Transfusion Syndrome (TTTS) [23]. TTTS is caused by inter-twin vascular anastomoses on a monochorionic placenta (a shared blood supply between fetuses) which causes an imbalance in the blood flow with one fetus receiving too much blood and the other too little. The condition can result in death or severe impairments of the fetuses. FLP is used to coagulate the vessels, thus severing the link and redressing the balance, by using a laser fibre through the working
channel of a fetoscope. During FLP, the surgeon uses a rigid fetoscope (straight or curved depending on the location of the placenta) in order to observe the placenta and, after visually determining the extent of the shared blood supply, selectively coagulate vessels. The fetoscope is inserted into the amniotic sac through a keyhole incision passing through the abdomen, uterus and amniotic membranes as illustrated in Fig. 1.3. This approach constrains the motion of the endoscope and limits the locations that can be treated. An additional challenge is that forces exerted at the entry port can cause undesirable weakening of the fetal membranes (and should therefore be minimised) [24]. Additionally, the surgeon must refrain from physical contact with the placenta as this may cause bleeding (obscuring visual observation) and lead to complications [23]. Therefore, maintaining the fetoscope and therapeutic laser at even distances from the tissue (to deliver appropriate laser power) is challenging and, depending on the position of the placenta, not always possible. Currently, in up to 33% of procedures, some shared vessels are missed or not fully
coagulated [23].

1.1.2 Fetoscopic Endoluminal Tracheal Occlusion

Fetoscopic Endoluminal Tracheal Occlusion (FETO) is used to treat Congenital Diaphragmatic Hernia, a condition that occurs when the diaphragm fails to form correctly. This causes the abdominal organs to form within the chest cavity preventing the lungs from forming fully [25]. Failure to correct this defect is associated with high mortality and morbidity, due to both the undeveloped lungs and an increase in blood pressure. Post-natal correction of the diaphragmatic defect has a survival rate of approximately 70%, but still suffers from medium to long term morbidity [26].

Previously, there was an attempt to repair the defect pre-natally through an open procedure. While safe for the mother, the technical difficulties encountered led to low fetal survival rate which restricted uptake of the procedure [27]. The FETO procedure is typically performed between 27-29 weeks under both ultrasound and endoscopic guidance. The fetoscope is introduced into the amniotic cavity under ultrasound guidance as illustrated in Fig.1.4. The insertion point is selected to be perpendicular to the nose of the fetus, and to provide a clear trajectory to the nose. The fetoscope is then inserted into the amniotic cavity and advanced through the nose to the trachea. Once positioned correctly, a balloon is inflated with saline to occlude the trachea and detached [28]. This occlusion causes an increase in lung fluid, increasing the intrapulmonary pressure promoting lung development [29]. The balloon is removed post-natally and the defect in the diaphragm is corrected.

1.1.3 Fetal Shunting

Genitourinary tract obstructions are among the most common congenital anomalies occurring in approximately 1% of pregnancies, of which Lower Urinary Track Obstructions (LUTO) are the most common [30]. Most commonly the obstruction is, in the posterior urethral valve (for male fetuses) and from the urethra atresia (for female fetuses), where the urethra blindly ends. This causes abnormal development of the urinary track which can also lead to under development of the lungs and renal damage and possible malformation. Fetal vesicostomy is an open procedure
1.1. Fetal Interventions

Figure 1.4: Schematic diagram of the FETO procedure for the treatment of Congenital Diaphragmatic Hernia showing the endoscope positioned at the trachea and a balloon being deployed through a working channel. Right shows deployment procedure of the balloon to obstruct the trachea and increase pulmonary pressure. [Image reproduced with permission from UZ Leuven, Belgium]

where an opening is made in the bladder through the abdomen to relieve the blockage. While this provides good post-natal results, it is associated with high maternal and perinatal morbidity which has prevented uptake [31]. However, a number of possible fetoscopic interventions to treat these abnormalities do exist; they include: vesicoamniotic shunting and fetal cystoscopy [32]. Vesicoamniotic shunting is an ultrasound guided percutaneous procedure used to treat LUTO (a schematic of the procedure shown in Fig. 1.5). A catheter is placed between the fetal bladder and the amniotic cavity, this bypasses the blockage and allows the bladder to drain. The shunt remains in place until delivery and is then removed postnatal [33]. In contrast, cystoscopy relies on endoscopic guidance to visually the obstruction visually, thus allowing the diagnosis to be confirmed. A curved endoscope is introduced into the fetal bladder and any blockage located (for example, a membrane structure over
1.1. Fetal Interventions

The urethral valve), can be fixed by removing or cutting the membrane; this may be achieved by using a number of methods including laser fulguration. However, if the blockage is not membrane-like a vesicoamniotic shunt is utilised instead. The main advantage of performing a cystoscopy is the diagnostic benefit, as it allows the blockage to be imaged. However, this procedure is technically more challenging and of higher risk than vesicoamniotic shunting [32].

1.1.4 Spinal Cyst Repair

Spinal dysraphism, the most common congenital abnormality of the central nervous system, is a cyst protruding from the base of the spine [34]. The cyst contains spinal fluid, nerves and part of the spinal cord, either damaged or deformed. While a non-lethal abnormality, it is a chronic and progressive disease that may cause paralysis (to varying degrees), and nerve damage. The surgical procedure to correct the abnormality involves closing the cyst to prevent damage to the exposed nerves and spinal cord, and stop the leakage of spinal fluid. The current approach involves closure using a two layer approach with the first layer being muscle tissue, or a patch made from an acellular-dermal-matrix, and the second being skin [35]. Traditionally, this procedure was performed postnatally, due to the high complexity and risk.

Figure 1.5: Schematic diagram of the Fetal Shunting procedure for the treatment of LUTO showing with the endoscope positioned to insert the shunt. [Image reproduced with permission from UZ Leuven, Belgium]
of the surgery. However, from a number of animal models and post-mortem studies detailing the continuous damage in utero, a coherent rationale for prenatal intervention was developed [36]. As discussed at the start of the section, the procedure has been performed as open fetal surgery and this is currently the preferred approach (due to the learning curve and associated affect on the efficacy of the procedures) [22]. A fetoscopic approach for the procedure was initially proposed however the technique was considered very complex. From the trial of 4 procedures, there was a 50% mortality rate with both the survivors requiring a ventriculoperitoneal shunt [37]. These outcomes led to the fetoscopic approach being abandoned for open approaches. More recently, fetoscopic approaches are once again being developed as a method of reducing the invasiveness procedure and improving outcomes for both the mother and fetus [38]. The fetoscopic approach is illustrated in Fig. 1.6. While similar to MIS approaches, one of the key differences is that the cannula is introduced through multiple interfaces to reach the surgical site, starting at the abdomen and penetrating the uterus. This approach involves acute technical difficulties; as the procedure requires multiple instruments and the surgeon can no longer dedicate a hand to manipulating the cannula. In contrast, the minimally invasive approach has the potential to reduce the abdominal and uterine scar for the mother and decrease the likelihood of preterm delivery from the reduction of membrane rupture rates [39].

1.2 Needs and Challenges

The instrumentation available and the technical challenges performing the aforementioned fetoscopic procedures are generally referred to as the main challenges facing fetal surgery [40, 41]. Integral to the problem is that instrumentation used in fetoscopic procedures is often adapted from its original purpose, for example instrumentation is often introduced using cannulas originally intended for vascular access. However, a range of purpose-made instrumentation is now available for fetal surgery. Fig. 1.7 shows a number of available fetoscopes and the quality of images obtainable. In relation to instrumentation designed specifically for fetal procedures,
a significant technical challenge is to minimise the instrument diameter. For fetoscopes, this may also involve a reduction in image quality in comparison to most other clinical endoscopes. It is also apparent that, due to anatomical constraints, some procedures necessitate a curved fetoscope. This then precludes the use of rod lens endoscopes and requires coherent fibre bundles or tip mounted cameras to be used instead, further decreasing the image quality. Additionally, and particularly relevant in terms of clinical utility, reducing the diameter of the instrumentation adversely affects robustness: meaning they are often quite fragile. Within the solely fetoscope based procedures such as FLP and FETO, the surgeon will often use their free hand to stabilise the cannula at the incision point. This is done, as discussed above, in an effort to reduce the force applied to the amniotic sac because excessive force can potentially cause the fetal membranes to rupture [24]. One of the potential improvements to minimally invasive surgery, including fetoscopic interventions, is articulation of the instrumentation at the tip (distal to the surgeon). This has been

Figure 1.6: Schematic diagram of the repair procedure for the treatment of spina bifida cystica showing the three entry points at the abdomen used to introduce an fetoscope and two rigid instruments. [Image reproduced with permission from UZ Leuven, Belgium]
implemented in laparoscopic surgery using manual instruments but also, and most notably through robotic instruments such as the da Vinci surgical system. However, the comparatively large instruments used in the da Vinci (< 5 mm) effectively precludes its use in fetal surgery. Robotic systems have significant potential to improve fetal interventions, for example, through the already mentioned addition of articulation at the tip but also through (a) stabilisation of the instruments outside of the body (proximal to the surgeon) and (b) controlling the forces exerted during surgery. Such systems and instrumentation could assist in reducing the stress on the uterine wall and potentially reduce membrane rupture rates, whilst also assisting in the precision of the procedures for example in the laser targeting for the FLP
procedure or in manoeuvring a fetoscope to the trachea in the FETO procedure. However, the intrauterine environment is a challenging application for robotics as the instrumentation must operate in amniotic fluid and within that environment are critical structures with which accidental contact must be avoided.

Fetoscopic interventions generally rely on white light imaging from the fetoscope and abdominal US for guidance during procedures. However, improvements to existing modalities and the introduction of new modalities could potentially improve the information acquired during the procedures. One such modality is Optical Ultrasound (OpUS), a method of generating and detecting ultrasound using optical elements rather than the conventional piezoelectric based elements [42]. Ultrasound is generated using pulsed lasers directed through a fibre to a highly optically absorbent coating which converts the energy to heat. The subsequent pressure increase within the coating then propagates as an ultrasound wave. The returned ultrasound waves are then detected using a Fabry-Pérot (FP) cavity at the end of a fibre beside the ultrasound source. The key advantage with the approach is the imaging probe is readily miniaturisable with imaging probes generally < 1 mm enabling it to be integrated within instrumentation. However, this provides only a single dimension of information known as an A-scan. Robotic instrumentation could enable the OpUS probes to be manipulated in such a way to acquire clinically useful information.

1.3 Contributions

This thesis presents the current work on improving fetoscopy using robotics. It directly addresses, through original research and design involving rigid and articulated instruments, the challenge of holding and manipulating the instrumentation outside the body and the integration of distal mechanisms at the tip of the instrumentation. Chapter 2 proceeds to review distal actuation mechanisms that have previously been applied to minimally invasive surgery focusing on miniaturisable continuum mechanisms that could be applied to fetal surgery. Chapter 3 presents an exploratory approach to the introduction of both distal Degrees of Freedom (DoF) and constrained proximal DoF to an endoscope. This was presented with a con-
centric tube mechanism with 2 DoF and the coupling of the distal mechanism to a redundant proximal manipulator constrained to a virtual RCM. The utility of the combined instrument is then demonstrated in ex vivo imaging of a human placenta, performing predetermined scanning trajectories and using the image data to determine the placenta geometry and enhance the field of view of the endoscope. Chapter 4 introduces robust control of the proximal DoF through the use of a motion planner. The design of a rigid endoscope with a stereo camera, light guide, working channel and Optical Ultrasound (OpUS) sensor is presented. The motion planner based control is assessed using an optical tracker to determine to over positioning accuracy and repeatability and additionally assessing the manipulators ability to follow the RCM constraint. The endoscope is demonstrated through dense scanning trajectories over a placenta phantom, in which the combination of the OpUS sensor and robot kinematic information can be used to construct a large area 3D pointcloud of the environment. Chapter 5 focuses on improving the distal capability of the system. A new mechanism design was produced which took into account the issues in the previous system presented in Chapter 3. The new model was employed as a scanning mechanism with a focus on trajectory repeatability. Firstly the mechanism is assessed using an Electromagnetic sensor; secondly, an OpUS is integrated through the channel of the instrument using a torque coil to maintain the orientation of the sensor while scanning. The mechanism is demonstrated creating 3D pointclouds through small area but fast scans. This approach allows scans to be generated faster than using only proximal manipulation. Additionally the information acquired from scanning known objects is used to improve the forward kinematics of the concentric tube mechanism, thus improving the quality of subsequent scans. The work in this thesis has contributed to the following publications:


Chapter 2

Flexible Manipulators for Fetal Interventions

As discussed in the previous chapter, distal DoF have been added to range of surgical tools for various procedures. These additional DoF are usually in the form of tendon driven articulated joints, most notably the da Vinci Endowrist™ instruments. Articulated instruments are difficult to miniaturise, mostly due to the complexity of the mechanism. The joints need to be structurally strong in order for the instrument to be stiff and capable of applying force.

Continuum manipulators are a class of robotics (like articulated or parallel), which rely upon the elastic properties of materials to deform the manipulator into a series of curves [43]. As the manipulator is generally considered a series of arcs, continuum manipulators are often divided into two categories: variable and constant curvature. This chapter will discuss some of the methods of introducing distal DoF to instrumentation, covering the actuation concept, modelling of the instrument, and the current applications. Following that a short review on previous work on manipulating imaging probes with both proximal and distal DoF. Finally, the clinical adoption of surgical robotics will be discussed.
2.1 Distal Actuation Mechanisms

2.1.1 Concentric Tube Manipulators

Concentric Tube Manipulator (CTM) are a series of precurved superelastic tubes aligned concentrically. The tubes apply internal moments on the adjacent tube causing the overall shape of the tubes to be an amalgamation of each. Through this mechanism the tubes can then be independently rotated and translated along the tube axis, thus controlling the overall shape. Concentric tube manipulators were developed concurrently but independently by Webster et al. and Dupont et al. \[44, 1\]. They were designed to be a form of a steerable needle that did not rely on the tissue interaction for control \[45\]. First presented in 2005 as an actuated needle guide for fetal blood sampling \[46\], the device featured two curved nitinol tubes of decreasing stiffness placed concentrically. The motion was restricted to rotation about each axis and translation of the entire device along the tube axis, allowing the curvature of the tubes to be controlled. As manipulators of this type rely on the internal moments for control, they do not rely upon tendons or muscles, and this provides a number of benefits. In particular: a constant working channel is provided throughout the manipulator; the tubes do not have joint limits; their capability and safety is an integral function of the tube design; all the actuation components are placed at the end proximal to the surgeon (i.e., not within the patient); and the tubes are compliant to external forces (Fig. 2.1).
2.1. Distal Actuation Mechanisms

The tube parameters are generally designed in pairs to assist in modelling the interactions [1]. The extremes of the interactions are referred to as dominating stiffness pairs (where the difference in the bending stiffness between the tubes is large) and balanced tube pairs (where the difference is extremely low). In a dominating stiffness pair the shape in the overlapping section will be dictated by the shape of the dominating tube. Whereas, in a balanced pair the shape in the overlapping section will be an average of the two tubes [44].

The initial actuation mechanisms for CTM generally features the tube joined to a actuator to provide the rotation of the tube, with the mechanism mounted on a linear stage for the tube translation [1]. Subsequent research focused initially on a sterilisable mechanism, that placed the unsterilisable actuators into a sterile drape and transferred the torque through the sterile barrier using an Oldham Couple [47, 48]. In addition, the superelastic materials normally used are compatible with magnetic resonance imaging scanners, allowing them to be used while imaging is active. This then requires that the actuation mechanism does not contain ferrous materials, therefore precluding the use of electromagnetic actuators. However, the use of pneumatic actuators has been demonstrated as a suitable alternative to electromagnetic actuators [49].

Typically, the tubes described above are made from nickel titanium, known as Nitinol. Alloys of this composition can have elastic strain limits of up to 11%, though typical properties are closer to 8%. Nitinol tubes are set to the desired curvature through a heat treatment process. The Nitinol is constrained to the desired shape in a jig and heated to its transition temperature, relieving the strain in the material and setting it into the fixed shape. The part is normally heated using a box furnace; however the time taken to heat and cool the sample to and from the transition temperature can introduce a number of undesired effects. The most noticeable of these is ‘spring-back’: where all the strain in the tube has not been relieved, causing the unconstrained form to attempt to revert back to the original shape. Although reversion is slight, it represents a significant limitation when producing a precision instrument. As an alternative to the box furnace, joule heating involves
passing an electric current through the material utilising the electrical resistance to heat the tube. This has the advantage of faster heat transfer to the tube and results in an appreciable reduction of spring-back. This process does, however, require a high-current power supply [50]. Recently an alternative to Nitinol and the shape setting process described above has emerged in the form of 3D printing. Allowing precurved elastic tubes to be fabricated using stereolithography and selective laser sintering with elastic limits of 5% and 20% respectively. This method also allows tubes to be tailored both to the procedure and the individual patient. Despite showing great promise, current printing techniques have a number of known limitations, specifically: the overall size and wall thickness they are capable of printing is limited and the materials have not been assessed fully in respect of their biocompatibility and resilience to sterilisation techniques [51].

Moreover, and as noted in relation to many other continuum robots, one of the main practical limitations of concentric tube manipulators is the need for complex mechanical models for accurate forward and inverse kinematics. The initial kinematic models focused on the bending moments applied by the tubes and used an idealised version of the Bernoulli-Euler beam equation [52]. However, while these models worked well in planar cases, when the tubes were rotated the model starts to fail due to torsion. Subsequent models then accounted for torsion along the straight section of the tube [44], and that development was closely followed by a model that accounted for torsion in both the straight and the curved sections [1]. However, in some cases it was noted that torsional energy accumulated during operation could cause the position of manipulator to suddenly change despite the input parameters remaining unaltered; the likelihood and severity of this adverse outcome increased in proportion to the number of tubes in the system. Whilst the tube configurations that cause this can be found in energy models for two tube manipulators, more work is needed when greater numbers of tubes are encountered [53]. The main disadvantage of existing models is they are computationally expensive and therefore slow to find solutions, and the models assume no external loading (e.g. from interaction with tissue). Closed form differential kinematic solutions have also been de-
2.1. Distal Actuation Mechanisms

Derived to provide velocity control which, within a visual servoing framework, enables fast computation of solutions. This approach has produced an average accuracy of \(< 1 \text{ mm}\) demonstrated in a eye-to-hand configuration (static camera) [54]. In an eye-in-hand configuration (where the manipulator holds the camera) the accuracy was measured at under one pixel [55]. Additionally, methods for computing the jacobian and compliance matrices for external loading have been presented, allowing trajectories to be formed but, in each case, the external load needs to be known [56]. That these models find single kinematic solutions, and do not take into account the environment in which the manipulator operates, are important considerations in respect of the anticipated surgical applications (such as neuro, vascular, fetal, etc.). Path planning for concentric tube robots has been demonstrated with the use of Rapidly-Exploring Roadmaps [57]. Using computationally efficient algorithms and highly parallelised processes, a collaborative teleoperation control scheme has been presented which performs rapid path planning and inverse kinematics allowing for online and stable control [58]. Alternatively data-driven approaches have also been presented that attempt to learn both the forward and inverse kinematics of a three-tube robot through artificial neural networks [59]. Using this technique, the forward kinematic network had an accuracy of 2.3 mm and 1.1° for position and orientation respectively. While the inverse kinematics network had an error of 4.0 mm and 8.3° for position and orientation respectively. However, the joint space of the robot was restricted for the training of both networks; this was due to the influence on the pose of the robot caused by the load from the EM sensor.

2.1.2 Tendon-Driven Continuum Manipulators

Tendon-driven Continuum Manipulators (TCM) operate in a similar way to the articulated variant. The tendons are placed through/along an elastic spine and fixed in a radial pattern at the same length along the spine. The placement of the tendons from the central axis of the spine determines the magnitude of the torque applied to the spine and subsequently the force the end effector can exert (Fig. 2.2) [2]. The shape of the spine is changed by pulling and releasing the tendons; unlike articulated manipulators that require four tendons (two antagonistic pairs), continuum manip-
2.1. Distal Actuation Mechanisms

Figure 2.2: Tendon driven manipulator at different curvatures [2]. [©2008 IEEE]

ulators can use only three to achieve the same two DoF. However, this placement leaves some instabilities in the manipulator and, due to the relatively small diameter of tendons, four tendons are more commonly used. Additionally, multiple 2 DoF mechanisms can be placed concentrically providing 2 additional degrees of freedom from the translation of each mechanism [11]. The elastic spine can be modified to changed the compliance of the instrument and the bending radius. Either through a single nitinol tube and cutting notches along the tube, with different cut patterns providing different properties. There are three main patterns used: symmetrical cut patterns allowing bidirectional bending, single side asymmetric cut patterns allowing bending in a single direction but with a reduced bending radius [60] and lastly double sided asymmetric cuts which allow bidirectional bending and increase the compliance of the instrument [61]. Alternatively multiple profiled tubes can be arranged concentrically to provide a self supporting but compliant mechanism [62].

TCM instruments are generally modelled in two stages: firstly the in-plane
curvature is determined, followed by the rotation of the plane to the parent coordinate system. Another option is that the system can be modelled using the concept of simultaneous rotation, describing the configuration using the Rodrigues formula [63].

As a further alternative to the use of elastic spines, multiple serial joints can be used to provide the necessary curvature. These mechanisms follow a serial articulated design but have a large and redundant number of joints on the same axis (though not necessarily adjacent) and actuate them with a single tendon pair [64]. Such systems can also group smaller sets of joints to actuators; this allows for more precise control, but only at the expense of additional complexity [65]. Generally, due to physical design constraints, these types of manipulators have relatively large diameters when compared with other continuum mechanisms.

The stability of TCM manipulators can, however, be increased through substituting the wire tendons for superelastic tubes or rods. The superelastic tubes are positioned around a central spine in the same radial spacing as the wire tendons. The tubes are fixed at the end of the manipulator and small disks are placed along the spine to maintain the requisite spacing. The tubes can be pushed and pulled - removing the need for antagonistic pairs - to control the shape of the central spine [3]. The use of superelastic tubes promotes greater stability and the dexterity can be increased further by placing additional tubes concentrically through each existing tube (Fig. 2.3) [66]. A further alternative involves manipulators that comprise of multiple structured planar modules which provide compliance [67]. This approach allows large bending motions decoupled from contraction motion, high back-drivability and low hysteresis. Quasi-static models are used to control these manipulators, they describe the elastic energy of the system as a function of bending allowing the current configuration to be found as the solution with the minimum energy. However, these models often ignore torsion and external forces [68]. The use of rods instead of tendons also presents the possibility of dispensing with a spine or backbone to support the mechanism. Instead these mechanisms can be assembled in a similar fashion to rigid link parallel mechanisms (where the load is shared
2.1. Distal Actuation Mechanisms

between the rods), thus improving the stability of the manipulators [69].

2.1.3 Pneumatic Artificial Muscles

Pneumatic Artificial Muscles (PAM) are the latest form of actuator to be applied to fetal surgery. There are many types of pneumatic actuators though the most commonly used as a distally placed actuator is the McKibben muscles. These muscles feature a thin membrane wrapped in a woven shell with ferrules at each end. When inflated, the thin membrane expands causing the shell to expand with it but simultaneously contract in length thus reducing the distance between the ferrules (Fig. 2.4)[70]. McKibben actuators exhibit a high power to weight ratio and, due to the compressibility of gas, are compliant. As McKibben muscles operate through a change in length, they are deployed in a similar way to tendons; however, due to their small physical size, they are not required to be positioned proximally. The distal placement of the muscle both reduces the effect of elasticity in the tendons and increases the flexibility in the manipulator up to the muscle. Artificial muscles
2.2 Manipulation of Imaging Probes

The robotic manipulation of imaging probes has different requirements to interventional instrumentation (such as forceps or needle drivers). For the majority of imaging modalities, the instrument does not need to be capable of applying large forces as tissue interaction is minimal.

2.2.1 Scanning mechanisms

A number of manipulators already exist which are not designed for direct control by the surgeon but, instead, provide repeatable motions with the aim of improving the imaging. Typically, these are employed with imaging modalities that provide...
low-dimensional information or have a limited field of view. The three examples illustrated in Fig. 2.5, demonstrate the different applications of scanning mechanisms when attempting to:

- increase the dimensions of data acquired
- improve the acquired data
- combine multiple modalities

The first example is a scanning fibre endoscope [5], where 1D information is utilised to develop a 2D image. This approach uses piezo actuators to vibrate a single optical fibre at its resonant frequency. Applying sinusoidal waves across 2 axes causes the fibres to follow a spiral pattern. The single fibre delivers laser light (through a lens assembly at the tip) to the imaging surface. The reflected light is then captured through a fibre bundle around the outside of the actuators and lens. Through a calibration sequence, the recorded reflected light with the motion of the fibre can be reconstructed into a colour image. This allows a flexible colour endoscope to be produced with a 1.2 mm diameter, a 500 pixel diameter resolution at 15 Hz.

The second example is an endomicroscopy scanner [6]. This is designed to improve the acquired data by increasing the field of view. Endomicroscopy is a form of microscopy that often relies on fibre bundles to miniaturise the distal section of the microscope, thus allowing it be introduced within the body as part of a minimally invasive procedure. These imaging devices are most commonly applied to providing in vivo assessment of tissue pathology because they provide high resolution (1 – 4 µm) images [72]. However, the clinical use of the high resolution images produced is inhibited due to the relatively low field of view achievable (≤ 600 µm) [72]. Such scanning mechanisms have been applied endomicroscopy in an attempt to improve the field of view by manipulating the probe in a known and repeatable pattern and then combining the recorded images. To counter trajectory errors introduced from tissue deformation (as the probe must be in direct contact with the tissue), a visual servoing approach is introduced to adjust the trajectory
2.2. Manipulation of Imaging Probes

Figure 2.5: Top Row - Scanning fibre endoscope which uses piezo elements to scan a fibre emitting white light in a spiral pattern, the light reflected from the surface is detected and an image can be formed of the entire scan pattern [5].

Middle Row - Endomicroscopy scanner which uses two motor driven cams to manipulate a probe in 2D, visual feedback is then used to adjust trajectories while scanning to improve the acquired mosaics [6].

Bottom Row - a scanning mechanism that combines endomicroscopy and ultrasound, using a tendon driven rail to scan the microscopy probe across the length of the ultrasound transducer [7].
while scanning. Images are then combined using visual information but each image is initialised in the estimated kinematic position.

Thirdly, a scanning mechanism that combines endomicroscopy with ultrasound has been developed [7]. Endomicroscopy provides high resolution images at a microscopic level on the tissue surface, while the ultrasound provides depth resolved information at a macroscopic level. The mechanism uses a tendon-driven rail to scan the microscopy probe along the length of the ultrasound transducer: this brings the images from the two modalities into a similar level (through mosaicing the microscopy images). A calibration phantom is then used to determine the offset between the ultrasound and microscopy probe, enabling information within the same coordinate system to be displayed.

2.2.2 Proximal Manipulation

The use of manipulators outside of the body (proximal to the surgeon) can improve the stability of the imaging probe motion; there are also a number of established modalities that can provide internal imaging whilst positioned outside of the body, such as ultrasound and x-rays. Ultrasound is often targeted for robotic manipulation due to the difficulties in manual manipulation of the probe, especially over long periods of time [8]. One example of this is a robotic ultrasound manipulator, utilised specifically for fetal ultrasound examinations. The system employs a dual arm system with 17 DoF; this is used to manipulate two ultrasound probes independently. The system is shown in Fig. 2.6. One of the key aspects of this arrangement is that each of the safety critical joints (ones that primarily control the translation of the probe) has a mechanical clutch, and this prevents excessive force from being applied to the patient: the clutch disengages the link from the driven joint when the load exceeds a predetermined threshold. However, some commercial robots have been developed to execute ultrasound manipulation tasks, such as the UR5 (Universal Robots A/S, DK) [73]. While intended to solve the same issues as identified in the previous example (the manipulation difficulties noted), the approach focuses on implementing a low-level controller, rather than on the mechanical design of the manipulator. The system uses a force/torque between the robot end-effector and
2.2. Manipulation of Imaging Probes

Figure 2.6: Left - The custom made dual probe ultrasound manipulator scanning a patient's abdomen [8]. Right - ARTIS Pheno® (Siemens Healthineers AG, DE) a 6 DoF robotic manipulator holding a C-arm allowing greater positioning control between the imaging system and the patient.

the ultrasound probe to feed the current wrench applied to a haptic controller, held by the sonographer; the sonographer then controls the force applied to the patient. Both of the presented systems have been designed for tele-operation by the sonographer, who controls the ultrasound probe through a haptic controller in free space (controlling all 6 DoF). Visual servoing approaches (as part of which image information can be taken into account during the manipulation of the probe [74]) have been proposed as a method of improving imaging probe manipulation. Additionally, the combination of imaging and force information has allowed for the patients motion to be compensated for while maintaining a consistent view [75].

It is apparent that robotics brings an additional advantage for imaging modalities placed outside the body, particularly the capability to manipulate large and heavy equipment, exemplified by the ARTIS Pheno® (Siemens Healthineers AG, DE) [76]: this is a C-arm attached to a 6 DoF robotic manipulator allowing more control on the positioning of the C-arm (as shown in Fig. 2.6).

For imaging probes that require access to the targeted organ/area, the proximal manipulator must first be constrained to a Remote Centre of Motion (RCM) at the insertion point for the instrumentation. There are two approaches to achieving this constraint: mechanically or through software (both are illustrated in Fig. 2.7). Mechanisms have a notable advantage in terms of reliability and robustness, as the RCM is mechanically enforced. However, the position of the RCM is fixed and
2.2. Manipulation of Imaging Probes

Figure 2.7: **Left** - the MiroSurge which uses a 7 DoF constrained to a virtual RCM [MiroSurge, DLR (CC-BY 3.0)]. **Right** - a patient side manipulator of the da Vinci surgical robot that is mechanically constrained to a RCM

This requires additional mechanisms to position the RCM initially [77]. In comparison, software enforced RCM utilise articulated manipulators then, using geometric knowledge of the robot and instrumentation, constrain the motion of the actuated joints. This allows the manipulator to move to the RCM and potentially change the RCM according to external motion. It is noted that the use of articulated manipulators introduces the possibility of singularities within the constrained workspace, and the accuracy of these are likely to lower from the longer kinematic chains [78].

Both approaches to constrain the system to the RCM have been applied to the control of rigid endoscopes for use in laparoscopic surgery. A number of surgical tasks require bimanual manipulation, and this can be particularly challenging because an endoscope is required in order to visualise the surgery. Often one surgeon will hold the endoscope while another hold the instrumentation required for the procedure. In operational surgical environments, this setup imposes a significant constraint on available space around the patient (with two surgeons working in the same space); it requires clear communication between the surgeons; the endoscope does not require constant manipulation, which leads to physical burden on the surgeon holding it. Robotic endoscope holders have been presented as a potential solution for this, however one of the main difficulties in this is the control interface for the surgeon. The first generation design was the AESOP™(Computer Motion Inc., US): a 7 DoF robotic endoscope holder capable of moving the endo-
scope during surgery by voice commands [79]. However, the technology utilised provides only coarse commands for positioning and motion, limited to: up, down, left, right, in, out and stop. Subsequently, interfaces that provided finer control of the endoscope were introduced, such as eye tracking, where the endoscope follows the motion of eyes on the endoscope display [80].

2.2.3 Distal Manipulation

For the manipulation of imaging probes within the body, the majority of systems available now rely on both proximal and distal manipulation. Flexible endoscopes, for endoluminal procedures, require the least amount of proximal manipulation, with only insertion into the lumen; tendon actuation is used to adjust the curvature of the tip and therefore the distal orientation. This results in limited accessibility and often substantial discomfort for the patient. To address such limitations, different manipulation strategies have been proposed to control the endoscopes within the body; two such approaches are magnetically actuated and distal locomotion [81] (examples to these approaches shown in Fig. 2.8). Magnetic actuation relies on the control of magnetic fields external to the body to induce forces on a magnet at the tip of the endoscope. The external magnetic fields are either generated from fixed coils positioned about the patient, rotating permanent magnets or permanent magnets controlled by an external articulated manipulator [9]. For distal locomotion, novel mechanisms are implemented at the tip, using the interaction with the lumen to propel the endoscope through the lumen. Three actuation systems have been presented: continuous tracks, legged and pneumatic. Continuous tracks use miniaturised tracks around a cylindrical housing driven by internal motors. They were considered appropriate for surgical applications as they could be used on planar surfaces and within a lumen [10]. The legged approach employs two sets of legs at either end of the tip, positioned radially about the tip. Through the utilisation of two motors within the endoscope (one to drive each set of legs), the endoscope can move forwards or backwards along the lumen [82]. Alternatively, the pneumatic approach incorporates two balloons, one at the tip and another that can travel along a length of the endoscope; the rear balloon reduces the insertion forces while the
2.2. Manipulation of Imaging Probes

Figure 2.8: Left - Magnetically actuated endoscope, where an External Permanent Magnet (EPM) is held by a robotic manipulator to guide an endoscope with an Internal Permanent Magnet (IPM) through a lumen, in this case a colon [9]. Right - Distal locomotion of an endoscope provided by tracks along each side of the endoscope [10]

The manipulation of imaging probes within in the body can be ergonomically challenging, for example in the use of a microscopy probe to scan across the surface of an organ [86]. Visual servoing can be used to assist in the motion using an external imaging sensor, in this case a stereo endoscope. In this approach a marker is attached to the microscopy probe while being held by a robotic manipulator. The stereo endoscope is used to reconstruct the surface of interest and a trajectory is generated accounting for the field of view of the microscopy probe (and they need for overlapping areas). The probe then executes the trajectory using the pose of the
marker, tracked with the endoscope as feedback for the motion but, additionally, as a pose estimate used to assist in generating a mosaic of the entire scan. Importantly, as the scan motion has been tracked by the endoscope, the mosaic can then be overlay directly on the endoscope image.

### 2.3 Clinical Adoption

Surgical robots first started to be used clinically in the mid-1980s, with two robots originating from North America. Initially, there was the Arthrobot: a custom-made system to position the patient’s leg during orthopedic surgery through voice commands by the surgeon [87]. The second was a modified industrial robot, registered to a Computed Tomography (CT) scanner, and used to position a needle guide for brain tumour biopsies. A calibration procedure of the system was presented to re-calculate the kinematic parameters of the robot, which brought the tip positioning accuracy to 1 mm, allowing effective use in biopsies [88]. These robots focused on positioning outside of the body (a constraint possibly due to approval mechanisms in place at the time and for patient safety reasons). Shortly after Arthrobot, the PROBOT was presented to assist in performing trans-urethral prostate resection. The system took a scan of the prostate using an ultrasound transducer, by which means the surgeon marked on each image the sections of the prostate to be removed. The system constructed a 3D model of the prostate and the section to be removed. The cutting trajectories were then planned and executed in a semi-autonomous manner allowing the surgeon to stop the robot and adjust the cutting parameters [89].

In the 1990s two companies started developing robotic surgical systems, Intuitive Surgical (US) and Computer Motion (US). Both companies focused on a teleoperation-based system, where the surgeon could be remote to the patient (i.e. different room, different building or different country). Computer Motion first developed AESOP (discussed above in §2.2.2) and, this was cleared for clinical use in 1994. However, the full surgical systems, da Vinci and Zeus (from Intuitive Surgical and Computer Motion respectively) did not gain FDA clearance until 2001; each
2.3. Clinical Adoption

The use of surgical robots has now been approved for use for almost 20 years (since 2000 for the da Vinci surgical robot) [93], yet continuum robots have only recently (and in limited applications) been adopted in the clinical environment. The da Vinci SP © (Intuitive Surgical Inc., US) was one of the first to be deployed which, rather than using multiple incision points, introduces an articulated camera and up to three multi-jointed instruments (as shown in Fig. 2.9). In addition, two tendon driven continuum platforms were introduced in 2018 and 2019: the Monarch ™ platform (Auris Health Inc., US) shown in Fig. 2.9 and the Ion ™ (Intuitive Surgical Inc., US). Both are catheter style instruments designed to navigate deep within the lungs, reaching significantly further into the lung than conventional manual tech-
Catheter surgical robots have been previously marketed for intravascular interventions through two companies Hansen Medical Inc. and Corindus Vascular Robotics Inc.. (Hansen, however, was purchased in 2016 by Auris Health Inc. and the intravascular systems were discontinued). The systems are teleoperation based, meaning surgeons can control the guide wire and rapid exchange balloon and stent delivery systems behind a radiation shielded barrier. On the patient side, a robot arm, holding the instrumentation in motorised drives, facilitates the use of standard instrumentation. Current evidence shows the robot procedures are safe and reliable, and often reduce the surgeons exposure to radiation by > 95% [94].

Lastly, while the previously described surgical systems have all been teleoperated systems (where the clinician directly controls the system during operation), there are two other interfaces used to control surgical robots: supervisory control and shared control. Supervisory control generally provides planning interfaces for the surgeon but the actual operation is autonomous, with the surgeon having the ability to start and stop the procedure. This method of control is exemplified by the CyberKnife®(Accuray Inc., US) shown in Fig. 2.10. The CyberKnife is a robotic radiosurgery system which is comprised of a linear accelerator mounted to a 6 DoF
robotic manipulator, a stereoscopic X-ray system, with each source placed orthogonal to the patient; an optical tracking system and a positionable bed for the patient which is held by a second robotic manipulator. The treatment is planned by the clinician on preoperative images. The system then performs a registration between the preoperative imaging and the patient using the stereo X-ray system (which allows the treatment trajectories to be known in the robot coordinate system). The imaging systems are used during treatment to track the patient’s breathing and adjust the robots motion accordingly. This configuration provides an overall targeting accuracy on a target undergoing respiratory motion of $\leq 1.5\ mm$ [95]. Shared Control system relies on a collaboration between the robot and the surgeon, and this case can be demonstrated by the Mako\textsuperscript{TM} (Stryker Corp., US) shown in Fig. 2.10. The Mako system is designed for joint replacement orthopaedic surgeries; it consists of a planning system where the surgeon, using preoperative images, plans the insertion of the replacement components for the procedure. The other element of this system is a robotic arm, designed to hold the range of surgical instruments required by the surgeon. Before the procedure starts, a registration step is undertaken to align the preoperative plan to the patient. During the procedure, the robot will be compliant allowing the surgeon to control the instruments similar to a non-robotic procedure. However, the robot will prevent the tool from extending beyond the planned motions. This approach, in comparison to exclusively manual surgery, provides increased accuracy and repeatability of the component placement [96].

### 2.4 Summary

Continuum manipulators commonly work from similar principles: that is to say, applying a moment to an elastic spine causing it to deform. This allows for the creation of manipulators using relatively simple structures, in comparison to articulated mechanisms that require joints. This allows a more straightforward approach for the manipulator to be miniaturised. Tendon and fluidic actuators have separate backbones and driving elements; a different principle to CTM, which use the backbone as the driving element. Combining the backbone and driving elements provides the
advantage of a constant working channel through the device. However, the diameter of the working channel reduces as the number of DoF increases. In contrast, and within known limits, tendon or fluidic manipulators can maintain a constant diameter while increasing DoF through the utilisation of a stepping radial pattern of the tendon path. The majority of the work on CTM is focused on the control of multiple-tube robots (i.e. 3 tube mechanisms with 6 DoF); often the instrumentation developed for testing this work uses a wire as the innermost tube (to simplify a portion of the control). However it is uncommon to find instrumentation that does not rely solely on the CTM for manipulation but couples the motion with an additional manipulator positioned proximally and, additionally, focuses on the integration of end-effectors (such as grippers and imaging sensors) with the manipulators.

This thesis will proceed to investigate the combination of proximal and distal actuation methods to manipulate imaging probes. The combining of actuation methods has the potential to improve the stability and precision of the motion from the proximal side, while the dexterity of the instrument can be increased through the distal side. Considering the proximal side, while parallel robots can display significantly higher stiffness and finer motion precision, articulated robots can provide accurate and dynamic motion and require only a small footprint for a comparatively large reach. Additionally, due to the RCM constraint on the proximal side, the manipulators often provide redundant DoF allowing greater control in the overall pose of the robot. Whereas, at the distal side, the use of CTM will be explored in relation to the low diameters possible from the structure of the manipulators. Additionally, as the purpose of the instrument is to hold imaging probes, the manipulator does not need to be capable of applying force. While one of the limitations of CTM is the need to check the stability of the tube configurations to prevent tube ”snapping”, this generally affects CTM with 2 or 3 tubes. In order to control the orientation, the instrument only requires a single curved tube, allowing it to take advantage of low diameters and high repeatability, while avoiding the issues in stability and complex kinematic models that afflict CTM with more curved elements.
Chapter 3

Coupled Proximal and Distal Manipulation for Stable Fetoscopy

3.1 Introduction

As presented in §1.1.1, FLP is a minimally invasive procedure used to treat TTTS. The procedure involves a fetoscope with a therapeutic laser being introduced into the uterine cavity and coagulate vessels on the placenta (which are part of a detrimental shared blood supply between the fetuses) [23]. Typically, this is a technically challenging procedure as current instrumentation is rigid and constrained to a remote centre of motion at the incision point. These limitations prevent the placenta from being viewed with a constant orientation compounding the difficulty of (a) the identifying the shared vasculature and (b) the aiming of the laser for consistent coagulation. To address the challenges of FLP it is possible to increase the dexterity and stability of the fetoscope by introducing actuated components to the fetoscope design. Greater dexterity of the tip of the instrument can facilitate observing and delivering therapy to the anterior placenta [40] while stability can be controlled by an articulated arm which constrains movement around a RCM at the incision point [97]. A major challenge for delivering robotic actuation to fetoscopic instrumentation is size. As discussed in Chapter 2, continuum mechanisms have been used in surgical robotics to facilitate smaller diameter instruments and while increasing the number of DoF. However, these mechanisms are often applied to single port
surgery, intravascular or neurosurgery, where the mechanism is fixed outside the body (proximal to the surgeon) and controls only the movement from within (i.e., distal to the surgeon). An example of continuum manipulators would be concentric tube robots [44, 1], a manipulator category that uses a series of precurved tubes positioned concentrically (the overall shape of the tubes being controlled by translating and rotating each tube). Concentric tubes have been demonstrated with six DoF, though they often have a comparatively low position accuracy (3 – 4 mm) when employed without the use of external sensors [54, 98]. In comparison, more established articulated mechanisms such as those used in the da Vinci surgical robot (Intuitive Surgical, US), utilise both proximal and distal motion from the parallel linkage and the wrist joints separately. This approach results in seven active DoF, four proximal and three distal (one of which being the end effector). Instruments featuring continuum mechanisms with both distal and proximal motion, have been presented previously using concentric tube robots coupled with a passive proximal arm for single port prostate surgery [99]. In this application, the concentric tube robot is inserted through an endoscope with a working channel, thus allowing the endoscope to be manipulated with the passive arm and the robot to be manipulated relative to the endoscope. However, it is important to note that the proximal arm was mainly used for the initial positioning of continuum mechanism and the co-manipulation of the device was not fully explored. Additionally, a concentric tube mechanism has been integrated into a da Vinci instrument an arrangement which provides three distal DoF along with the da Vinci three proximal DoF constrained to a mechanical RCM [100].

This chapter is an extension of work presented in [101] and presents the design and control of a concentric tube manipulator; one that provides two DoF at the distal end of the instrument and allows the orientation of the tip to be decoupled from the position controlled at the RCM. The design of the instrument is broken down into three main components: mechanism design, where a generic mechanism design is presented allowing it to be customised according to the tube parameter, while maintaining a small form factor; the tube design, where the tube parameters
are determined from the clinical requirement and workspace available; and lastly the integration between the concentric tube manipulator and a 7 DoF robotic arm constrained to a RCM incorporating a stereoscopic micro Complementary Metal-Oxide-Semiconductor (CMOS) camera at the tip of the concentric tube manipulator. Fig.3.1 shows a Computer Aided Design (CAD) drawing of the complete system.

§3.3 presents the how the instruments are controlled individually and when coupled together, followed by the validation of the kinematic model using an EM tracker in the experiments and results section. The experiment and results section presents the experimental setup of the distal instrument, a method for measuring the arc parameters of the tube, the accuracy of the mechanism and overall accuracy of the distal instrument. The workspace, accuracy and repeatability of the coupled mechanism is then demonstrated through tracking the tip of the nitinol tube using electromagnetic sensors in addition to visual odometry.
3.2 Instrument Design

For the purpose of decoupling the orientation of an imaging sensor from the position, the distal side of the instrument requires 2 DoF, rotation around the tube axis and extension along the tube axis. The distal mechanism was designed to be easily adaptable to tube parameters while retaining a small and compact form. The design utilised a carriage to transfer the torque from the actuators to the inner tube, similar to a design presented in previous work [48]. Additionally, the use of a carriage allowed the actuators to be fixed in place on the mechanism. The actuation components of the mechanism consists of a leadscrew and square shaft of equal length, positioned parallel to the tube axis and constrained to rotation along their axis by bearings placed either end of the mechanism. The leadscrew performs the translation of the carriage through a nut fixed to the carriage; while rotation of the tube is achieved through the square shaft, running the length of the mechanism and a gear with a square bore fixed to the carriage. When the square shaft is rotated it drives a gear fixed to the tube housing which in turn, rotates the inner tube in the opposite direction. Additionally, the square shaft constrains the translation axis of the carriage, acting as the linear guide required for the leadscrew to operate. At the distal end of the mechanism, the stainless steel outer tube is fixed in place concentrically to the curved nitinol inner tube. The design of the carriage provides a mechanical interface, allowing the tubes to be swapped without a requirement to dismantle the mechanism. The inner tube is joined to the carriage through two bearings and a spur gear, the same size as carriage gear. A CAD drawing of the assembled carriage and driving mechanism is shown in Fig. 3.2.

The mechanism houses the carriage and constrains the lead screw and square shaft but does not hold the motors. Instead the end of the leadscrew and square shaft has a female section of a hexagonal coupling. This allows the design to be motor agnostic (i.e., any motor can be used if coupled to a male hexagonal connector), allowing motors to be chosen according to the tube parameters and required torque. The motors used are mounted together onto a plate (the width of which matches the mechanism housing) and fixed using four machine screws positioned in each corner.
Figure 3.2: (a) Diagram of the carriage with: the lead screw nut (yellow); tube holder with the bearings and gear (blue); and rotation component with the square shaft couple, bearings and gear (green). (b) Assembled mechanism without carriage; (c) Assembled mechanism with carriage attached, also shows the female connector for the motor couple.
3.2. Instrument Design

The mechanism housing and carriage are fabricated using a Ultimaker 2 (Ultimaker B.V., NL) 3D printer, printing Polylactic Acid (PLA) (RS Components, GB). The Ultimaker is a Filament Deposition Modelling (FDM) printer which provides high accuracy on the X and Y axes from the use of highly repeatable stepper motors and a constant layer thickness making it suitable for the main components in the carriage and mechanism. While PLA is thermoplastic polymer made from renewable resources, displaying similar properties to conventional thermoplastics (such as Acrylonitrile Butadiene Styrene (ABS)) but with the addition of being biodegradable, it is commonly used for prototyping and single use components. However FDM printing is unable to fabricate small detailed parts or print large overhanging sections. Therefore the parts that required small detailed features or a low tolerance joints, which included both the inner and outer tube housings, were printed using an Ultra 3SP (ENVISIONTEC INC, US). The Ultra 3SP uses a printing technique called Scan, Spin and Selectively Photocure, that works in a similar way to stereolithography, using a photocurable resin and a high resolution projector to selectively cure areas of resin with high accuracy.

The mechanism was actuated using Dynamixel MX-28 motors (Robotis Inc., US), as these motors have encoders and inbuilt controllers allowing direct position control and can provide up to 2.5 Nm of holding torque. The motors communicate using Transistor - Transistor Logic (TTL) allowing the motors to be connected together through a serial connection with communication to the main controller through a Universal Asynchronous Receiver/Transmitter (UART).

3.2.1 Tube Design

FLP requires access to the main area of the placenta at a approximately 26 weeks gestation age, where the placenta has a diameter approximately 200 mm. Current literature states the required length and diameter of fetoscopes is 200 – 300 mm for full access to the uterine cavity and 1.0 – 3.8 mm respectively [40]. While the laser has an optimum effect 10 mm from the placenta at a 90° angle. Utilising the generic mechanism described in the previous section, the tubes can the be designed according to the clinical application, with the tube design influencing the length of
mechanism, square shaft and lead screw.

From this, the tube parameter for the instrument can be determined. There are four main parameters which can be used to drive the remainder:

- outer tube length
- outer tube diameter
- inner tube arc angle
- inner tube bending radius

The outer tube length determines the straight reach of the instrument which will be set to 200 mm, as shown in the literature [40]. The outer diameter determines the cannula size needed and should be minimised as much as possible while allowing a channel for the required end-effector and not exceeding the present maximum of 3.8 mm. The inner tube arc angle represent the maximum angular deflection between the outer tube axis and the inner tube axis. As the instrument length is set as 200 mm and the placenta diameter is considered to be approximately 200 mm, if the port placement is centred above the placenta the arc angle would be \(\text{arcsin}(\frac{100}{200}) \approx 24^\circ\) in order to allow the tip to remain perpendicular to the placenta across the placenta. To add some redundancy the arc angle can be set as 35°. The inner tube bending radius is the curvature of the curved section of the inner tube; this determines the length of the curved section and the length of the mechanism housing. The bending radius is generally driven by the components in the working channel and the maximum curvature they can adhere to. For many fibre based components including therapeutic lasers, this is approximately 30 mm. However a straight section of approximately 7 mm is needed at the tip to provide a constant profile allowing an interface for components of the end-effector to join to. Following these constraints, and additionally the availability and compatibility required between the tubes, the tubes are designed to have the following parameters:

- outer tube
  - outer diameter: 2.45 mm
3.2. Instrument Design

- inner diameter: 1.8 mm
- length: 200 mm

• inner tube

- outer diameter: 1.59 mm
- inner diameter: 1.4 mm
- length: 250 mm (rounded sum of the arc length, the outer tube length and the carriage thickness)

The inner tube then needs to have a 7 mm straight section followed by a curved section with a deflection of 35° and bending radius of 30 mm.

The inner tube is made by shape setting the nitinol tube [102]. This is a multi-stage process. Firstly, a jig with the desired shape is made in aluminium using a CNC mill. The nitinol tube is then constrained to the desired shape within the jig, and then placed into a furnace at 550° for 20 minutes. The jig and tube is then immediately quenched in room temperature water to avoid ageing effects. Typically, the new unconstrained shape of the inner tube is very similar to the desired shape; but springback is inevitable and results in the inner tube having a larger bending radius and lower angular deflection than desired. Therefore a direct measurement of the arc parameters are required for accurate control, and this will be discussed in section 3.4.

3.2.2 End-effector Design

A housing for an electromagnetic tracking sensor (NDI Aurora, Northern Digital Inc., CA), held concentrically at the tip of the instrument is required for accurate measurement of the tube pose. The sensor is a cylindrical probe measuring 0.8 mm in diameter and 9 mm in length, and can display the full pose in 6 DoF. The sensor is mounted to the inner tube through a simple collar component to ensure the sensor is concentric in the tube. The second end-effector is a Naneye Stereo camera (AWAIBA Lda, PT) that has a rectangular footprint of 2.2 \times 1.0 mm and therefore a bounding diameter of 2.42 mm; and a length of 1.6 mm. A stereo camera was
selected due to the potential of using the stereo information to derive geometric information on the environment, for example distance to the tissue surface. A prototype mount for the camera was made featuring a rectangular channel to hold the main body of the camera and a cylindrical tube that fits concentrically into the inner tube (Fig.3.1a). This solution is temporary because it effectively occupies the entire working channel; thus, there is no room to accommodate a light source or a therapeutic laser, for example. Both end-effector housings were printed using a Connex1 Objet500 (Stratasys Ltd., US), due to the low wall thickness of the parts required (< 0.150 mm). These printers rely on jetting a photocurable resin from the print head while also curing the resin at each level, allowing multi-material and high resolution parts.

3.2.3 Proximal Manipulation

An articulated arm (KUKA LBR iiwa 7 R800, KUKA Gmbh, DE) is used for the proximal manipulation of the instrument. The arm is a 7 DoF robot with torque sensors on each joint. The distal manipulator was fixed to the proximal manipulator through the motor coupling. Because the actuators were already detachable from the mechanism housing, a suitable means of mounting the manipulators together was readily available. The actuators were fixed to the flange of the proximal manipulator and this arrangement allowed the mechanism to be attached with a single machine screw (through the base of the mechanism housing to the flange), as shown in Fig.3.1.

3.3 System Control

3.3.1 Distal Mechanism Control

The joint control of the instrument is implemented using the Robotic Operating System (ROS) [103]. First, the mechanism control is implemented as a node using the Dynamixel SDK[104] in C++, providing a publisher and subscriber for the mechanism in joint space (i.e., orientation and length of the tube).
3.3. System Control

3.3.2 Coupled Control

As a minimally invasive procedure, FLP requires the instruments to be constrained to a RCM; this is due to the instruments accessing the surgical site through a small incision in the abdomen. An RCM constrains the motion to 4 DoF: translation and rotations along the instrument shaft (Z axis), and rotations about the X and Y axes. However, the Z axis rotation is restricted as the distal actuation mechanism provides continuous rotation along the Z axis while the proximal manipulator has a limited range of motion and velocity. A number of control models, with increasing levels of complexity, were presented in §2.1.1. However, and as discussed in the previous chapters, these models are generally made for instruments with multiple actuated curved tubes. As the presented instrument has only a single actuated tube and the outer tube has a significantly greater stiffness than the inner tube, the domination stiffness model can be used.

The domination stiffness model is dependant on the bending stiffness of the outer straight tube to be significantly larger than the stiffness of the inner tube [1, 44]. It assumes that the inner tube will solely adapt to the shape of the outer tube.
while overlapping. This allows the overall shape of the instrument to be modelled as the shape of the outer tube, followed by the shape of the inner tube with the length extruded from the outer tube. With the inner tube having the fixed parameters, $l_{\text{ArcLength}}$, the length of the curved section of the tube; $l_{\text{TipLength}}$, the length of the straight section at the tip extended from the outer tube; $k$, the curvature of the arc equivalent to the inverse of the bending radius; and $l_{\text{Outer}}$, the length of the straight outer tube. While the controllable parameters, length extended from outer tube, $p$ and bending plane given by the orientation of the tube, $\psi$ (Fig. 3.3). This can be represented as:

$$l_{\text{Tip}} = \begin{cases} p, & \text{if } p \leq l_{\text{TipLength}} \\ l_{\text{TipLength}}, & \text{otherwise} \end{cases} \quad (3.1)$$

$$l = \begin{cases} 0, & \text{if } p \leq l_{\text{TipLength}} \\ p - l_{\text{TipLength}}, & \text{otherwise} \end{cases} \quad (3.2)$$

$$Shaft_{\psi} = \begin{bmatrix} \cos(\psi) & -\sin(\psi) & 0 & 0 \\ \sin(\psi) & \cos(\psi) & 0 & 0 \\ 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad (3.3)$$

$$\psi_{\text{cur}} = \begin{bmatrix} \cos(kl) & 0 & \sin(kl) & \frac{\cos(kl) - 1}{k} \\ 0 & 1 & 0 & 0 \\ -\sin(kl) & 0 & \cos(kl) & \frac{\sin(kl)}{k} \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad (3.4)$$

$$\text{cur}_{\text{tip}} = \begin{bmatrix} 1 & 0 & 0 & 0 \\ 0 & 1 & 0 & 0 \\ 0 & 0 & 1 & l_{\text{Tip}} \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad (3.5)$$

$$Shaft_{\text{Tip}} = Shaft_{\psi} \psi_{\text{cur}} \text{cur}_{\text{tip}} \quad (3.6)$$
Where bold capital letters are used to represent $4 \times 4$ transformation matrices and each kinematic variable is limited as follows $\psi \in [-\pi, \pi], p \in [0, l_{\text{ArcLength}} + l_{\text{TipLength}}]$. The frames declared are defined as follows: Shaft is the end of the outer tube, $\psi$ is used to represent the orientation of the inner tube according to the rotation of the actuator, cur represents the end of the curved section of the inner tube, and Tip is the end of the inner tube.

$$
\begin{align*}
RCM_T\theta &= \begin{bmatrix}
\cos(\theta) & -\sin(\theta) & 0 & 0 \\
\sin(\theta) & \cos(\theta) & 0 & 0 \\
0 & 0 & 1 & 0 \\
0 & 0 & 0 & 1
\end{bmatrix} \\
\theta T\phi &= \begin{bmatrix}
\cos(\phi) & 0 & \sin(\phi) & 0 \\
0 & 1 & 0 & 0 \\
-\sin(\phi) & 0 & \cos(\phi) & 0 \\
0 & 0 & 0 & 1
\end{bmatrix} \\
\phi T_{Shaft} &= \begin{bmatrix}
1 & 0 & 0 & 0 \\
0 & 1 & 0 & 0 \\
0 & 0 & 1 & r \\
0 & 0 & 0 & 1
\end{bmatrix}
\end{align*}
$$

With the total transform from the RCM to the tip being:

$$
RCM_T_{Shaft} = RCM_T\theta \theta T\phi \phi T_{Shaft} Shaft_T\psi \psi T_{cur} cur T_{Tip}
$$

Where each kinematic variable is limited as: $\theta \in [-\pi, \pi], \phi \in [0, \pi]$ and $r \in [0, l_{\text{OuterTube}}]$ (with $l_{\text{OuterTube}}$ being the length of the outer tube). As the instrument shaft (Outer Tube) is rigidly fixed to the kuka, the transform between $T_{Flange}$, given by the KUKA application programming interface (API), and $T_{Shaft}$ is found by moving the proximal manipulator about a series of poses keeping the tip of the shaft stationary, geometrically fitting the flange position to a sphere, where the the tip is the origin. While the orientation can be found by aligning the shaft with the world
3.3. System Control

axis. $T_{RCM}$ is then set with respect to the shaft coordinate system at the start of the procedure.

3.3.3 Control Implementation

The control of the instrument is separated to each individual mechanism, where the cartesian position of the instrument shaft relative to the RCM is set explicitly in the path planning by the surgeon, while the orientation of the tube is set to maintain a constant orientation. The position $(x, y, z)$ of the tip of the outer tube can be set explicitly with respect to the RCM through:

\[ r = \sqrt{x^2 + y^2 + z^2} \]  \hspace{1cm} (3.11)
\[ \theta = \arctan \left( \frac{y}{x} \right) \]  \hspace{1cm} (3.12)
\[ \phi = \arccos \left( \frac{z}{r} \right) \]  \hspace{1cm} (3.13)

In controlling the instrument shaft, the position of the tip can be constrained to remain parallel to the Z axis of the RCM. This constraint may be resolved by calculating the orientation of the tube, $\phi$, that aligns the bending plane of the tube perpendicular to the desired imaging plane; and secondly, length of the tube extended, $l$, must be found to achieve the perpendicular constraint. The orientation of the tube can be found by finding the difference between $T_{RCM}$ and $T_{Shaft}$ around the Z axis. The length can be found by finding the angle between the RCM, the Z axis and the instrument shaft axis, $\theta$.

The control system is implemented to run on the Sunrise Cabinet (KUKA) running Windows embedded Operating System (OS) (Microsoft Corp., US) and VxWorks (Wind River Systems Inc., US) providing a soft real-time system. The KUKA is controlled through the DirectServo API, a Java library running on the Windows operating system communicating with the VxWorks OS. The DirectServo API allows the robot’s flange to be controlled in Cartesian space. The entire kinematic chain is implemented in Java, while joint control for the concentric tube mechanism is implemented using the Dynamixel SDK in C++ and integrated with the controller through a Java wrapper.
3.4 Experiments and Results

Due to constraints concerning the availability of materials and equipment for tube forming, the tube design used in the experiments is slightly different to those presented in §3.2.1. The inner tube was set to a bending radius of 30 mm, and arc angle of 80° (arc length, 41.89 mm), with a straight section, 7 mm long at the tip to join the instruments, and a transmission length of 170 mm.

3.4.1 Distal Mechanism Kinematic Assessment

To validate the precision and accuracy of the mechanism, an EM sensor was attached to the tip of the inner tube. The EM tracker was polled for the sensor pose from a ROS node at 10Hz. A controller was implemented in ROS that generated a joint trajectory, sent the joint position, and then logged the commanded and current joint positions, and the EM tracking data once the position had been reached. Two datasets were collected with each measurement repeated 10 times; the first along the entire linear workspace, from 0.0 mm to 45 mm in steps of 0.25 mm; the second was rotation only around 360° in 5° steps with the tube extended 45 mm. The data was then imported to and processed in Matlab (Mathworks, US) using the geom3d toolbox (David Legland, INRA). The repeatability of the mechanism was assessed by calculating the standard deviation of the tracker position for each commanded position. As discussed in §3.2.1, while the tube has been set to an arc with 30 mm bending radius and 80° arc angle. The actual parameters are likely to differ slightly from that, due to effects in the manufacturing process (such as springback during the shape setting). Therefore to improve the control of the system, tube parameters need to be determined using an external measurement system, in this instance the EM tracking system. These parameters can be found using just linear motion from the tube. Firstly, as the curvature of the tube is planar, the recorded positions of the tip in the tracker coordinate system is fitted and then projected to a plane. A circle is then fitted to the projected points using least square regression. The matrix equation for circular regression is shown in equation 3.14. Where \( x_i \) and \( y_i \) are the position of a projected point on the x and y axis respectively, \( n \) is the number of data points and \( A, B, \) and \( C \) are coefficients in the linearised equation of a circle 3.15. Allowing the
3.4. Experiments and Results

centre point \((x_c, y_c)\) and radius \(r\) to be found using equation 3.16.

\[
\begin{bmatrix}
\sum x_i^2 & \sum x_i y_i & \sum x_i \\
\sum x_i y_i & \sum y_i^2 & \sum y_i \\
\sum x_i & \sum y_i & n
\end{bmatrix}
\begin{bmatrix}
A \\
B \\
C
\end{bmatrix}
=
\begin{bmatrix}
\sum x_i(x_i^2 + y_i^2) \\
\sum y_i(x_i^2 + y_i^2) \\
\sum (x_i^2 + y_i^2)
\end{bmatrix}
\tag{3.14}
\]

\[x^2 + y^2 + Ax + By + C = 0 \tag{3.15}\]

\[x_c = -\frac{A}{2} \tag{3.16}\]

\[y_c = -\frac{B}{2} \tag{3.17}\]

\[r = \frac{\sqrt{4C + A^2 + B^2}}{2} \tag{3.18}\]

The arc angle is calculated using the arc cosine of the dot product of vectors from the arc centre point to the average position at the minimum and maximum lengths of the tube.

In the repeatability measurements, for the linear dataset, consisting of 181 commanded positions each repeated 10 times, the mean, maximum and minimum standard deviations were 0.05 mm, 0.2 mm and 0.02 mm respectively. For the orientation dataset, consisting of 72 commanded positions each repeated 10 times, the mean, maximum and minimum standard deviations were 0.10 mm, 0.28 mm and 0.02 mm respectively. Figure 3.4 shows the standard deviation against the commanded joint for each dataset. The arc parameters were fitted using the linear dataset and the fitted parameters were 30.16 mm bending radius and 85.15°. The tracking data and fitted circle is shown in Figure 3.5.

3.4.2 Coupled Instrument Trajectory Assessment

To validate the motion capabilities of the coupled mechanism, an EM tracker was passed along the concentric tube; it was used to compare the motion of the forward kinematics to the EM measurements. The RMS accuracy of the tracker is specified as 0.8 mm and 0.7°. The experiment was performed on an optical table (Nexus, Thorlabs Inc., US) with the proximal manipulator, EM tabletop field generator, and
Figure 3.4: Top - The standard deviation of the position of the tool against commanded tube length. Where the mean, maximum and minimum values for the norm standard deviation were 0.05 mm, 0.2 mm and 0.02 mm. Bottom - The standard deviation of the position of the tool against commanded tube orientation with the tube extended to 45 mm. Where the mean, maximum and minimum values for the standard deviation of the tube position was 0.10 mm, 0.28 mm and 0.02 mm.
3.4. Experiments and Results

Figure 3.5: EM tracking data fitted then projected to a plane (green) and and the fitted arc overlayed (red).

EM control unit as shown in Fig 3.6a. The tabletop field generator was used as there is an EM shield at the base which should minimise any potential distortions from the optical table [105]. The transformation between EM tracker coordinates and the RCM frame was found by using an adaptation of the hand eye calibration method described in [106] with 30 synchronised poses.

The trajectory of the tip was assessed using the EM tracker and commanding the instrument to follow a series of open-loop scanning trajectories while the tip was constrained to remain perpendicular to the XY plane. Two types of scanning trajectories were followed: spiral scans and raster scans. The spiral trajectory to be applied to the shaft on a XY plane was defined by: number of complete revolutions, \( c \) and final radius, \( R \), while each position is generated from iterating along \( i \) within its limits and the density of the path is determined by the number of \( i \) values used.
Figure 3.6: (a) Image of the assembly instrument with EM sensor integrated at the tip; Scanning trajectories from the kinematic data and EM tracker projected to the XY plane as the z position of the shaft remains constant throughout the scan; (b) Raster scan with an area $50 \times 50 \text{mm}^2$ with 1.5$\text{mm}$ steps along the y axis; (c) Spiral scan with a maximum radius of 50$\text{mm}$ and 5 revolutions

<table>
<thead>
<tr>
<th></th>
<th>Translation Error (mm)</th>
<th>Angular Error (deg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Spiral Scan</td>
<td>3.10 ± 11.91</td>
<td>22.31 ± 5.83</td>
</tr>
<tr>
<td>Raster Scan</td>
<td>0.77 ± 0.25</td>
<td>10.17 ± 0.59</td>
</tr>
</tbody>
</table>

Table 3.1: Metrics on trajectories, showing the mean and standard deviation

\[ a = i^2 R \]  
\[ x = a \cos(\sqrt{i(2\pi c)^2}), \ y = a \sin(\sqrt{i(2\pi c)^2}) \]  
where \( i \in [0,1] \)
The raster trajectory was defined by an area to scan and the step along each axis, the scan area was centred to the initial position. The instrument trajectories from the kinematics data and EM tracking data were aligned and due to the difference between the sampling frequencies of each device each pose from the EM tracker was matched to a kinematic pose through a distance criterion. Once matched the distance between the points was calculated and the orientation difference was determined by finding the euler angles in ZYZ convention then calculating the difference of the rotation along the Y axis. The kinematic and tracked paths from a raster and spiral scan are shown overlayed in Fig. 3.6(b) and (c) respectively. For each experiment the instrument was inserted 100 mm past the RCM. The raster scan (Fig. 3.6 b) had an area of \((50 \times 50)\text{mm}^2\) with 1.5 mm steps along the y axis. The average error between the kinematic position and tracked position was 0.77 mm while the average orientation error was 10.17°. In comparison, the spiral scan had a maximum radius of 50 mm and 5 revolutions, the average translation error was 3.10 mm and the average angular difference was 22.31° as shown in Table 3.1. The repeatability of the scans was assessed though tracking 5 identical trajectories, matching the poses along each trajectory to a reference then finding the average distance to the reference trajectory. The repeatability of the path was 0.77 mm while the combined average translation error was 3.50 mm.

3.4.3 Placental Scanning

To experiment under more realistic conditions, and aim towards demonstrating the potential clinical use of the reported instrument, a full-term (approximately 40 weeks gestation age) human placenta was collected following a caesarean section delivery and after obtaining a written informed consent from the mother at University College London Hospital (UCLH). The Joint UCL/UCLH Committees on the Ethics of Human Research approved the study (ethical approval UCL 08/H0714/87). The placenta was transported to the laboratory and immersed in water within a container. The instrument was immersed in the water and introduced above the placenta as shown in Fig 3.7a. Additionally a stereo laparoscope was positioned alongside the instrument to illuminate the workspace as the instrument
3.4. Experiments and Results

Figure 3.7: (a) Image of the placenta imaging setup, showing the position of the probe and laparoscope relative to the placenta; (b) Image of the instrument partially submerged in water with the tip extended to view the placenta along the normal to the placenta surface; (c) *ex vivo* imaging of human term placenta

The immersed placenta was imaged a number of times using both raster scans and spiral motion patterns acquiring stereoscopic data. (The position of the instrument relative to the placenta can be seen in Fig 3.7b.) The camera acquires synchronised images at 27 Frames Per Second (FPS) with a resolution of 250 × 250 in each camera. (An image from one of the cameras can be seen in Fig 3.7c.) To demonstrate the potential information that can be derived from the scans of the placenta, stereoscopic reconstruction of the placenta was undertaken using the method described in [107] and shown in Fig. 3.8a and b. In addition, the camera trajectory was reconstructed using ORB-SLAM [108] and aligned with the instrument kinematics. However due to the fast "wind back" motions during the spiral scan, the video and kinematics were subsampled to 60 keyframes with the path shown in Fig. 3.8c and d. A mosaic using the method described in [109] was constructed of three scanned areas: a raster scan; a spiral scan of a real placenta; a spiral scan of a model placenta (see Fig. 3.9a, b and c
3.5 Discussion

FLP is challenging due to the small Field of View (FoV), which inhibits localisation and makes it difficult to be certain that all vessels are ablated. Intra-operative motion of the fetuses and physiological signals from the mother make it difficult to maintain a measured energy delivery and ensure consistent therapeutic delivery.
3.5. Discussion

The distal instrument developed improves the dexterity of the fetoscope by providing additional DoF at the tip allowing the orientation of the scope to be changed. The prototype was fabricated using either 3D printed components or commercial off-the-shelf components. The current design, as described above, allows the tubes to be easily swapped and the joint workspace to be modified according to the clinical requirements of the instrument. The interface with the actuators provides a separation between the mechanism and actuator. Additionally, the layout of the mechanism and the interface with the actuators, within the coupled design groups the non-sterilisable objects (i.e., the actuators and proximal manipulator) together. This approach could potentially allow the system to be sterilisable through (a) constructing the mechanism from sterilisable materials and (b) the use of a single sterile drape with two coupling adaptors to transmit the motion from the actuators through the sterile barrier. The end-effectors have been attached to the tip of the instrument directly, which while providing a rigid attachment between sensor and instrument, prevents the continuous rotation of the tube due to the sensor cabling. This is currently compensated for by preventing the joint position from wrapping and enforcing the joint limits between $-180^\circ$ and $180^\circ$.

The end-effectors integrated were either a stereo camera or an EM tracking sensor. For clinical use the instrument would require an internal light source (rather than the external laparoscope used in this chapter’s experiments) and a working channel for an interventional laser. With the diameter of fetoscopic instruments averaging $3\ mm$ and generally not exceeding $5\ mm$ [40]. This is likely possible as the current instrument diameter is $2.45\ mm$, with the stereo camera. The interventional lasers used generally consist of a fibre with a core diameter between $400 – 600\ \mu m$ which with a plastic sheath for protection is $< 1\ mm$. While light can be provided through fibres approximately $1\ mm$ in diameter, however multiple smaller diameter fibre can be used to utilise the space available after placing the camera and working channel.

The distal mechanism described here is assessed through the repeatability of the commanded positions with an average standard deviation of the tube extension...
and orientation of 0.05 mm and 0.10 mm respectively. The linear dataset show the tube extension to be highly repeatable with the standard deviation increasing with the tube extension with a range of 0.20 mm and 0.02 mm. This indicates good coupling between the lead screw and actuator and low backlash between the leadscrew and carriage. The orientation dataset showed the tube orientation was comparable to the tube extension but slightly less repeatable, with the standard deviation ranging between 0.28 mm and 0.02 mm. The reduction in repeatability is likely due to backlash in the gear train going from the interface between the square shaft and the drive gear, followed by the tube and drive gears. However, Computer Numerical Control (CNC) machining of the gears and square shaft interface could allow for smaller tolerances between the parts reducing backlash, and the material used could be optimised for wear and surface friction.

The tube parameters were determined using a least squares regression method. An arc was fitted from 1810 synchronised positions using the linear dataset. The arc parameters determined were 30.16 mm and 85.15°. These parameters set the arc length as 44.82 mm, while the maximum tube extension is 45 mm. This method, it should be noted, fits the arc to the position of the EM sensor embedded at the tip of the tube rather than the curved section of the tube itself.

The current accuracy of the combined system is quite low considering the scale of fetal surgery and is reflected in the trajectory comparisons, shown in Fig 3.6. The pose accuracy of the instrument was shown to be 3.10 ± 11.91 mm and 22.31 ± 5.93° for the spiral scans and 0.77 ± 0.25 mm and 10.17 ± 0.59° for the raster scans. This disparity in accuracy information is likely due to the comparison of unsynchronised trajectories, and the individual poses being incorrectly matched. Synchronisation of the instrument controller and tracker could allow poses to be directly measured against one another. However, while the accuracy measurements are relatively high, the repeatability of the trajectories was shown to be 0.77 mm. This comparison between the repeatability and the accuracy measurements indicates the system is repeatable but the kinematic models and parameters used were not sufficient to determine the pose of the instrument accurately. Potentially learning approaches
could be applied to parameterize the control of the concentric tube while maintaining computational performance. Alternatively, determining the tube parameters using the endoscope image information could allow for more accurate control of the system.

The experiments show the combination of the proximal motion controlled by the manipulator constrained to a RCM and the distal motion from the concentric tube. We performed two scanning motions that have regular structure for analysis - raster and spiral. Spiral motions are often applied in visual scanning mechanisms [110, 111], they require the imaging component to be capable of continuous rotation or to be "wound back" to an initial position. The system as developed and described here resolves this by limiting the rotation to a single revolution. While this prevents the sensor cables from winding up, it also affects the trajectory of the instrument during scans especially spirals as the rotation has to reset during the scan interrupting the commanded trajectory. This can be seen in Fig. 3.6c along the left side of each revolution. In comparison, the raster scan has an offset at the peripheral; however, it does not have the sudden "wind back" motions. Stereoscopic reconstruction has the potential to provide additional information during the procedure, for example the distance from the camera to the vessels and vessel size. The mosaics presented show the potential to increase the field of view intra operatively. The mosaicing technique did not use the kinematic data in the reconstruction process which, as noted in relation to other tracking systems (i.e., EM or optical tracking) would allow larger and more accurate mosaics [112]. However, as the technique relies solely on image data, they can be robust to motion from either the instrument or the imaging target. Additionally mosaicing techniques have made use of deep learning methods to improve the robustness whilst still relying solely on image data [113].

3.6 Conclusions

This chapter presented research and design concerning the coupling of a redundant proximal manipulator with a low diameter concentric tube manipulator holding a
3.6. Conclusions

stereo camera at the tip. The concentric tube mechanism has been designed for fetoscopic procedures as part of which the instrument design requires a small diameter and additional DoF at the tip. The control system implemented was developed to improve fetoscopic interventions by providing a stable view of the placenta at a constant orientation: this capability will facilitate surgical assist in the delivery of laser photocoagulation to placental vasculature for the treatment of TTTS. Chapter 4, will focus on improving the proximal side of the system, allowing for closer integration with the sensors on the endoscope.
Chapter 4

Proximal Manipulation of Fetoscopes

4.1 Introduction

This chapter focuses on the improvement of the proximal DoF presented in §3, these DoF are manipulated by a serial robot manipulator proximal to the surgeon. The disadvantages of proximal manipulation system (presented in the previous chapter) can be summarised as:

- The control function is separated from the imaging sensor: the control of the endoscope was situated directly on the Kuka Sunrise controller, while the sensors were controlled on a separate computer.

- The inverse kinematics of the iiwa robot are provided as a closed-form solution; where the solution requires a pose and the seventh DoF is expressed as the third joint angle. Overall, this arrangement limits the workspace when constrained to a RCM.

- Trajectories provided to the controller are not capable of checking for collisions within the known scene.

The proximal control was therefore redesigned within ROS, which allows for: the simpler transfer of data between the robotic control and sensors; the integration of existing robotics work (such as inverse kinematics solvers and path planners); allows the presented work to be robot agnostic.
Due to their small form factor, it is noted that fetoscopes often suffer from a limited field of view, which complicates the manipulation of these types of endoscope [40]. While fetoscopy is traditionally performed manually, precise control of the instrumentation through the incorporation of robotics has the potential to improve the stability of imaging devices [101, 114]. In addition, the ability to execute pre-computed trajectories and utilise feedback from other sensors (such as strain gauges for force control) can improve image acquisition strategies and enable high-quality multi-modal imaging [110, 115].

Recent research has focused on improving intrauterine visualisation: both by improving established modalities (for example white light imaging with mosaicing [112, 116]) and introducing new imaging modalities (such as Optical Ultrasound [42] and photoacoustic imaging [117]). These additional modalities can provide an increased field of view, yield information below the tissue surface and enable visualisation in situations where the amniotic fluid is not optically clear. Optical Ultrasound (OpUS), where ultrasound is both generated and received using light [42], is emerging as a particularly versatile modality and one that is readily miniaturised (and therefore well-suited to interventional applications). OpUS can provide high resolution imaging from small form factor devices and has recently been demonstrated for real-time in vivo guidance from a surgical needle [118]. In combination with robotics, the application shows significant potential for improving control during surgical procedures [114].

The content in this chapter is an extension of work presented in [119] and presents the design and operation of a rigid, robot-mounted endoscope; one that for the first-time integrates a stereo camera with an OpUS sensor (detailed in §4.3.2). Through a ROS interface, imaging data obtained from both modalities were acquired simultaneously and in real-time during robotic manipulation. After performing a calibration between the sensors and the robot, a tissue mimicking phantom (with anatomically realistic vascular structures) was imaged using various scan apertures; the resulting 3D data were collected, processed and displayed in real-time.
4.2 Proximal Instrument Control

4.2.1 KUKA iiwa ROS Interface

The KUKA iiwa was controlled through the Fast Robot Interface (FRI), a proprietary API supplied by KUKA. The API is a real-time interface to the robot controller from an external computer and utilises universal datagram packet communication. The FRI provides the ability to monitor the current joint positions and torques, and to set: joint position, joint torques and Cartesian wrench at the flange of the robot at 500 Hz control loop. This is implemented as a ROS control hardware interface for controlling the joint positions (joint position interface and joint state interface) [120]; it allows joint position, velocity and effort limits to be set and enforced before being sent to the controller. The Cartesian control and path planning of the manipulator flange is then managed using the Moveit motion planning framework [121]. This utilises the TRAC-IK plugin [122] and the open motion planning library [123] as the inverse kinematics solver and path planner, respectively.

4.2.2 Remote Centre of Motion Constraint

When used in minimally invasive procedures, endoscopes are inserted into the body through the point of incision. In order to minimise stresses on the incision, the endoscope tip position $\text{RCM}^\text{T}_\text{Tip}$ is constrained to a RCM, which is represented as four virtual joints (4 DoF) at the incision point; three revolute joints allowing rotation around each of the three axes, and a prismatic joint allowing translation along the long axis of the endoscope (referred to as the Z axis), as shown in Fig. 4.1. The endoscope tip position can be described as

$$\text{RCM}^\text{T}_\text{Tip} = \text{Rot}_X(\psi)\text{Rot}_Y(\phi)\text{Rot}_Z(\gamma)\text{Trans}_Z(r),$$  \hspace{1cm} (4.1)

where $\text{Rot}_X(\psi)$ denotes a transform representing a rotation over angle $\psi$ around the X axis, and $\text{Trans}_Z(r)$ denotes the transform representing a translation along the Z axis over a distance $r$. Rotations around the Y and Z axes are denoted likewise. Eq. 4.1 was then analytically inverted to obtain the RCM joint co-ordinates for the four DoFs for a given position $\text{RCM}^\text{T}_\text{Tip}$. This inversion is implemented as a ROS
4.2. Proximal Instrument Control

Figure 4.1: Illustration of the RCM constraint imposed on the robot manipulator and the available DoF

The ROS node then: publishes the current joint position, Cartesian position and pose of the RCM; subscribes to the desired joint position and Cartesian position; advertises a service to set the RCM and to send trajectories in joint or position space. When receiving new commands, either as a path or point, the desired position of the tip in joint space is interpolated from the current position in 3.0° and 5.0 mm increments. The desired pose is then transformed to the end-effector of the iiwa and the interpolated path is passed to the Cartesian path planner in Moveit, which attempts to plan the trajectory in 1.0 mm steps. This node is executed alongside the
4.3 Instrument Design

4.3.1 Mechanical Design

A rigid endoscope with a diameter of 5.00 mm and a length of 300.00 mm was designed and developed. At its proximal end, a custom 3D-printed housing was used to mount the instrument to a robotic manipulator. The endoscope comprised of a stainless steel tube (5.00 mm outer diameter, 4.35 mm inner diameter) and a custom sintered stainless steel endoscope tip. The endoscope tip matched the profile of the instrument shaft and housed a Naneye Stereo camera (ams AG, Austria), two fibre optic lighting channels and an 1.10 mm working channel. The endoscope tip was printed in two sections to facilitate assembly around the stereo camera. The entire assembled endoscope and endoscope tip is shown at Fig. 4.2.

4.2.3 Tool Calibration

The fetoscope is controlled relative to the tip of the instrument. Therefore, to control the robot, the transformation between the robot flange and the tip is needed. The translation of the tip is calibrated to the robot flange through a pivot calibration which involves pivoting the instrument about the endoscope tip. The calibration is performed by moving the manipulator across a number of joint configurations where the position of the endoscope remains constant but the orientation changes. At each configuration, the pose of the robot flange is recorded. To locate the position of the scope tip, a sphere is then fitted to the recorded flange positions. The cameras and OpUS are calibrated geometrically from the CAD model of the 3D printed endoscope tip and by physical measurement.

4.3 Instrument Design

4.3.1 Mechanical Design

A rigid endoscope with a diameter of 5.00 \text{ mm} and a length of 300.00 \text{ mm} was designed and developed. At its proximal end, a custom 3D-printed housing was used to mount the instrument to a robotic manipulator. The endoscope comprised of a stainless steel tube (5.00 \text{ mm} outer diameter, 4.35 \text{ mm} inner diameter) and a custom sintered stainless steel endoscope tip. The endoscope tip matched the profile of the instrument shaft and housed a Naneye Stereo camera (ams AG, Austria), two fibre optic lighting channels and an 1.10 \text{ mm} working channel. The endoscope tip was printed in two sections to facilitate assembly around the stereo camera. The entire assembled endoscope and endoscope tip is shown at Fig. 4.2.
4.3.2 Optical Ultrasound

An OpUS probe comprising a fibre optic ultrasound transmitter and a fibre optic ultrasound receiver was fabricated. The transmitter comprised a multi-walled carbon nanotube (MWCNT) with a polydimethylsiloxane (PDMS) composite [42, 125] coated onto the distal end surface of a 400 $\mu m$ core, multimode, optical fibre. The receiver comprised a plano-concave Fabry-Pérot cavity coated on the distal end surface of a single mode optical fibre (SMF-28) [126]. The two optical fibres were held adjacent and heat shrink tubing was used to align their distal end surfaces. For robustness, this pair of optical fibres was housed within an acoustically transparent polymer tube, with an outer diameter of 1.2 $mm$ (TPX, Mitsui Chemicals Inc., JP). The ultrasound probe generated pressures in excess of 2 $MPa$ at a distance of 1.5 $mm$, with a corresponding $-6$ dB ultrasound bandwidth of ca. 30 $MHz$. As described here, the OpUS imaging probe achieved an axial imaging resolution of ca. 60 $\mu m$ [118].

4.3.3 White Light Imaging

A Naneye stereo camera was selected because of its a small footprint (measuring 2 $\times$ 1 $mm$), and the camera is constrained to the printed endoscope tip through a friction fit between the two halves of the tip. The fibre optic lighting channels measured 1.1 $mm$ in diameter and each channel contained 9 fibres (200 $\mu m$ core, 500 $\mu m$ coating, 0.5NA) (FP200ERT, Thorlabs Inc., US). Each fibre was stripped to the cladding with a length of approximately 10 $mm$ to span the length of the lighting channel (of approximately 7 $mm$); the fibres being inserted through the channel and fixed in place with optically clear epoxy (F120, Thorlabs Inc., US). Once the epoxy set, the protruding fibres were cleaved off and polished to the face of the endoscope tip down to 1 $\mu m$ lapping sheets (LF1P, Thorlabs Inc., US). At the proximal end, the 18 optical fibres were fixed within a single Sub Multi Assembly (SMA) connector, again using epoxy and similarly polished moving down to 1 $\mu m$ lapping sheets. The resulting bi-furcated fibre bundle was coupled to a standard surgical light source (Cold Light Fountain Power LED 175 SCB, Karl Storz, GmbH, Germany).
4.3. Instrument Design

Due to the protective TPX sheath around the sensor, the OpUS sensor could not be placed within the working channel of the endoscope. It was therefore mounted to the side of the instrument shaft, with a consequential increase in the overall diameter of the endoscope to 6.20 mm. The assembled endoscope, shown in Fig. 4.2, was mounted to a KUKA LBR iiwa 14 R820 robotic manipulator (KUKA AG, Germany): this is a 7 DoF manipulator with a payload of 14 kg, a repeatability of ±0.15 mm, and a reach of 820 mm to enable robotic manipulation.
4.4 Sensor Interfaces

4.4.1 Camera Interface

The Naneye cameras were driven through an FPGA controller configured to poll the two cameras simultaneously to provide synchronised frames. The API for the camera controller is only available for Windows, hence the ROS.Net library was used to interface with the controller using a Windows computer [127]. The resulting images were then published into ROS with the respective camera information at a frame rate of 15 Hz. One example of a pair of synchronised images (corresponding to the left and right camera) is shown at the top of Fig. 4.3.

4.4.2 US Interface

The OpUS imaging probe was coupled to a console similar to that previously described [42, 128]. Ultrasound was generated using a Q-switched pulsed laser (DSS-1064 Q, Crylas, Germany) with a wavelength of 1064 nm, a pulse width of 1.5 ns, and a repetition rate of 100 Hz. The pulse energy delivered to the absorbing coating on the ultrasound probe was 20 µJ. The Fabry-Pérot ultrasound receiver was interrogated using a continuous wave tuneable laser (TSL-550, Santec, UK), with a wavelength range 1500 – 1600 nm, via a circulator. The reflected optical signal from the Fabry-Pérot cavity was measured using a photodetector and digitised at 62.5 MS/s (M4i.4420-x8, Spectrum, Germany), and the low frequency optical signal was used to track the optimum bias point for the sensor. The received ultrasound data was frequency filtered (Butterworth, 4th Order, Bandpass, 3 – 25 MHz), had the cross-talk removed [42], and underwent Hilbert and log transforms to display the A-scans.

The OpUS acquisition was performed using a custom LabVIEW script that streamed both raw and signal-processed A-scans over ROS (an example of a raw A-scan is shown in the bottom of Fig. 4.3). Using the position and orientation of the endoscope tip, obtained from the forward kinematics of the manipulator, the A-scans were transformed and rendered in the robot base frame in real-time. Thus a 3D volume scan could be acquired and displayed in real-time by traversing the
endoscope tip across an arbitrary 3D scan pattern. The surface information of the scan was also constructed as a point cloud by finding the maximum intensity point along each A-scan. If this maximum intensity exceeded a certain threshold, it was transformed to the world frame and the colour was mapped to the distance of the point from the endoscope tip. Any outliers of the point cloud were removed using the pcdenoise function in MATLAB (The MathWorks Inc, USA), where outliers are detected by having less than 4 neighbours closer than the average distance to the neighbours of all points.

4.5 Experiments and Results

4.5.1 Experimental Setup

The Kuka LBR iiwa 14 R820 is mounted rigidly to an optical table (Thorlabs) through an adapter plate that provides M6 holes for the optical table and M8 for the manipulator. The manipulator is configured in Sunrise workbench (Kuka GmbH, DE) for use with the fast robot interface and a tool, with dynamic properties but no geometry (i.e., tool tip calibration information), attached. Within ROS the manipulator is described with the optical table and endoscope housing attached.

To provide external tracking of the manipulator and instrument, an Optotrack Certus optical tracker (Northern Digital Inc., CAN) is used. The tracker uses three converging cameras and active wired infrared markers; these are used to provide 3D positions of each marker, and poses assembled from rigid bodies made of multiple markers (a minimum of 3 markers). The accuracy of the 3D positions is rated to 100 $\mu m$ and the measurements are provided to a resolution of 10 $\mu m$. The tracker communication is through Ethernet and a native linux API is provided. A custom application was written to poll the tracker periodically at 167 $Hz$ for the current position of each marker and any defined rigid bodies, and then publish each to ROS, using the transform library [129]. The optical tracker is mounted vertically to a four wheeled cart, via an anti-vibration mat, and positioned approximately 2 $m$ away from the base of the manipulator. Four markers are rigidly attached using an adhesive to the base of the manipulator, while another six markers are attached to
4.5. Experiments and Results

Figure 4.3: Top - Example of unprocessed white light images underwater of a placenta phantom obtained from the Naneye Stereo camera using only the integrated lighting channels. Red lines indicate the edges of the placental vessels. Bottom - a raw OpUS A-scan displayed over the full imaging range obtained from a tissue-mimicking acoustic placenta phantom. The pulse-echo signals from the phantom can be observed at a depth of approximately 12 mm. The signal observed for depths < 5 mm is due to direct cross-talk between the fibre-optic ultrasound source and receiver, and is suppressed in subsequent signal processing.

an acrylic sheet which is then mounted to the endoscope housing. Fig. 4.4 shows the complete tracking setup.

To obtain the position of the endoscope tip, markers are attached to the endoscope body and used to perform a pivot calibration. The pivot calibration is done using NDI 6D Architect (Northern Digital Inc., CA) and output as a rigid body. This is then used to perform a hand-eye calibration between the robot base and the
4.5. Experiments and Results

4.5.1 RCM Kinematic Chain

Using the optical tracker aligned to the robot coordinate system, the tip positioning accuracy and repeatability is assessed. 100 positions in the RCM joint space were generated across a uniform distribution bound by the joint limits; these were set to $\pm 20^\circ$, $\pm 20^\circ$ and 100 mm for the rotation about X, rotation about Y and translation along Z. The controller is then commanded to move to each position and, when the list is completed, it is randomly shuffled and the controller is commanded to return to each position: this procedure is repeated 20 times. Upon completing the trajectory to each position, the tracker is triggered and the tracked rigid bodies, the

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**Figure 4.4:** Experimental setup for the assessment of the endoscope positioning accuracy and repeatability.
4.5. Experiments and Results

The kinematic pose of each robot link and the desired pose is captured.

After collection, each set of poses with the same desired position is found. The repeatability of the endoscope is assessed through the standard deviation between the measured tracker positions. The controller accuracy of the endoscope is determined by examining the difference between the desired position and the measured tracker position; while the kinematic accuracy is the difference between the achieved kinematic position and measured tracker position. Lastly, the controller resolution is calculated by establishing the difference between the desired position and the achieved kinematic position.

Results from the assessment of data were: 0.82 ± 0.80 mm for the repeatability, 2.13 ± 1.20 mm for the controller accuracy, 2.17 ± 1.12 mm for the kinematic accuracy and 0.33 ± 0.16 mm for the controller resolution. The increase in error at higher angles presents the possibility of the tracker error contributing to the endoscope error. Therefore, to determine the contribution of high viewing angles, the experiment was repeated but the rotation about the Y axis was reduced to ±10°; the purpose of this was to reduce the viewing angle between the optical tracker and the markers attached to the endoscope. The results from the assessment were 0.54 ± 0.27 mm for the repeatability, 1.35 ± 0.62 mm for the controller accuracy, 1.21 ± 0.59 mm for the kinematic accuracy and 0.30 ± 0.07 mm for the controller resolution. A comparison of the average error of the controller accuracy, for each repeated position, was plotted against rotation about X axis and Y axis (shown in Fig 4.5). For the fitted error lines on each graph, the gradient and Y axis intercept (average error at 0°) for the ±20° data set was −0.010 and 2.12 mm; and 0.06 and 2.04 mm for the X and Y axis rotations, respectively. While for the ±10° data set was −0.01 and 1.34 mm; and 0.05 and 1.34 mm for the X and Y axis rotations, respectively.

The RCM is assessed using both the pose from the kinematics and an external tracker. The tool calibration is used to determine two points on the instrument shaft, the tip and the base, transformed to the parent reference frame using the measurement system (kinematic or external tracker). The distance of the instrument to the RCM is then determined by forming a line between the two points on the
4.5. Experiments and Results

Average error over repeated positions (mm) against rotation about the X axis (red) and Y axis (blue) of the RCM. Red and Blue lines are fitted to the X and Y axis data respectively. **Top** - the rotation about the Y axis is constrained between ±20°. **Bottom** - the rotation about the Y axis is constrained between ±10°.

**Figure 4.5:**

For repeated positions against rotation about the X axis (red) and Y axis (blue) of the RCM. Red and Blue lines are fitted to the X and Y axis data respectively. **Top** - the rotation about the Y axis is constrained between ±20°. **Bottom** - the rotation about the Y axis is constrained between ±10°.

Instrument shaft, and finding the distance between the line and a point on the RCM using Eq. 4.2 [131], where $P_{\text{tip}}$ and $P_{\text{base}}$ is the position of the tip and the base of the endoscope respectively and $P_{\text{RCM}}$ is the position of the RCM.

$$d = \frac{|(P_{\text{RCM}} - P_{\text{tip}}) \times (P_{\text{RCM}} - P_{\text{base}})|}{|P_{\text{base}} - P_{\text{tip}}|} \tag{4.2}$$

This is implemented as a ROS node, where the transforms of each frame are polled using the ROS package tf2 [129] in order to obtain synchronised data; the distance between the instrument shaft and RCM is then calculated and logged. Using the same data as described in the previous section, the distance to the RCM was assessed with both the kinematic position and tracker position. In both instances, the base point was provided from the robot kinematics and knowledge of the geometry of the endoscope. The average errors were $1.22 \pm 0.77 \, mm$ and $1.41 \pm 0.74 \, mm$ for the kinematic and tracker positions, respectively.
4.5.3 Optical Ultrasound Imaging

The placenta phantom (i.e., the imaging target) was fabricated using gel wax as a tissue-mimicking material. Vascular structures were replicated from a human placenta with 3D drawing software and then converted to a 3D printed mould. Solid glass spheres were added to melted gel wax to provide acoustic backscattering. The phantom featured clinically-realistic acoustic properties, such as propagation speed of sound, echogenicity and speckle pattern [132]. It was placed into a water bath onto an acrylic sheet and clamped down with metal posts along the side of the phantom (shown in Fig. 4.6).

The endoscope was scanned across the phantom with a linear trajectory planar to the surface. The trajectory was 40 mm long and A-scans are collected in 1 mm increments. The A-scans sample points were then transformed into world coordinates, and the intensity of points were bound between $-50$ and $-10$ dB, which was then color mapped. The resulting B-scan took 16 seconds to acquire and is shown at Fig. 4.7.

Using the endoscope (as described above), the phantom was scanned with two independent scanning trajectories: a raster and a spiral trajectory. The RCM was placed approximately in the centre of the phantom at a height of 160 mm above the phantom. The raster scan was generated on a plane over the entire 80 $\times$ 80 mm area, in 1 mm increments, at approximately 10 mm from the phantom. The spiral scan was limited to a total diameter of 80 mm, and comprised 40 cycles around the start point (to yield a comparable scan density). Due to difficulties in synchronisation between the OpUS acquisition system, the robot controller and the ROS master; the robot motion to each position was followed by a pause of 0.1 s to ensure a stationary endoscope tip. This start-stop acquisition strategy resulted in a total scanning time of 43 minutes and 45 minutes for the raster and spiral scans, respectively.

The resulting 3D OpUS images are shown in Fig. 4.8. In these images, the surface vessels on the placenta are clearly visible across the entire imaging aperture. In addition, the supporting acrylic base can be observed at a depth of 35 mm (as can artefacts caused by reflections off the posts used to fix the placenta phantom). The
Figure 4.6: Photograph of the placenta phantom from above to show the positions of the posts to maintain the position of the phantom. White box shows area presented in Fig. 4.7.
4.5. Experiments and Results

Figure 4.7: Top - Top down view of placenta phantom with green line to indicate scan paths. Bottom - Post processed B-scan image of the placenta phantom.

The apparent absence of signal along the top edges of the visualisations is likely due to a sub-optimal choice of threshold when displaying the data. However, it was noted that the raw A-scans did, indeed, capture the placenta phantom surface.

4.5.4 Scan Repeatability and Reliability

Following the same experimental setup as §4.5.3, a $40 \times 40$ mm raster scan and $40$ mm diameter spiral scan were repeated 10 times. Point clouds of the surface were then generated for each of the scans; each subsequent scan was then registered to the
Figure 4.8: 3D visualisation of the OpUS images of the placenta phantom acquired through robotic probe manipulation. **Bottom Right** - A photograph of the gelwax placenta phantom is shown. **Top Row** - 3D OpUS images obtained with the instrument shown in angled view and **Middle Row** - shown in top-down view. The scanning trajectories used are spiral and raster scans on the left and right respectively. **Bottom Left** - XZ and YZ views of the 3D OpUS image obtained using a spiral scan pattern. In each image the distance of the points from the endoscope tip are colour-encoded using the colorbar (in mm) displayed in the bottom centre.
first scan of the same type by using the point-to-point variant of the Iterative Closest
Point (ICP) algorithm [133]. The root mean square error for each registration (this
being the euclidean distance between each point cloud for each registration) is found
and the mean and standard deviation calculated. For the raster and spiral scan, the
mean errors were $0.45 \pm 0.14 \text{ mm}$ and $0.27 \pm 0.02 \text{ mm}$, respectively. Each scan was
manually segmented to determine the number of outliers from the surface of the
placenta phantom (attributable to inaccurate data due to noise or other artefacts).
The percentage of outliers was $3.12\%$ and $2.96\%$ for the raster and spiral scans,
respectively.

4.6 Discussion

The results described above represent the first demonstration of OpUS and optical
imaging integrated into an endoscope and controlled by a robotic manipulator. The
OpUS imaging probe provides an A-scan with an imaging depth of 30.0 $\text{ mm}$ and
an axial resolution of 60 $\mu\text{m}$. However the OpUS transmitter based on a 400 $\mu\text{m}$
optical fibre, exhibited partial divergence [118]. Thus the lateral resolution of the
system was dependent on the imaging depth and decreased with depth: at its widest,
the lateral resolution was approximately $2.0 \text{ mm}$. The use of a larger diameter OpUS
transmitter, or one with a concave surface, would lead to improved directivity and
an improvement of lateral resolutions [134].

The imaging probe selected for this novel application had a protective sheath
around it with an overall diameter of 1.2 $\text{ mm}$, which precluded it from being inserted
into the working channel. Instead the sensor was fixed to the side of the endoscope,
increasing the width of the instrument to 6.2 $\text{ mm}$. With the diameter of fetoscopic in-
struments generally not exceeding 5 $\text{ mm}$, this precludes the current instrument from
being suitable for procedures. However, the width at the largest lateral extent of the
two optical fibres comprising the ultrasound probe was $< 800 \mu\text{m}$; an arrangement
which would allow them to be directly integrated into the working channel of the
endoscope. In future iterations of the endoscope design, a dedicated channel within
the endoscope tip would provide protection for the OpUS probe whilst minimizing
the total diameter of the instrument.

The placenta phantom studied in this work emulated anatomically accurate placental surface vasculature [132]. It was designed to achieve clinically relevant acoustical properties and, through the addition of purple dye, the phantom was observed to exhibit a passable visual likeliness to an actual placenta. Using this phantom, OpUS imaging was demonstrated at a penetration depth of up to 20 mm, and the vasculature was clearly visualised and accurately depth-resolved. However, the absence of colour contrast rendered the phantom sub-optimal for white light imaging; a limitation which revealed only geometry and not function. In future work, a placenta phantom will be developed that has both clinically relevant acoustical and optical properties. This solution will further increase the value of the combined white light and OpUS imaging as demonstrated by the rigid endoscope presented here.

One further advantage of the OpUS modality is the ability to integrate complementary modalities; this is achieved by using wavelength selective coatings [128] that selectively absorb or transmit light of different wavelengths. (An experiment incorporating selectively absorbing coatings and combining OpUS and photoacoustic imaging has recently been demonstrated [128].) Without compromising device size, photoacoustic imaging provided tissue specificity while OpUS yielded structural information. In addition, such selectively absorbing coatings could be used to allow for laser light delivery for ablation or photocoagulation, or to deliver illumination for stereoscopic cameras. Furthermore, introduction of the ability to provide multiple modalities through a single instrument channel will allow for further miniaturisation of endoscopes, and simultaneously provide more information to the clinician.

The point cloud generated by the OpUS trajectories covers an area of $7127 \, mm^2$ and $6474 \, mm^2$ for the raster and spiral pattern, respectively. The constructed point clouds clearly show the vasculature of the phantom; however, the quality of the scan deteriorates at increasing lateral distances from the incision point, as the angle between the surface normal and the instrument increases. This
undesirable effect is due to the signal processing performed in this work. To address this limitations in future implementations, a signal-dependent threshold could be used to improve the homogeneity of the image. In addition, it should be noted that the point clouds presented in this work merely show the voxels exhibiting maximum intensity; this results in the visualisation of only the surface. In future work, different visualisation techniques (e.g., mapping the signal intensity to the voxel alpha channel) will also be implemented to enable sub-surface visualisations. The repeatability assessment of the scans demonstrates the potential for the endoscope to be returned to previously scanned positions after the point cloud has been constructed. This could be used clinically to assist in enforcing the distance and angle constraints for laser coagulation; after which it could be used to assess the quality of the coagulation.

The current system requires a stop in the motion to allow for OpUS data to be synchronised with the step position, resulting in an acquisition rate of approximately 2.5 points per second. Although the OpUS and ROS interfaces allow for data transmission at a rate of approximately 60 A-scans per second, the acquisition could be accelerated by synchronising the acquisition PCs and timestamping the OpUS data. This change would allow for a smooth continuous motion during acquisition and enable faster acquisition speeds. Within the current robot control strategy, this would result in scans of equal area and density shown previously (Fig. 4.8) acquired in 2′35″ and 4′5″ for the raster and spiral scans, respectively. This is a notable enhancement and a significant benefit when working to precise clinical timescales.

Lastly, the scans are currently generated in reference to the robot coordinate system using the robot kinematics and tool calibration to determine the sensor position. However, motion of the imaging target is not accounted for, which would adversely influence the scan quality used in surgery due to respiratory and cardiac motion. Closer coupling between the stereo camera and the optical Ultrasound could potentially be utilised to compensate for external motion. For example, using stereo reconstruction techniques [107] to determine the position of the imaging target to the endoscope, and generate the scans in terms of the imaging target and
4.7 Conclusions

This chapter has presented the design, control and operation of a multimodal robotic endoscope with integrated white light optical and OpUS imaging. The endoscope consists of a miniature stereo camera, fibre optic light channels and an OpUS sensor within a 6.2 mm overall diameter. The instrument is manipulated by a KUKA LBR iiwa R820 arm, constrained through software to a remote centre of motion.

The performance of the endoscope is demonstrated through generating scanning paths over a gel wax placenta phantom. A-scans acquired by the OpUS sensor are processed into B-scans and 3D surfaces using the kinematics of the endoscope. The vessels can be clearly delimited from the main body of the placenta. Results from this work were published in 2019[119] and at that time, and to the author’s best knowledge, the OpUS scans presented in Fig. 4.8 were the largest published in peer-reviewed literature. The main limitation of the system was the acquisition speed of the OpUS C-scans: even with continuous motion there is a safety limitation to the speed of the instrument due to the RCM.

The following chapter will present the use of a distal manipulator, one developed specifically to perform small area, fast scans using the OpUS sensor. The aim of the new instrument is to improve the acquisition speed of scans while maintaining the requisite density.
Chapter 5

Distal Manipulation of Fetoscopic Imaging Probes

5.1 Introduction

OpUS, as an interventional imaging modaility, has a number of advantages over conventional piezo-electric based US. This is because optical fibres are:

- readily miniaturised
- EM compatible
- relatively low cost

However, one of the main challenges in the use of OpUS is that low diameter probes, of the type available currently, produce A-scans giving one-dimensional information only. One dimensional structural information is inherently difficult to interpret as clinical information (for example, when attempting to identify vessels on and below the tissue surface). However, these probes can be used to assist other modalities. One such example is the use of the A-scan to determine the distance to a surface: a measurement which can be used as feedback for a controller to maintain a distance constraint with a robotic manipulator [114, 135]. This has been demonstrated in an instrument with a camera and a therapeutic laser, where the distance constraint provides improved white light camera images and improves performance in FLP. Alternatively, the transmitter can be embedded into a needle, and used with
5.1. Introduction

a conventional ultrasound probe to assist in needle tracking [136]. In that example, only a Fabry-Pérot ultrasound receiver is integrated into the needle and is used to detect encoded transmissions from a modified transabdominal US transducer. This arrangement incorporates 2 arrays of unfocused transducers either side of the main imaging transducers; the strength of the signal detected by the embedded receiver from the arrays can be used to determine the 3D position of the needle relative to the US transducer.

However, to acquire structural information similar to conventional ultrasound, information is needed across multiple axes. One means of achieving this is the use of a galvano mirror and a cylindrical lens above the linear membrane of the ultrasound generator. The mirrors scan across the membrane generating US pulses which are detected by a single receiver; these can be processed to form a B-scan (two dimensional data, resembling a traditional US image) at 15 frames per second [134]. For radial images, the transmitter fibre can be polished to 45°, coated with a mirror finish and rotated along the fibre axis to acquire a radial B-scan (similar to those acquired with intravascular ultrasound probes [137]). Through the use of a robot manipulator, and scanning the probe across the target, a B-scan or C-Scan (volumetric data) can be obtained (as presented in Chapt. 4 and [119]). However, this approach increases the size of the instrument and introduces large acquisition times when obtaining the imaging information.

Dedicated scanning mechanisms have been applied to other modalities with limited fields of view or providing only one dimensional information: namely endomicroscopy and optical coherence tomography (OCT). In each example, the aim was to increase both the field of view and image acquisition rates [6, 7, 110, 111]. Microscopy is often the targeted modality for scanning mechanisms due to the high resolution, despite the limited field of view (approximately 2 µm and 1 mm respectively). The approach is problematic in terms of obtaining clinically useful information. However, scanning mechanisms attempt to improve this by moving the imaging probe across the target in a repeatable trajectory.

This chapter presents the use of a concentric tube manipulator as a scanning
5.2 Instrument Design

A new instrument was designed using the experience gained from the previous device, as presented in §3.2. The main modifications to the instrument are in three
5.2. Instrument Design

areas: miniaturising the overall size; incorporating linear rails into the carriage section; integrating the carriage section into a modular housing. The core purpose of these changes was to accommodate the integration of an imaging probe within the instrument. However, a number of other, smaller, changes were made to improve the overall reliability and robustness of the system. For example, the housing around the mechanism provided structural support for the mechanism and protection to the components, in addition to protecting the imaging probe within the instrument. The linear rails also reduced the load on the nitinol tube and square shaft, thus reducing the torque required to drive mechanism; this, in turn, allows for smaller profile motors to be used and subsequently the overall profile of the mechanism to be reduced.

5.2.1 Mechanism Design

Following the previous design, the instrument can be separated into two sections: motor pack and carriage pack. The motor pack has been expanded to hold four actuators. (For this application, it should be noted, only a single tube is required and the control of two tubes will not be considered further within this thesis.) The actuators to drive the mechanism are arranged about the central axis of the tube and have been changed to now incorporate stepper motors. Stepper motors were selected due to their high repeatability and the ability to apply relatively high torque at low speeds. The motor pack is joined to the carriage pack through 2 alignment pins and the motion of the actuators is transferred through Oldham couplings. Oldham couplings consist of 3 components: 2 hubs, typically metal, that couple to the input and output shafts and have a single rail across the face of the hub; a disc, constructed typically from a plastic with low friction. The disc has a slot for the rail on each face and the slots are offset 90° from each other. This arrangement allows for some translational error in the positioning of the shafts as the disc will slide along the rails to compensate. The key advantages of the Oldham coupling are: almost no backlash; the use of a disc between the hubs operates as a torque limit for the transmission (if too much torque is applied the disc will break allowing both shafts to move freely without colliding with the other); with only minor modifications, a sterile drape can be placed between the motor pack and carriage pack, thus allowing the instrument
to be sterilised [48]. Figure 5.2 shows the coupling between the motor pack and carriage pack where the alignment pins are extended beyond the Oldham couplings, a design feature incorporated to ensure alignment before the couplings are engaged. The alignment pins are then fixed with a M2.5 cap screw to prevent the packs from decoupling unintentionally.

The main components of the carriage pack are: the carriage (which holds the tube gear, drive gear and lead screw nut); the lead screw assembly; the square shaft assembly. Each shaft assembly consists of the lead screw or square shaft held by a small coupling component that constrains the shaft to the centre of a cylindrical outer profile. Radial bearings hold the coupling component to the housing at each end. The coupling at the proximal side of the carriage section has a 4 mm pin to fix the Oldham coupling and on the opposing side, the shaft profile with grub screw fixings placed radially about the profile. The distal side coupling component has the shaft profile throughout to prevent length alignment issues between the other shaft
5.2. Instrument Design

Figure 5.3: CAD rendering of the carriage pack, showing the shaft assemblies running the length of the pack constrained by bearings at either end and an Oldham hub to couple with the motor pack. The carriage is supported in the centre of the housing by the shaft assemblies; the cylindrical rails run across the centre of the housing and the nitinol tube is within the instrument shaft at the front of the housing (right).

assembly and the housing. The shaft assemblies are held within the carriage housing which has separate front and back plates that hold the shaft assembly bearings and a uniform housing profile. A significant advantage of this arrangement is that it allows for the length of the carriage pack, and therefore the linear joint range of the mechanism, to be easily modified. Fig. 5.3 shows the carriage pack with the carriage and shaft assemblies within. The carriage holds the drive gear (shown green in Fig. 5.3), with the square shaft running through it and allowing the gear to traverse along its length and rotate if the square shaft is driven. The drive gear is meshed with the tube gear (shown blue in Fig. 5.3) that holds the nitinol curved tube, while the leadscrew nut is constrained within the carriage housing and uses the two linear rails and square shaft to act as the runner when the lead screw is engaged. Linear bushings (GSM-0304-03, igus GmbH, DE) provide the interface between the carriage and the linear rails and are friction fit on the sides of the carriage.

The majority of the components were printed using a Mark Two (Markforged
5.2. Instrument Design

Inc., US) in Onyx, with the exception of the central housing of the motor pack and carriage pack, the coupling pins, and the drive and tube gears in the carriage. Onyx is Nylon, mixed with chopped carbon fibre to strengthen the material and this was used for the main components due to the printers higher dimension accuracy (compared to an Ultimaker S5). The material is significantly stronger than PLA and provides heat deflection at a temperature of 145° Celsius. This property is required as the heat deflection temperature of PLA is generally below 60° Celsius and the temperature of stepper motors can readily exceed 60° Celsius. The use of Onyx prevents the deformation of the motor pack whilst under load. The tube and drive gears are made in Iglidur I6-PL (igus GmbH, DE), using a selective laser sintering printer. This material was chosen as it has high abrasion resistance and good sliding properties. Both gears have the same modulus and pitch circle diameter of 0.75 and 18 mm respectively giving a 1:1 gear ratio. The main housing sections are printed from PLA using an Ultimaker S5 (Ultimaker B.V., NL), as the load applied to them is not large and they do not need the heat resistance properties of Onyx (discussed above). The coupling pins were printed using a Form 2 (Formlabs Inc., US) with clear resin and this provided a rigid and smooth finish to ease coupling. The square shaft is stainless steel 5 mm square bar stock, and the lead screw is stainless steel with a 6 mm diameter and 1 mm lead. The lead screw nut is made from iglidur J (igus GmbH, DE) and has a filleted square profile that allows it to slot into the housing with a friction fit; it does not, therefore, require an additional radial locking mechanism, such as a clamp or grub screw.

The stepper motors are NEMA 8 (ST2018, Nanotec Electronic GmbH & Co. KG, DE), and these can provide 3.6 Ncm of holding torque and operate in steps of 1.8°. The motors can apply a maximum torque of 3.0 Ncm at speeds up to 6.7 rev/s but are capable of applying 2.8 Ncm at 10.0 rev/s. The motors have a frame size of 20 × 20 mm and 48 mm length; this allows the mechanism a linear velocity at maximum torque of 6.7 mm/s and a resolution of 5 µm, and a rotational velocity of 6.7 rev/s and a resolution of 1.8°. The housing of the motor section of the instrument has slotted vents along the side and a small fan to cool the motors while
Figure 5.4: Images of the assembled scanning mechanism: **Top** - the coupled carriage and motor section, **Left** - the motor section with the coupling pins, **Right** - the carriage section

in operation. The final assembled instrument, shown in Fig. 5.4, has a total length of 235 mm, consisting of 125 mm and 110 mm for the motor section and carriage section respectively.
5.2. Instrument Design

5.2.2 OpUS Probe Integration

In order to integrate the OpUS probe into the concentric tube mechanism a number of challenges need to be addressed. Firstly, the fibres cannot rotate continuously with the curved inner tube, they must retain their original orientation. Secondly, the fibres need to remain at the tip of the curved inner tube while the tube is translated.

To accommodate the design constraints outlined above, the OpUS probe is removed from its protective polymer tubing and inserted into a section of Nitinol Helical Hollow Strand (HHS) tubing (Fort Wayne Metals Ireland Ltd, IE). The HHS tubing has an outer diameter and inner diameter of 1.26 mm and 0.88 mm and consists of 8 strands of Nitinol formed into a single layer helical strand. The main advantage of HHS is that while it is highly flexible and compliant it exhibits high torsional rigidity. The probe within the HHS sheath is then inserted through the concentric tube to the tip, where the sheath is then fixed in position on the carriage. This allows the probe to translate with the carriage but the orientation is held while the tube is rotating. The sensor assembly at the distal end of the scanning mechanism is shown in Fig. 5.5.
5.2.3 Mechanism Control

Each axis of the mechanism is controlled using a dedicated controller (CL3-E-1-0F, Nanotec Electronic GmbH & Co. KG, DE) and powered from a 12V power supply. Communication to the controller is through the Controller Area Network (CAN) protocol (a multi-master message based communication protocol). Each controller allows the motor to run in open loop mode but with the rotation motor running in velocity mode and the linear motor running in profile position mode.

The arrangement, as described above, provides the control in actuator space but the mechanism information needs to be accounted for in order to move in joint space. Equation 5.1 shows the mapping between actuator values and joint values, where \( q \) is the joint state, \( k \) is the gear ratio, \( A \) is the actuator state and \( \lambda \) is the zero offset. The actuator state is the angle of the actuator and the zero offset is the actuator angle at the joint zero position. The gear ratio for the rotation joint is the gear ratio between the drive and tube gears within the carriage, and the linear joints the gear ratio is the lead of the lead screw. The joint state is composed of the rotation of the tube and extension of the inner curved tube from the outer tube (steel instrument shaft).

\[
q = k(A + \lambda)
\]  

(5.1)

For the assembled mechanism, this resulted in a gear ratio of 1.0 and 0.001 for the rotation and linear joint respectively. A forward kinematics solver with backlash compensation was implemented using the domination stiffness method [1]. Each joint position is compared with the previous position; this means that if the direction of the motion has changed, the compensated position remains unaltered until the difference in position has exceeded the backlash value. The compensated position is then used to calculate the tube transform, as previously shown in §3.3.2.

5.2.4 Optical Ultrasound Processing

The processing of the optical ultrasound was initially presented in §4.4.2. However, closer integration of the scanning mechanism with the kinematic information could significantly improve the imaging results, as discussed previously. During
operation, the unprocessed A-scans and kinematic information are output as ROS messages with timestamps. These are recorded as rosbags, a file format that stores ROS messages with metadata such as receipt time. The completed scans are then post-processed in Matlab using the following procedure. The kinematic information is synchronised with the unprocessed A-scans using the timestamps; as the A-scans are published at approximately 100 Hz while the joint states are published at 500 Hz; the joint states are compared to the A-scans and the joint state closest in time to each A-scan is found. Following the synchronisation, the A-scans are processed through frequency filtering, envelope detection and log-compression; this stage of the OpUS signal processing is presented in further detail in [137, 138]. The output processed A-scans are transformed using the matched kinematic pose from the sensor frame to a common coordinate frame which, in this instance, is the tip of the outer tube. Each output A-scan is formed as a point cloud and sequentially added together to form a C-scan with a cone shaped volume. After the addition of an A-scan, a 10 µm voxel filter is applied to the C-scan preventing large sections of overlapping data. Finally, an intensity threshold is applied to the constructed C-scan and values empirically optimised for display.

5.2.5 Scanning Mechanism Calibration

With the mechanism and controller assembled and operational, a number of unknown parameters remain preventing accurate imaging. The first parameter to be determined is the position of the OpUS probe relative to the tip of the nitinol tubing, the second is the nitinol tube curvature; the third is the free axial motion or backlash between the OpUS sensor and the linear joint. However, the backlash of the rotation joint is not a consideration because the direction of rotation is constant. Solving the first two problems involves identifying of the kinematic parameters of the concentric tube robot, while the third is a combination of actual backlash from the lead screw driving the carriage and elastic effects from the curved nitinol tube and torque coil. These parameters can be tuned by imaging objects with a known geometry. For example, imaging a plane will allow for backlash compensation as there will be two constructed planes from each linear direction of travel, while the
tube curvature can be seen from the curvature of the plane.

The scanning mechanism is used to image a phantom with known geometric properties: in this case, a planar surface and a surface with a line 3 mm wide and 5 mm high through the centre (both 3D printed using an Ultimaker S5). The forward kinematics for each joint state, synchronised with an A-scan, are recalculated over a range of curvatures and backlash values. The A-scans are then processed following the procedure presented in the previous subsection (§5.2.4). For both calibration phantoms, the low intensity points are then filtered out to remove the noise and the points from the underside of the phantom. The points are then fit to a plane and the error of the fit calculated, with the parameters that provide the minimum fitting error used as the kinematic parameters. In the case of the line phantom, two surfaces can be found (the base of the phantom and the raised line); the width of the raised line can also be used to determine the parameters further.

5.3 Experiments and Results

5.3.1 Scanning Mechanism Trajectory Assessment

As the mechanism is designed to consistently follow the same spiral trajectory, the repeatability of the motion needs to be assessed. To enable this, the OpUS probe is removed from the torque coil and replaced with an EM tracking sensor. The sensor is 11 mm long and has a diameter of 0.8 mm, allowing it to be inserted into the torque coil at the tip. Fig. 5.6 shows the sensor and the overlapping section of torque coil extended from the nitinol tube. This is necessary to prevent deformation of the sensor from the interactions between the nitinol inner tube and steel outer tube.

With the sensor attached at the tip of the mechanism, the mechanism is set to follow a repeated scanning trajectory: where the extension of the tube, from the linear joint, is set to travel between 0.0 mm and 10.0 mm with a trapezoidal velocity profile where the maximum velocity is 10 mm/s, the maximum acceleration is 5 mm/s²; while, the rotational joint is set to rotate the tube at a constant velocity of 2 rev/s. The scanning mechanism is set to follow this trajectory for 20 min-
utes and the joint states and tracker transforms are timestamped to the system timer and recorded. Once completed, the tracker positions are aligned with the joint states through the timestamps (the tracker outputs at approximately 40 Hz while the mechanism controller outputs joint states at approximately 500 Hz). The synchronisation procedure results in 48,021 aligned joint states and transforms. Each joint state is then compared to all the other joint states and, if they match within 0.2° and 50 µm, their indices are added to a list and the count for the index increased. Once all the joints were grouped, the duplicate groups were removed by sorting by size, then removing any duplicate index. This approach resulted in 1465 unique groups with at least 4 matched joints in each group. The respective transforms of the matched groups were then taken and the standard deviation across the groups was calculated, resulting in an average repeatability of 145 µm, and a maximum and minimum of

![Figure 5.6: The experimental setup used to assess the repeatability of the scanning mechanism. The EM sensor is held in the torque coil which has been extended out from the curved nitinol tube.](image)
Figure 5.7: The repeatability measurements of the scanning mechanism, using 7632 transforms assigned to 1465 groups. The measurements are displayed against the extension of the linear joint. The average repeatability of 145 µm, and a maximum and minimum of 793 µm and 17 µm respectively. (Fig.5.7 shows the measured repeatability over the linear range of the mechanism.) Additionally, tracked poses from the EM tracker were recorded over the range of linear motion in a scan but at a single tube orientation. A sphere was fitted to the points to measure the bending radius of the curved tube: the fitted sphere had a radius of 49.5 mm with a mean error of the distance between the points and the model of 91 µm.

5.3.2 Phantom Scanning Setup

Fig. 5.8 shows the experimental setup for acquiring the scan data using the mechanism described here. The scanning mechanism is held over a tank of water with a 3D printed phantom holder at the base of the tank. The phantom holder has a square tapered centre with an 6.5 mm hole through the centre. Each phantom has a square footprint, with the underside tapered and a tapped M6 hole in the centre, allowing for an M6 cap screw to fix the phantom to the holder. The mechanism is positioned so the instrument shaft is orientated approximately normal to the phantom surface, centred on the phantom and raised 20 mm above it. Scans are acquired over 45 seconds with the mechanism running at 2 rev/s on the rotation joint and, for the linear
5.3. Experiments and Results

5.3.3 Scanning Mechanism Calibration

Two phantoms are used to acquire data for calibrating the scanning mechanism: a planar sheet and a planar phantom with a raised line through the centre. The joint, at between 0.0 mm and 10.0 mm with a maximum velocity of 10 mm/s; the maximum acceleration is 5 mm/s².

Figure 5.8: Top - the experimental setup for acquiring scans of phantoms: with the phantom held in the mount and the scanning mechanism positioned above (~ 20 mm) and the centre of the phantom with the instrument axis normal to the phantom. Bottom - the range of printed imaging phantoms used to assess the quality of the C-scans. From the left: line phantom, cross phantom, stepped phantom and multiple width cross phantom.
5.3. Experiments and Results

Precalibrated data of the planar phantom is shown in Fig. 5.9 and shows each A-scan resolved against time, and the A-scans transformed to the pose of the tip of the scanning mechanism from the initial kinematic parameters and merged over time to form a C-scan. These assume the curvature of the tube conforms to the curvature of the jig in which the nitinol was set (35 mm) and that the mechanism has no backlash.

For both phantoms, the calibrated parameters are found through a search over both the backlash and the tube curvature. The backlash range was from 0 mm to 2 mm, in 10 µm increments, while the tube curvature is from 35 mm to 45 mm in 500 µm increments. For the planar phantom, a plane is fit to the constructed C-scan and the planar error is used to assess the tube parameters. The fitted error against both the backlash values and curvature values are shown in Fig. 5.10. The minimum errors for the planar calibration were 65 mm and 1.42 mm for the tube curvature and backlash, respectively. For the line phantom, Fig. 5.11, shows the
5.3. Experiments and Results

Figure 5.10: The error of fitting a plane to the constructed C-scan of the planar phantom: **Top** - average fitting error against tube curvature; **Bottom** - average fitting error against mechanism backlash

height and width error with respect to the tube curvature and backlash separately. The set parameters that resulted in the minimum error from the line phantom were **53.5 mm** and **1.49 mm** for the tube curvature and backlash respectively.

### 5.3.4 Reconstruction Assessment

The quality of the reconstructions is then assessed through the use of a range of 3D printed phantoms. The phantoms used are: a plane, a raised line, a raised cross, a raised cross with multiple widths, and a stepped phantom. The first two phantoms demonstrated are the same used for the calibration but the scans presented are not from the calibration dataset. The raised cross is same the width as the line but along two axis. The multiple-width cross was designed with widths of **1, 2, 3, 4 mm**, while the stepped phantom has a border around the outside of the phantom; in the centre a grid layout, four levels are set at **2, 4, 6, 8 mm** and offset from the border. Figure 5.8 shows the phantoms designed to assess the C-scans, each phantom was printed using an Ultimaker S5 using a **0.25 mm nozzle** and PLA, the phantoms were then measured with digital calipers (Axminster Tool Centre Ltd, UK), with a resolution...
5.3. Experiments and Results

Figure 5.11: The two error metrics of the line calibration: **Left column** - height error between two fitted planes of the line and the base of the phantom; **Right column** - the error between the width of the top surface of the C-scan and the width of the physical phantom top surface. For each column, **Top** shows the average fitting error against tube curvature, and **Bottom** shows the average fitting error against mechanism backlash of 0.01 mm, to determine the physical dimensions. The cross and line phantoms had a measured width of 3.11 mm and a height of 5.02 mm. For the multiple-width cross phantom the measured dimensions, starting from the smallest, were 0.86 mm, 2.36 mm, 2.91 mm, 4.30 mm. For the stepped phantom the measured heights were 2.02 mm, 4.00 mm, 6.01 mm, 8.02 mm.

Fig. 5.12 shows the C-Scan of the plane in both XY and XZ projections. The top surface of the plane (shown in blue in Fig. 5.12) is manually segmented from the bottom surface and is assessed in relation to two considerations: how planar the surface is and the deviation from the mean. A plane is fit to the surface where the mean and median distance of the points to the plane were 0.19 mm and 0.14 mm, with a maximum distance of 0.76 mm. The standard deviation of all the points is 0.15 mm. Fig. 5.13 shows the line and cross phantoms in XY projection, with the Z depth shown by the colour of the points. For the line phantom, the top and bottom surfaces are assessed in the same way as the plane; but in addition, the width of the
5.3. Experiments and Results

<table>
<thead>
<tr>
<th>Phantom</th>
<th>Plane</th>
<th>Line</th>
<th>Cross</th>
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<tbody>
<tr>
<td>Base plane deviation (mm)</td>
<td>Mean</td>
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<td>0.31</td>
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<tr>
<td></td>
<td>Median</td>
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<td>Max</td>
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<tr>
<td>Top plane deviation (mm)</td>
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<tr>
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<td>Median</td>
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<tr>
<td></td>
<td>Max</td>
<td>N/A</td>
<td>0.99</td>
</tr>
<tr>
<td>Distance between planes (mm)</td>
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<td>5.03</td>
<td>4.94</td>
</tr>
<tr>
<td>Percentage of points within template</td>
<td>N/A</td>
<td>98.36%</td>
<td>95.90%</td>
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Table 5.1: Metrics on scan quality for the plane, line and cross phantom.

raised line is assessed along with the distance between the top and bottom planes. For the top plane, the mean and median distance of the points to the plane are 0.20 mm and 0.14 mm with a maximum distance of 0.99 mm. The standard deviation of all the points was 0.19 mm. For the bottom plane, the mean and median distance of the points to the plane are 0.31 mm and 0.25 mm with a maximum distance of 1.09 mm. The standard deviation of all the points was 0.23 mm. The distance between the fitted planes is 5.03 mm. It was found that 98.36% of points lie within a width of 3.11 mm and 99.01% of points lie within a width of 3.66 mm.

For the cross phantom, the top and bottom surfaces are assessed in the same way as the plane; in addition, the width of each strut of the cross is assessed along with the distance between the top and bottom planes. For the top plane, the mean and median distance of the points to the plane are 0.26 mm and 0.21 mm with a maximum distance of 0.92 mm. The standard deviation of all the points is 0.20 mm. For the bottom plane, the mean and median distance of the points to the plane are 0.31 mm and 0.28 mm with a maximum distance of 0.94 mm. The standard deviation of all the points is 0.22 mm. The distance between the fitted planes is 4.94 mm. The cross is found using a moving filter which searches through X and Y coordinates to find the maximum number of points in a cross centred on the X and Y. Using the width of the printed cross, 3.11 mm, 95.9% of the points were within the template, while 99% of the points fit within a cross of 3.6 mm. The results for the plane, line and cross can be seen in Table. 5.1.

Fig. 5.14 and Fig. 5.15 shows the multiple width cross and stepped phantoms.
5.4 Discussion

The mechanism presented in this chapter is a continuation of the design presented in §3.2. The overall size of the instrument has been reduced and the mechanism
Figure 5.14: Top Row - CAD renders of the stepped phantom model, coloured to the C-scan data; Angled view (Left) with the border sectioned out and Top down view (Right). **Bottom Row** - C-scans of the stepped phantom shown from two views, side on YZ projection (Left) and Top down XY project (Right). The colour of the points is encoded to the depth of the scan; while the size of the points is the intensity of the signal recorded.
Figure 5.15: **Top Row** - CAD renders of the multiple width phantom model, coloured to the C-scan data; Angled view (Left) and Top down view (Right). **Bottom Row** - C-scans of the multiple width phantom shown from two views, side on YZ projection (Left) and Top down XY project (Right). The colour of the points is encoded to the depth of the scan; while the size of the points is the intensity of the signal recorded.
enclosed within a housing for easier handling and protection of the components; in addition all the electrical components in the motor section are routed into a single multi-core cable back to the controllers and power supply. The mechanism to couple the motor section and carriage section has been changed to Oldham couplings. These allow easy alignment in coupling and the central disc can act as a sacrificial element in case of jams and breakages (however, as part of this application, custom discs would need to be manufactured for the appropriate torques). In previous work, these couplings have been modified to transmit torque through a sterile barrier, allowing the carriage section to be sterilised (using an autoclave) and the motor section to be draped [48]. As also noted in previous designs, the nitinol tube and square shaft described here acted as the rails for the leadscrew; this arrangement applied a load to both elements. In turn, it increased the torque needed to drive the rotational joint and also adversely affected the offset between the nitinol tube and steel instrument shaft. To counteract this undesirable outcome, linear rails have been introduced on either side of the nitinol tube: this maintains concentricity between the nitinol and steel instrument shaft and reduces friction while in motion.

One of the main limitations of the instrument presented in chapter 3, was that the instrument could not rotate continuously and, thus, required the orientation to be reset or "wound back" when approaching its operational limits. This is a fundamental flaw for a scanning mechanism as the purpose is for a scanning trajectory to be in operation continuously allowing sustained scan acquisitions. However, the use of a torque coil allowed for the imaging sensor to be translated with the curved nitinol tube and remain at the tip of the instrument; while remaining uninfluenced by rotation of the nitinol tube itself. When coupled to the lateral flexibility and torsional stiffness of the torque coil, there is also an axial elastic element to the design. This introduces some ‘uncertainty’ into the sensor position relative to the nitinol tube’s tip position because the torque coil is fixed relative to the tube at the carriage, rather than at the tip.

A key aspect of scanning mechanisms is the repeatability of the motion [139, 110]. The repeatability of the scanning mechanism was assessed using an EM
tracker; the sensor being designed to fit directly within the channel of the torque coil. The tip of the instrument was tracked for 20 minutes, collecting 7632 sensor transforms and joint values. However, as Fig. 5.7 shows, the majority of the grouped transforms are at the limits of the linear joint. A potential reason for this is that the tracker outputs transforms at 40 Hz and the linear joint of the mechanism runs a trapezoidal velocity profile, with an acceleration and deceleration of 5 mms$^{-2}$: the sensor will therefore spend more time close to the joint limits. The repeatability measurements at the limits, compared to within the centre, are also significantly different. This is most likely attributable to the EM sensor tracking losing some stability at higher velocities, thus increasing the error of the measurements [140]. A higher accuracy tracker could undoubtedly provide more information but would not be without limitations: the tracker used in Chapter 4, for example, utilised markers of 8 mm diameter; almost 4 times the size of the instrument’s diameter. Such an arrangement could not be mounted rigidly on the instrument described here without deforming the motion of the instrument due to the load.

The OpUS calibration allows for an approximation of the configuration of the mechanism. While the curvature of the nitinol tube can be measured, the position of the sensor relative to the torque coil, the position of the torque coil compared to the nitinol tube and the zero position of the nitinol tube are significantly harder to measure physically. The calibration procedure therefore attempts to compensate for these unknowns by imaging known objects and assessing the reconstruction from certain joint parameters from two parameters: these are backlash (the lost axial motion from changing directions) and curvature (the path followed by the sensor when extended). Two phantoms models were developed specifically for the purpose of calibration. Due to the geometric simplicity of the model required, the first used a plane and is shown in Fig. 5.10, the minimum error for the backlash value can be identified with relative ease. The error with respect to tube curvature is considerably noisier, and the error gradient moves to zero as the curvature increases. In contrast, the use of the line phantom allowed the identification of the optimum parameter for both the tube curvature and the backlash; an arrangement offering a significant ad-
vantage here. This is possibly due to the reliance of each metric on opposing axes of information (because the width of the line requires lateral accuracy and this is mostly affected by the tube curvature). Whereas, the distance between the line and the base relies strongly on the backlash value. The parameters identified here using the line phantom were 53.5 \textit{mm} and 1.49 \textit{mm} for the curvature and backlash respectively. The width error and height error were 85.7 \textit{\mu m} and 76.1 \textit{\mu m} respectively.

The calibration was then assessed using four printed phantoms with increasing complexity, as shown previously in Fig. 5.8. The raised line and cross phantoms could be distinguished clearly from the constructed C-scans. The error in the height offset between the top and base surfaces, as compared to the measured value, was 10 \textit{\mu m} and 80 \textit{\mu m} for the line and cross respectively. The points on the top level of the C-scan fit the measured geometric model of the line phantom 100%. For the cross phantom, 95.9\% of the points were within the measured geometric model. All of the points were fit within a width of 3.60 \textit{mm}, and the points outside of the measured width can be identified visually as close to the corners of the cross. For the stepped phantom and multiple width phantom, the levels of the phantoms can be distinguished readily, as seen in the bottom left of Fig. 5.15 and Fig. 5.14. In the stepped phantom reconstruction, four separated levels can be discerned: they are coloured as dark blue, light blue, green, and red, with the top of each level separated by approximately 2 \textit{mm} and fitting with the printed phantom. Similar axial detail in the cross phantom can be seen with the separation between the two planes of approximately 5 \textit{mm} which matches to the geometry of phantom. However, the actual structure of either is not easily interpreted from the reconstruction: the cross phantom displays a single strut of high intensity points along the positive Y axis but no other strut is discernible. This, lack of lateral information, is also apparent in the stepped phantom, where the top two surfaces and the bottom surface can be identified, the bottom surface being the base of the phantom but the 2 lower levels of the phantom only being seen from a side view.

There are a number of possible reasons for this. One is a misalignment of the phantom under the scanning mechanism, attributable to the fact the phantoms are
currently placed in view of the scanning mechanism manually (which is assessed from visually inspecting the live A-scan over a single scan iteration). This would correspond to the views seen if the phantom was offset from the central axis of the instrument. Additionally, increasing the offset in the Z axis between the phantom and scanning mechanism would increase the field of view and potentially show a larger portion of the phantom. However, an increase of the distance also results in a reduction of signal and, possibly, movement of the phantom out of the depth of field when the sensor is at the lower limits of the linear joint. Another possibility is that the pose of the OpUS sensor determined by the kinematics may be inaccurate, either from uncompensated motion of the sensor from within the torque coil during the scanning trajectory or from the calibrated kinematic model. The sensor is currently not constrained within the torque coil at the distal end, therefore, it can still move within the coil. An additional consideration is that the current kinematic model assumes there is no deformation of the steel instrument shaft, and no torsion in the nitinol tube. Both of these factors may introduce torsional "wind-up" when the sensor is at the lower limits of the linear joint, which is then released when the sensor is at the upper limits of the linear joint. This would cause a difference in the tube rotation depending on the linear position. Lastly, degradation of the sensor over time could potentially be effecting the imaging: the OpUS imaging probe, it is noted, has been used in multiple projects including for the experiments presented in Chapter 4.

The focus of this chapter has been on developing the scanning mechanism and implementing a process to generate a C-scan from the data. The acquisition time for the C-scans has not been fully explored, as the C-scans were generated offline from the recorded data. However the aim for this mechanism is to provide high speed scans of the surgical environment, the acquisition rate is an important aspect. The acquisition rate is dependant on the scanning trajectory parameters. Currently the scanning trajectory is set so that the imaging probe goes between the linear limits in approximately 5 seconds, in which time the tube has rotated around 2.5 times. However, the probe does not then follow the same cartesian position, though it does
follow a set joint trajectory for each joint. This results in the density of the scan increasing over time, limited by the lateral resolution of the OpUS probe. One of the key advantages in this approach is tuning the C-scan area and acquisition rate through the tube parameters (tube curvature and arc length) and the scanning trajectory parameters (linear joint position and velocity profile, and the rotational joint velocity profile). Additionally, external motion compensation is another avenue that has not been explored within this section. The scanning mechanism currently only holds the OpUS probe and the C-scans are generated in reference to the instrument shaft. With a scan across the linear joint limits currently taking 5 seconds external motion would adversely affect the scan quality. One potential solution to this would be to increase the speed of a single scan, then employ a motion model when registering multiple scans together. This would allow sparse scans to be generated at a high rate with minimum distortion from motion, then registering multiple of these scan together while compensating for the motion could enable the generation of dense C-scans using a similar technique as presented here [141].

5.5 Conclusions

This chapter has presented the design, fabrication and initial characterisation of a concentric tube based scanning mechanism. A new concentric tube mechanism has been designed that differs from that presented in Chapter 3. The instrument follows a similar modular design as presented previously; however, the coupling method now uses Oldham couples, with two pins either side to align and transmit torque between the carriage section and the motor section. The carriage is supported by linear rails either side of the nitinol tube to reduce load on the tube. An OpUS probe was integrated into the instrument through the use of a torque coil, and this allowed the nitinol tube to rotate continuously without passing the torsion onto the sensor. The repeatability of the mechanism was assessed by replacing the OpUS sensor with an EM sensor and have the mechanism scan for 20 minutes. The repeatability was found to average 145 μm but increased with positions recorded at higher velocities. A calibration procedure was presented that relied on a small phantom
5.5. Conclusions

incorporating a raised line in running through the middle, and this allowed the kinematic parameters of the instrument to be identified. The quality of reconstructed C-Scans were then assessed using four phantoms, each with increasing geometric complexity. The C-scans for the first two phantoms (the line and the cross) provided clear and accurate scans: here, the planarity of the surfaces averaged $0.27 \text{ mm}$. For the later phantoms, the axial information can be seen clearly; however, the lateral detail remains unclear. These results illustrate the potential utility of the concentric scanning mechanism for fast acquisition of C-scans utilising OpUS sensors. The design of the instrument means it is highly adaptable both in terms of the density and the field of view of the scans obtained.
Chapter 6

Conclusion

This thesis has developed novel robotic instrumentation solutions to actuate, navigate and image the anatomy during fetoscopic interventions. The clinical requirements were presented in the first chapter. This was followed by a review of distal actuation mechanisms applied previously to the specific requirements of fetal surgery. Fetal surgery is a relatively new field with few hospitals offering procedures; those fetoscopic approaches that are available are technically extremely challenging. Despite the potential to reduce the morbidity to the mother and the fetus, open approaches are considered the gold standard due to the practical and clinical advantages they offer (discussed in detail at Chapter. 1).

6.1 Contributions

Chapter 3 presented an exploratory approach to the introduction of distal DoF and constrained proximal DoF to an endoscope. Here, the distal DoF was provided by a single tube concentric tube robot and the proximal DoF was provided by a 7 DoF articulated manipulator, constrained to a RCM allowing 4 DoF of motion. This provided the DoF required for the orientation of the tip of the endoscope to be decoupled from the position. A concentric tube mechanism was chosen because the structure of the endoscope also comprised the actuation element: an arrangement devised to provide a constant diameter working channel throughout the instrument. Additionally, the maximum curvature of the tube is set at the design stage, thus enabling the bending limits of the components operated through the channel to be
accounted for. The single tube mechanism designed was fabricated using a combination of 3D printed components and commercial off the shelf items. The design allowed for the nitinol curved tubes to be easily decoupled from the mechanism and, additionally, for the motors to be detached. Two end-effectors were made for the mechanism: one to hold an electromagnetic tracker and one to hold a small stereo camera. The proximal manipulation was provided by an industrial robot (KUKA LBR iiwa 7 R800) and the motor section of the concentric tube mechanism connected to the end-effector of the robot. The proximal manipulator was then kinematically constrained to a RCM using the robot controller. Proximal and distal controls were then coupled and demonstrated by performing a selection of scanning trajectories; during this process, the endoscope was constrained to a fixed orientation, first over a phantom placenta and subsequently a real ex vivo placenta. Mosaics were constructed using data collected from the scans demonstrating the stability of the system. The accuracy and repeatability of the trajectories were also assessed and determined to be 3.10 ± 11.91 mm and 22.31 ± 5.93° for the spiral scans and 0.77 ± 0.25 mm and 10.17 ± 0.59° for the raster scans. These results presented the potential improvements possible through the combination of proximal and distal manipulation: with the proximal motion improving stability and the distal motion increasing dexterity. However, a number of limitations were noted: control of the articulated robot remained quite limited; the control architecture made adjusting trajectories (either user based or from sensor feedback) impractical; the concentric tube mechanism was a prototype design of limited robustness and a compromised end-effector design.

Chapter 4 focused on improving the control of the proximal elements of the system. The previous chapter set out the necessary reliance on the built in control functionalities of the sunrise controller, which limited the level of control possible. For example, sensor integration was not readily possible and there was no path planning available; while the robot had 7 DoF, the redundant DoF is represented as the joint angle of the third joint, limiting the workspace and potential solutions for a given pose. The control of the robot was therefore re-implemented within a robotics
framework, ROS: this allowed the integration of a motion planning framework and simplified the application of kinematic constraints (such as the RCM constraint and limiting the workspace). This approach also allowed closer integration with the imaging sensors, enabling the collection of synchronised imaging and kinematics data to improve scan quality, and allowing the control to be robot agnostic. A rigid endoscope with a stereo camera, optical ultrasound sensor, 0.9 mm working channel and fibre guide for lighting, with an overall diameter of 6.2 mm, was fabricated. The tip of the endoscope was produced in stainless steel using a metal laser sintering 3D printer. The tip was laser welded to a stainless steel tube (i.e., the instrument shaft) which was held in a plastic housing allowing it to be coupled to the robot. The utility of the endoscope was demonstrated by scanning a placenta phantom, using both raster and spiral scans. The optical ultrasound data collected was constructed into C-scans (using the kinematic data) and produced large area 80 × 80 mm scans. The accuracy and repeatability of the endoscope was also assessed using an optical tracker: 100 positions were randomly generated and each position was repeated 20 times. The repeatability of the endoscope was 0.54 ± 0.27 mm, the controller and kinematic accuracy was 1.35 ± 0.62 mm and 1.21 ± 0.59 mm respectively. Additionally, the endoscope performed the same scan multiple times and the C-scans generated were compared to assess the scan repeatability. The scans were repeated 10 times each and the repeatability was 0.446 ± 0.139 mm and 0.267 ± 0.017 mm, for the raster and spiral scan respectively. The scan quality allowed vessels to be clearly delimited from the surface of the placenta phantom. However, it was noted that scan quality deteriorated as the angle between the surface normal and the instrument axis increased which, due to the RCM constraint, is coupled to the radial distance from the incision point.

Chapter 5 presented design improvements to the distal elements of the system. As part of which, the concentric tube mechanism was redesigned using feedback from the design presented in Chapter 3. The modular design of the mechanism was improved through the use of: Oldham couplings, utilised here to transfer torque (with minimal backlash) from the motors to the mechanism; and coupling pins,
used to align the sections before engaging the couplings. Linear rails were also introduced to support the carriage and minimise the load on the nitinol tube. The overall size of the mechanism was reduced and the individual components enclosed within a housing (for enhanced protection). An optical ultrasound sensor was integrated at the tip, through the use of a torque coil as a protective sheath between the nitinol tube and the sensor. The torque coil was constrained in position at the back of the carriage. This arrangement allowed the torque coil to move linearly with the curved nitinol tube but not to rotate with the tube: the sensor therefore maintained a constant axial orientation whilst the mechanism rotated. The mechanism controller was set to perform a repeating trajectory from a constant rotational velocity and a recurring trapezoidal velocity profile between a start and end position, creating an approximately cone-shaped field of view. This could be tailored according to the curvature of the nitinol tube but also adjusted during scanning by altering the trajectory parameters (start and end positions and velocity). The repeatability of the mechanism was characterised by replacing the optical ultrasound sensor with an EM sensor located within the torque coil. The repeatability was assessed from running a scanning trajectory for 20 minutes and collecting 7632 tracker poses with synchronised joint positions. The mechanism repeatability averaged 145 $\mu m$ but ranged between 17 $\mu m$ and 793 $\mu m$, with the higher repeatability measurements generally collected at higher velocities. However, EM trackers have been shown to decrease in accuracy at higher velocities and the overall influence this has on the repeatability measurements have yet to be determined comprehensively.

The optical ultrasound sensor pose in the system was found by performing a calibration procedure over a phantom with known geometrical properties. The sensor pose was calibrated to the domination stiffness model with backlash, leaving two parameters to be found, the curvature and the linear backlash. Within the calibration procedure, two phantoms were used: the first being a planar sheet and the second a planar sheet with a raised line in the centre of known width and height; both were 3D printed in PLA. The calibration was performed by scanning the sensor over each phantom. From this, C-scans were constructed using a range of parameters which
were then compared against the known properties of the phantoms used. The parameters that resulted in the lowest fitting error between the C-scan and phantom model were selected. Once calibrated, the scanning mechanism was assessed using four 3D printed phantoms of increasing geometric complexity. The line and cross phantom C-scans produced geometrically accurate models of the phantoms, with > 95% of the points lying on the surface of the printed model. In contrast, while the multiple-width cross phantom and stepped phantom each displayed clear axial information, the lateral detail was notably poor. This was most likely attributable to a combination of three factors: the low field of view scanned, the relatively high lateral resolution and not fully centering the phantom under the scanning mechanism. However the results obtained from the repeatability experiments and the initial optical ultrasound C-scans demonstrated the potential utility of the scanning mechanism to acquire C-scans.

6.2 Future Work

The results described here have demonstrated significant clinical potential and suggest several directions for further development. This section will outline a number of potential technical developments, that would improve the performance of the instruments (e.g. improve the scan quality or acquisition speed) and a number of barriers that would need to be overcome in order to be used by clinicians.

6.2.1 Technical Areas

It is intended that future work will therefore focus on four interrelated areas. Firstly, on coupling the proximal control presented in Chapter 4 and the distal scanning presented in Chapter 5. This could allow the system to create dense large-area scans with shorter acquisition time, as the proximal side could provide the macroscale motions (for example a large sparse raster scan), while the scanning mechanism could provide small dense-area scans at each point along the scan. This approach could also allow for motion external to the instrument to be accounted for by including small overlaps (between each scan from the distal mechanism) and registering between the scans before merging. Additionally, the scanning trajectories of the
mechanism could be modified over the scanning area to remain within an angular range of the scanned surface, with a view to maintaining scan quality across the area. The second focus will be on improvements to individual aspects of the proximal and distal systems. On the proximal system, the force applied at the RCM could be measured using either additional sensors or the torque sensors integrated within the joints of the robotic manipulator. This innovation could be used to introduce some compliance to the RCM in an attempt to minimise the force applied. While on the distal system, future work will investigate potential solutions for constraining the distal end of the sensor to hold the fibres at the tip, with the aim of reducing the linear motion of the fibre within the torque coil whilst still allowing rotation. Thirdly, the use of alternative kinematic models, especially ones that are that are capable of modelling the backlash effect more accurately (a limitation observable in the imaging presented here) could improve the quality of the scans. Also, utilisation of kinematic models that account for torsional effects on the tubes and deformation of the instrument shaft could improve the accuracy of the sensor position. The scanning mechanism acquires a cone shaped volume where the current field of view at 18 mm from the tip is a 4 mm diameter. Optimisation of the tube curvature (with respect to the dynamic bending radius of the imaging sensors); and increasing the joint velocity the mechanism can achieve could improve both FoV and the acquisition rate of C-scans. Lastly, integration of white light imaging and interventional components will need to be considered, either by including the sensors within the scanning mechanism or by integrating the scanning mechanism at the tip of an endoscope.

6.2.2 Barriers to Clinic

This section will focus on the areas that would require further development in order to progress to clinical use. However the regulatory process for approving medical devices will not be covered, instead this will cover aspects such as ensuring sterility, the safety of the instrumentation, and the interfaces required for the surgeons to utilise the instrumentation effectively. Sterilisation of the instrumentation is a key aspect for use in surgery. Methods to achieve this can generally be separated
6.2. Future Work

into three categories: sterile barriers, reusable sterilisable, and single-use. Sterile barriers involve covering the instrument in a single use or sterilisable sheet which prevents contact between the patient and the instrument. Sterile barriers are often used to cover pieces of equipment that are too large or sensitive to be sterilised in a different way such as interventional imaging devices such as x-ray imaging systems and robot manipulators. Reusable sterilisable refers to the instrumentation being designed and manufactured in a way that allows the instrument to be sterilised multiple times using techniques such as steam sterilisation using an autoclave or Ethylene Oxide (EtO) sterilisation. Most of the core instrumentation such as laparoscopes and scalpel holders can be sterilised multiple times. Lastly there are single-use items which are manufactured sterile and then packaged to maintain the sterility until they are opened and used for the operation then subsequently disposed of after the operation. Occasionally, these disposable items can be used in a limited number of operations allowing them to be sterilised between each operation.

For the system, as described in the previous section on technical advancements, involving components presented throughout the thesis, multiple techniques would be employed depending on the components involved. For example, the articulated manipulator and the actuation section for the concentric tube mechanism would be draped with a sterile barrier that allows the torque to be transmitted through to the carriage section. This would be the most suitable option due to the large size and components involved preventing sterilisation of the components and single use would understandably not be cost effective. From the design of the concentric tube manipulator, presented in 5.2, the carriage section contains only mechanical components. Minor changes to the materials of these components could allow for it to be sterilised potentially along with the instrument shaft (stainless steel) and the curved tube (nitinol). Nitinol is shape set at 550°C so the temperature increase from using an autoclave would unlikely to influence the set shape [142]. Lastly, the end-effector and sensors integrated need to sterile, a number of sensors can be or can be modified to allow sterilisation in autoclaves or EtO. For example the Naneye imaging sensors used can withstand 10 autoclave cycles. However the OpUS probes,
being a relatively new modality, have not been tested for sterilisation methods. Due to the relatively low cost of optical fibres, they could potentially be assemble sterile within a clean room and shipped as a single use item.

One of the inherent disadvantages in complex instrumentation like this requiring the use of multiple sterilisation techniques is the assembly of the instrumentation before the procedure. This will likely increase the preparation time needed for the procedure and include an additional training requirement for the theatre staff. The assembly process will need to be carefully designed to minimise the time and complexity of the process to reduce the chance of error and the burden on the theatre staff.

The next barrier to clinical use is the safety of the instrument namely, what contingencies are in place, firstly to prevent unintended damaging interactions between the instrument and the patient, and secondly to enable the instrument to be removed from the patient if it fails. This mainly relates the articulated manipulator and concentric tube presented within the thesis. The instrumentation presented within this thesis has focused on the manipulation of imaging probes which do not require contact with the tissue and the instrumentation is relatively lightweight. This enables two potential approaches to preventing damaging interactions. Firstly using a force sensor, or the integrated torque sensors within the articulated manipulator, to assess the force being applied by the instrument providing warnings over a set threshold. Secondly, introducing compliance in the curved tube, either through the physical properties such as wall thickness and material selection or through a mechanical process such as profiling, causing the tube to deform on interaction with the patient reducing the chance of damaging the tissue. For the removal of the instrument in the case of an error, the first step would be to retract the curved nitinol tube if it was extended from the instrument shaft. As the tube tip follows the path of its curvature, this can be done without visual observation from the surgeon. The main obstacle in this is the use of a lead screw for the linear joint of the mechanism as the pitch of the leadscrew prevents the joint from being backdrivable. One potential solution to this is to use a higher pitch lead screw and an actuator capable of higher torques and
lower speeds the motor could then be disconnected for back driving the joint. Once the curved tube is retracted the instrument could be decoupled from the articulated manipulator allowing it be removed manually. As the articulated manipulator does not enter the body, once the instrument is removed the manipulator can be moved away from the patient.

Finally, an interface has to be provided to display the information to the surgeon but also for the surgeon to control the instrumentation. While US imaging has been in clinical use for a long time and has established visualisation methods, the intended use case of OpUS is different and often involves combination with other modalities. One approach is to display the information as an overlay on the endoscope video, showing subsurface information [143] with reference to the endoscope view or providing functional information such as the presence of blood flow in a vessel following a similar approach presented for laser speckle imaging [144]. In terms of how the surgeon will control the instrumentation, the interfaces developed and used within this thesis can generally be categorised as a supervisory control interface either through command line interfaces or simple graphical user interfaces. This interface is appropriate for lab based experimentation or if the instrument were capable of performing the procedure autonomously in a similar use case to the CyberKnife [95]. However for use in surgery with other instrumentation, a teleoperation or shared control interface is likely to be more appropriate. From the surgical setup and the reliance on other instrumentation, a shared control interface is likely to be the more suitable control method. Allowing the surgeons to still be positioned around the patient and interacting directly with the instrumentation.

These areas for development have been based on the use of an articulated manipulator for the motion outside the body and a concentric tube mechanism for the internal manipulation of the sensors. However some simplifications or modifications to the instrumentation could be made to accelerate the entry to clinic. For example, integrating the OpUS probe into a working channel of an endoscope. This could either be used standalone by calibrating the OpUS probe to the image and providing an A-scan to the side of the image with a point on the camera image to
indicate the position of the A-scan. Alternatively, the endoscope could be localised with a tracking system or integrated with a clinically approved surgical robotic system to determine the motion allowing C-scans to be constructed from the motion of the instrument.

6.2.3 Additional Applications

The focus of this work has been towards the use of the instrumentation in fetal surgery. However the instrumentation and techniques could potentially be applied to other clinical disciplines and potentially non clinical applications. One of the key aspects of fetal surgery is that it is performed within the amniotic sac in amniotic fluid which allows the use of US imaging without tissue contact. The work of this thesis could potentially be applied to other surgical procedures that are performed in a fluid environment. For example, urological procedures within the bladder could be targeted to provide guidance in locating tumours. Additionally, intravascular and cardiac surgery could also potentially be targeted however the instrumentation would need to be modified having a flexible actuated instrument shaft rather than a rigid instrument shaft.

Alternatively, procedures that traditionally use gas for insufflation can instead use fluid to expand the cavity. Such as water filled laparoscopic surgery where the abdominal cavity is filled with an isotonic solution, this results in the expansion of the cavity for manoeuvring the instrumentation, in addition to providing a medium to facilitate the use of US guidance. However this approach requires new or modified instruments to provide irrigation and drainage of the fluid, as well as technical challenges from manipulating buoyant organs and the surgical view being obscured from blood diffusing into the solution [145]. A similar water filled approach has also been applied to colonoscopies, leading to reduced discomfort in the procedure but introducing potential visualisation issues from insufficient preparation of the colon [146]. The instrumentation developed could be applied to these techniques to improve the visualisation available during surgery. Outside of surgery, US features heavily in non-destructive testing techniques for identifying defects and flaws in materials [147]. Robotics can assist in this to automate and improve the con-
sistency of inspections. Additionally the use of OpUS and continuum mechanisms could reduce the diameter of the instrumentation used and enable access further internally within the imaging target.
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