

# **UNGROUNDED HAPTIC-FEEDBACK FOR HAND-HELD SURGICAL ROBOTS**

Christopher James Payne, MEng ACGI

Supervisors: Professor Guang-Zhong Yang and Professor Lord Ara Darzi

Department of Computing

Hamlyn Centre for Robotic Surgery

Institute of Global Health Innovation

Imperial College London

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This thesis is submitted to Imperial College London in partial fulfilment of the requirements for the degree of Doctor of Philosophy. Except for where indicated, it presents entirely my own work and describes the results of my own research.

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# Abstract

Surgical robotic technology has evolved over the last few decades: from autonomous systems, to master-slave and cooperatively-controlled *assistive* robots. Whilst these various approaches have proven to be technically successful, clinical adoption of robotic technology remains moderate, largely as a result of the financial cost of such technology. An alternative approach that has been recently explored is the integration of mechatronic technology in to surgical devices that are held by the hands of the surgeon and are unattached to a grounding frame. These ungrounded hand-held devices exploit the existing dexterity of the surgeon's hand that allows them to be simpler, physically compact, lower cost, more easily integrated in to the surgical workflow and with fewer barriers to clinical translation.

This thesis explores the use of mechatronic technology in ungrounded, hand-held surgical tools for the purpose of augmenting a surgeon's haptic perception. During microsurgery in particular, the tool-tissue manipulation forces are often so low that they cannot be perceived by the operating surgeon. This thesis initially proposes a hand-held device that can amplify these sub-threshold forces to magnitudes that can be perceived by human subjects. The mechatronic force amplification concept is further evolved for use in microsurgical forceps designs. In this case, haptic perception is diminished by the elastic spring return of the forceps which is significantly greater in magnitude than the micro-scale manipulation forces. Having investigated the force amplification concept, vibrotactile-based feedback of pre-defined force-thresholds is investigated. The concept is studied through the clinical exemplar of microneurosurgery: a device is proposed which can inform the operating surgeon if they are exerting excessive force based on a force threshold at which iatrogenic injury of neurovascular tissue is known to occur. Finally, an ungrounded force-feedback strategy is investigated for use with a hand-held device that incorporates position-based active constraints of the tool tip.

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# List of Acronyms

CNC – Computer Numerical Control

cRIO – Compact Reconfigurable Input/Output

DAQ – Data Acquisition

DoF – Degree of Freedom

FBG – Fibre Bragg Gratings

FE(A) – Finite Element (Analysis)

FIFO – First-in, First-out

FPGA – Field-Programmable Gate Array

F/T – Force/Torque

IGES – Initial Graphics Exchange Specification

IMU – Inertial Measurement Unit

I/O – Input/Output

JND – Just Noticeable Difference

LED – Light Emitting Diode

LM – Linear Motor

MEMS – Micro-Electromechanical Systems

MIS – Minimally Invasive Surgery

MLE – Maximum Likelihood Estimation

MRI – Magnetic Resonance Imaging

NRES – National Research Ethics Service

OSATS – Objective Structured Assessment of Technical Skill

OCT – Optical Coherence Tomography

PC – Personal Computer

PCI – Peripheral Component Interconnect

pCLE – Probe-based Confocal Laser Endomicroscopy

PID – Proportional-Integral-Derivative

RAMS – Robot Assisted Microsurgery

SEMI – Semiconductor Equipment and Materials International

UDP – User Datagram Protocol

VCA – Voice Coil Actuator

# Chapter 1: Introduction

*“There would be no hesitation in his work, no stumbling, no quivering, no mistakes.”*

This is the description of a robot surgeon of superhuman precision by the visionary science fiction novelist Isaac Asimov in his 1976 novelette *The Bicentennial Man*. It was a pursuit for precision that was the driving force behind the pioneers of the first surgical robots a decade later. Robots were exploited to improve stereotactic needle positioning accuracy in neurosurgery [1] and for precise drilling in orthopaedic surgery [2]. Whilst these pioneering robotic systems were not autonomous humanoids like the robot surgeon in *The Bicentennial Man*, they were designed to execute their surgical tasks *automatically*, which allowed laborious, repetitive tasks to be performed consistently. Indeed, one of the very first robotic platforms to remove tissue from a patient was devised for the automation of transurethral prostate resection [3].

Today, surgical robots are designed to *assist* rather than *replace* the surgeon [4][5]. It is this assistive approach that has seen robotic technology adopted in to clinical practice and across a broad spectrum of clinical disciplines. Robotic systems have been devised to aid orthopaedic surgeons in precisely resurfacing bone based on pre-operative, patient-specific data [6]. The fields of interventional radiology and cardiology have embraced robotic catheter technology that enables radiologists to navigate tortuous vasculature whilst remaining isolated from ionising radiation [7]. In neurosurgery, robotics can allow a surgeon to perform microsurgical interventions within a Magnetic Resonance Imaging (MRI) scanner [8]. At the forefront of clinically-adopted medical robotic technology is the da Vinci™ robotic system developed by Intuitive Surgical™ (CA, USA). It uses a master-slave architecture which is designed to allow complex minimally invasive surgery (MIS) procedures. MIS (or “keyhole surgery”) is challenging for surgeons for a number of reasons including visual-motor misalignment, poor ergonomics and diminished depth perception [9]. The da Vinci™ system is now in its fourth generation following the recent release of the da Vinci™ XI and this technology is now ubiquitous across North America and increasingly so in Europe too.

Despite the da Vinci™ success story, the field of medical robotics is not without its critics. The cost of robotic technology has been a long standing barrier to widespread, world-wide adoption [10]. To take the da Vinci™ as an exemplar, it costs over \$2m, the semi-disposable surgical instruments cost \$1300-\$2300 and have a maximum of ten uses [11] and there are additional maintenance costs. Critics argue that these significant costs cannot be justified given the lack of evidence to prove the da Vinci™ system’s efficacy over conventional MIS [12]. The robotic systems in current clinical practice are expensive for a number of reasons. The lack of competition is likely to be a significant contributor, but the intrinsic complexity of such robotic systems also contributes to their high cost. Complex medical equipment is not only costly to physically manufacture and assemble, it is costly to analyse and approve. Robotic systems like the da Vinci™ must undergo excruciatingly detailed failure mode and effects analyses in order to meet stringent certification requirements. Not only are surgical robotic systems expensive, they are also physically large, a vestige of their industrial origins, imposing infrastructural requirements on the hospitals they inhabit. The set up time of such robotic systems is a further point of criticism [13]. Robotic systems such as the da Vinci™ must be methodically positioned around the patient and draped to maintain sterility. In light

of these barriers to clinical adoption, researchers have looked to develop mechatronic technology that is simpler, smaller and lower cost [10][14]. One way that this can be achieved is by augmenting existing hand-held surgical instrumentation with mechatronic enhancements [15]. These enhanced surgical devices are completely mechanically-ungrounded; they are held by the hands of the surgeon and incorporate specific augmenting functions. This approach allows the surgeon to exploit their existing dexterity so that the role of their hand is not duplicated by a robotic manipulator.

Another challenge for surgery, and robotic surgery in particular, is the provision of effective haptic feedback. The da Vinci™ system for example, cannot provide haptic feedback to the operating clinician. This is because the surgeon no longer directly manipulates their surgical instrumentation but instead, they control their surgical tools through “fly-by-wire” teleoperation which removes all intrinsic haptic feedback. Conversely, hand-held surgical tools already provide haptic feedback to the surgeon as force is transmitted from the tip of the instrument to the fingertips of the surgeon. It is for this reason that haptics in hand-held tools is an area that has not received the same attention from the research community as teleoperated systems. However, the integration of mechatronics in to hand-held devices can allow for various forms of haptic augmentation. These may be of particular significance in disciplines such as microsurgery whereby excessive force application to delicate anatomical structures can cause permanent and irreversible iatrogenic injury to the patient. This thesis explores the use of different haptic feedback strategies in ungrounded hand-held surgical devices to improve force perception, application and collision avoidance.

## **1.1 Thesis Overview**

**Chapter 2** will review the state-of-the-art in hand-held smart surgical devices and advances in haptic feedback systems for surgery. This chapter will discuss the advances in mechatronic surgical technology towards lower cost devices that are accessible to the wider population. An overview of intrinsic human haptic perception is given and the role of haptic feedback in surgery is provided to highlight deficiencies in current clinical practice. Haptic interfaces are reviewed including conventional grounded haptic manipulators and means for providing ungrounded haptic feedback. Surgery-specific haptic feedback methods and devices are also reported in this chapter.

**Chapter 3** investigates an ungrounded hand-held architecture that seeks to augment haptic perception by amplifying tool-tissue forces. The forces exerted during microsurgery can often be below the limits of human perception and so the developed device is configured to scale these forces to levels that are better perceived by a human operator. The device is validated in bench tests which examine its intrinsic force-amplifying ability. A user study was devised to assess the ability of the device to improve human haptic perception in identifying the minimum threshold at which users can detect contact force exerted by the tool tip.

**Chapter 4** builds on the work presented in Chapter 3 by investigating force amplification in microsurgical forceps. Forceps are used in a multitude of surgical procedures, but haptic feedback is significantly hampered by the elastic restoring force of the forceps that contaminates the surgeon's perception of the grasping force exerted by the forceps. This chapter presents two incremental prototypes that build on the engineering limitations of the prototype presented in Chapter 3. Force sensor assemblies were devised to have high sensitivity at low force levels to improve the amplification fidelity of the device. Bench tests demonstrate that scaling factors of 80 are possible and a user study was conducted to examine the minimum force threshold detection with the proposed configuration.

**Chapter 5** investigates the use of tactile feedback to aid a surgeon limit the amount of force applied during microsurgery. The haptic augmentation in Chapters 3 and 4 demonstrate improvements in user perception of force through amplification, this chapter addresses the importance of warning the operating surgeon as to when dangerous force levels are being applied. The dissector device developed provides vibrotactile feedback in order to indicate when a pre-determined force level has been breached. User studies demonstrate that users exert varying force levels and can exert excessive force when distracted. A pre-clinical study was performed on the brain of a human cadaver and demonstrates reductions in force levels.

**Chapter 6** investigates a means by which active constraints with haptic feedback can be incorporated in to an ungrounded hand-held device. In the previous chapters, haptic feedback was provided based on tool-tissue interactions. The device presented in this chapter is

designed to avoid tool-tissue collisions using active constraints for use in surgical scenarios where anatomically-critical areas are to be avoided. The haptic feedback provided is based on the pre-programmed active constraint so that the operating surgeon can perceive the virtual boundary whilst the surgical tool tip is automatically retracted. Bench tests and a user study demonstrate the ability of the device to constrain the surgical tool away from the active constraint and inform the user as to when this virtual interaction has been made.

**Chapter 7** summarises the findings of this thesis. This chapter provides conclusions and offers perspectives on future work directions in to hand-held haptic feedback devices.

The key aims of this thesis are to:

- To develop a means by which tool-tissue forces can be amplified in an ungrounded hand-held device; and to validate the developed methodology in terms of intrinsic performance and in its ability to augment human perception.
- To develop a means by which tool-tissue forces can be relayed to a human operator, specifically in a forceps configuration whereby force perception is otherwise contaminated by the elastic return force in conventional forceps. To validate the developed methodology in terms of intrinsic performance and in a human perceptual study.
- To develop a means by which an operating surgeon can perceive when a force limit is reached when using an ungrounded hand-held microsurgical device. To validate this methodology with bench tests, human perception studies and to demonstrate pre-clinical proof-of-concept.
- To develop a system that can simultaneously provide haptic and position-based active constraints to prevent undesirable tool-tissue collisions. To validate the proposed system with engineering bench studies and with a human perception study.

The work presented in this thesis has resulted in a number of peer-reviewed full conference papers and journal publications:

- **A Smart Haptic Hand-Held Device for Neurosurgical Microdissection.**  
Payne C.J., Marcus, H.J, and Yang G-Z. Annals of Biomedical Engineering, 2015 (in press).
- **Hand-Held Medical Robots.**  
Payne C.J. and Yang G-Z. Annals of Biomedical Engineering, 2014. Aug;42(8). pp1594-605.
- **Hand-Held Microsurgical Forceps with Force-Feedback for Micromanipulation.**  
Payne C.J., Rafii-Tari H., Marcus H.J. and Yang G-Z. IEEE International Conference on Robotics and Automation, Hong Kong. 2014. pp284-289.
- **An Ungrounded Hand-Held Surgical Device Incorporating Active Constraints with Force-Feedback.**  
Payne C.J., Kwok K. and Yang G-Z. IEEE Intelligent Robots and Systems, Tokyo, Japan. 2013. pp2559-2565.
- **A New Hand-Held Force-Amplifying Device for Micromanipulation.**  
Payne C.J., Tun Latt W. and Yang G-Z. IEEE International Conference on Robotics and Automation, St Paul, USA. 2012. pp1147-1152.

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# **Chapter 2: Hand-Held Mechatronic Devices and Haptic Feedback in Surgery<sup>†</sup>**

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<sup>†</sup>Content from this chapter was published as:

**Hand-Held Medical Robots.**

Payne C.J. and Yang G-Z. Annals of Biomedical Engineering, Aug;42(8). 2014. pp1594-605.

The role of the current generation of surgical robots is to *assist* rather than *replace* the operating surgeon [4]. This task of assistance is achieved by combining the respective advantages of the robot and its operator. Robots can achieve high precision, repeatability and dexterity but often lack complex decision-making capabilities and high-level cognitive power that is second nature to humans [5] (Table 2.1). Nonetheless, the very first medical robots devised in the 1980s were autonomous industrial robots that had been applied to clinical problems. These included a stereotaxic needle placement system for neurosurgery [1] and robots for orthopaedic applications [2][16]. Neurosurgery and orthopaedics were the first applications for surgical robots because they could be interfaced with image guidance systems and the rigid nature of the anatomy allowed for autonomous positioning of the surgical tools. Following these developments, it was an autonomous robot that was the first robotic system to remove tissue from a patient in a transurethral prostate resection procedure [3]. Despite these pioneering efforts, it was clear that autonomous robots allowed very little intra-operative control to the operating surgeon. With safety and liability issues being of paramount concern, surgical robots evolved away from autonomous systems and towards assistive platforms that give the operating surgeon considerably more control over their interventional tools. This was achieved through the parallel development of “hands on” cooperatively-controlled robots and tele-operated, master-slave systems.

In a grounded, cooperatively-controlled system, the robot and the surgeon hold the surgical tool together. This enables the surgeon to have real-time control of the surgical tool, whilst the robot assists by steadying the tool position or by providing haptic guidance. The Acrobot™ [6] is an example of this synergistic control approach for use in knee replacement surgery. The robot physically restrains the surgeon from milling beyond a preoperatively-defined optimum boundary that is registered with the patient’s knee. Cooperatively-controlled robots have also been developed for microsurgical applications. The Steady-Hand robot, developed by Taylor *et al* allows tremor-free manipulation of a surgical tool and the system provides scaled force-feedback that augments the surgeon’s sense of touch [17][18].

**Table 2.1: The strengths and limitations of humans and robots. Adapted from Howe and Matsuoka [5].**

|                    | Humans                                                                                                                                                                                                                                                                                | Robots                                                                                                                                                                                                                                            |
|--------------------|---------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------|---------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------|
| <b>Strengths</b>   | <ul style="list-style-type: none"> <li>• Cognition and judgement</li> <li>• Hand-eye coordination</li> <li>• Adaptable</li> <li>• Able to use qualitative information</li> <li>• Dexterity at human scale</li> <li>• Easy to instruct and debrief</li> </ul>                          | <ul style="list-style-type: none"> <li>• Positioning accuracy</li> <li>• Repeatability</li> <li>• Unfatiguing</li> <li>• Can operate at different physical scales</li> <li>• Immune to radiation and infection</li> <li>• Sterilisable</li> </ul> |
| <b>Limitations</b> | <ul style="list-style-type: none"> <li>• Exhibit inherent physiological tremor</li> <li>• Limited accuracy</li> <li>• Susceptible to ionising radiation and infection</li> <li>• Dexterity outside of human scale</li> <li>• Prone to fatigue</li> <li>• Limited sterility</li> </ul> | <ul style="list-style-type: none"> <li>• Lacking judgement and intelligence</li> <li>• Expensive</li> <li>• Challenging to manufacture and debug</li> <li>• Difficult to guarantee robustness</li> <li>• Technology in flux</li> </ul>            |

Tele-operated robots allow the surgeon to remotely control a patient-side slave robot through manipulation of a master manipulator at an operating console. This robotic approach is particularly useful for minimally invasive surgery (MIS). MIS is conventionally challenging to perform because of the instrument fulcrum effect, visual-motor misalignment and diminished instrumentation dexterity [19]. The Zeus™ system (formerly Computer Motion™ CA, USA) and the da Vinci™ system (Intuitive Surgical™ CA, USA) allow the surgeon to sit at an ergonomic master console and remotely-operate wristed instruments. A stereo endoscope and the wristed instruments are anthropomorphically-aligned to the surgical scene so that the surgeon is immersed inside the patient, as if they were performing open surgery. Another significant advantage of the master-slave approach is the ability to attenuate the motion of the surgical tool tip [20]. The Robot Assisted Micosurgery (RAMS) system [21][22], the MM-1 [23] and NeuroArm [8] are examples of master-slave platforms

for microsurgical interventions where fine motion scaling of the surgical instrumentation is particularly useful to the operating surgeon during delicate manipulation tasks.



**Figure 2.1 The master-slave da Vinci™ XI system ©2015 Intuitive Surgical, Inc.**

Currently, the surgical robots that have seen the greatest commercial success are tele-operated and cooperatively-controlled robots, the most commonly adopted surgical robot being the master-slave da Vinci™ system (Figure 2.1). More recently, a number of commercial master-slave systems for endovascular intervention have been developed to isolate the surgeon from radiation exposure during fluoroscopic imaging [7]. Additionally, the RIO® MAKOpasty® system (Stryker Corp., MI, USA) is a commercially-available cooperatively-controlled robot for orthopaedic surgery. However, despite a growing interest in robotic surgical systems, the global uptake of surgical robots remains relatively low and is generally constrained to the US and Europe. A key factor is the considerable cost of these systems, which is prohibitive to most healthcare providers. Tele-operated robots are expensive because they require complex master and slave assemblies. Cooperatively-controlled robots also require multi-linkage mechanisms that have sufficient degrees-of-freedom (DoF) to provide the necessary workspace. In addition to the capital investment, the maintenance and consumable costs of medical robots can also be significant. Grounded robots also require time-consuming set up procedures prior to clinical deployment. Once these systems are set up, physical access to the patient is limited and so the robot must typically be used to perform the entire procedure, even if it is only advantageous during key surgical steps. Finally, these robots are still not yet capable of performing interventions without the assistance of theatre staff, as is the case with conventional surgery.

## 2.1 Hand-Held Surgical Mechatronic Devices

Ungrounded, hand-held medical robotic devices are an alternative to robots that are affixed to a grounded reference frame. Hand-held robots or ‘smart’ devices have integrated sensing and on-board actuation so as to provide specific assistive functions to the operating surgeon [15]. These devices have a physically smaller footprint and make use of much of the surgeon’s existing dexterity, making them lower in cost and warranting minimal set up time. Hand-held devices are also unobtrusive to manipulate: there are no linkages that constrain the motion or orientation of the device. This ungrounded architecture allows these devices to be manipulated in exactly the same way as conventional surgical tools. Figure 2.2 provides an illustrative example of the three robot-assistance paradigms outlined: master-slave [21], grounded cooperative-control [18] and hand-held [24]. In all three cases, the clinical application is microsurgery and each system is primarily designed to improve surgical precision using robotic technology. An overview of the hand-held mechatronic devices reviewed in this following section is provided in Table 2.2.

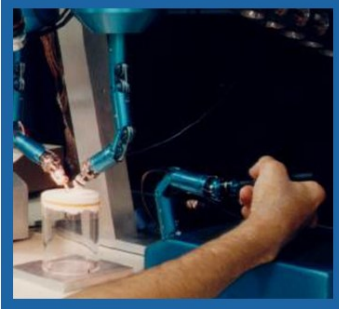
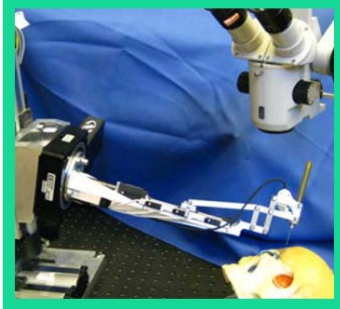
| Master-Slave                                                                                                                                                                                    | Cooperative Control                                                                                                                                                                                          | Hand-Held                                                                                                                                                                                                    |
|-------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------|--------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------|--------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------|
|                                                                                                              |                                                                                                                           |                                                                                                                         |
| <ul style="list-style-type: none"> <li>• Grounded</li> <li>• Tele-operated tool manipulation</li> <li>• High cost</li> <li>• Setup time</li> <li>• Complex, multi-linkage assemblies</li> </ul> | <ul style="list-style-type: none"> <li>• Grounded</li> <li>• Robot and surgeon hold the tool together</li> <li>• Moderate cost</li> <li>• Setup time</li> <li>• Complex, multi-linkage assemblies</li> </ul> | <ul style="list-style-type: none"> <li>• Ungrounded</li> <li>• Mechatronics integrated in to tool</li> <li>• Low cost</li> <li>• Easy integration in to surgical workflow</li> <li>• Less complex</li> </ul> |

**Figure 2.2: Exemplars of different precision-enhancing robotic architectures for microsurgery: the master-slave RAMS system [21], courtesy NASA/JPL-Caltech (left), the cooperative-conyol Steady-Hand system (centre) ©2010 IEEE [18] and the hand-held approach using Micron (right) ©2014 IEEE [24].**

## 2.2 Hand-Held Tremor Suppression Devices

A significant challenge that is inherent to the adoption of ungrounded hand-held devices is the physiological tremor of the operating surgeon. This is particularly important in microsurgical applications that require precise motion at similar magnitudes to human tremor, which can be as low as a few hundred microns [25]. The first implementation of a hand-held robotic tremor compensation was Micron [26][27], which was developed at Carnegie Mellon University by Riviere *et al.* Micron and other such devices work by 1) sensing their own motion, 2) filtering out the components that are considered erroneous motion and 3) actuating their surgical end-effectors in the opposite direction to the filtered erroneous motion. These devices have predominantly been applied to retinal microsurgery applications, [28][29] but they are also applicable to otolaryngology [30], cell manipulation [31] and Optical Coherence Tomography (OCT) scanning [32]. Hand-held tremor suppression devices are advantageous over grounded microsurgical systems in terms of safety, compactness and ease of use. The limited workspace of the instrument mitigates the effects of a catastrophic system failure and the compensation can be instantaneously switched off to render the device passive. These devices are also easier to remove from the surgical site in the event of a failure, unlike their grounded counterparts that are rigidly clamped into position. The introduction of sensing, actuation elements and electrical hardware into surgical devices increases their weight and inertia and this has the potential to induce more tremor in the operator rather than suppress it [33]. The sensing, control and actuation systems also have to be capable of operating over a wide bandwidth because the erroneous motion that must be corrected is between 8-20Hz [34]. Whilst these issues are particularly acute for tremor suppression devices intended for microsurgery, these challenges of bandwidth and weight minimisation are relevant to hand-held mechatronic systems in a global sense. Some examples of tremor suppression devices are presented in Figure 2.3.

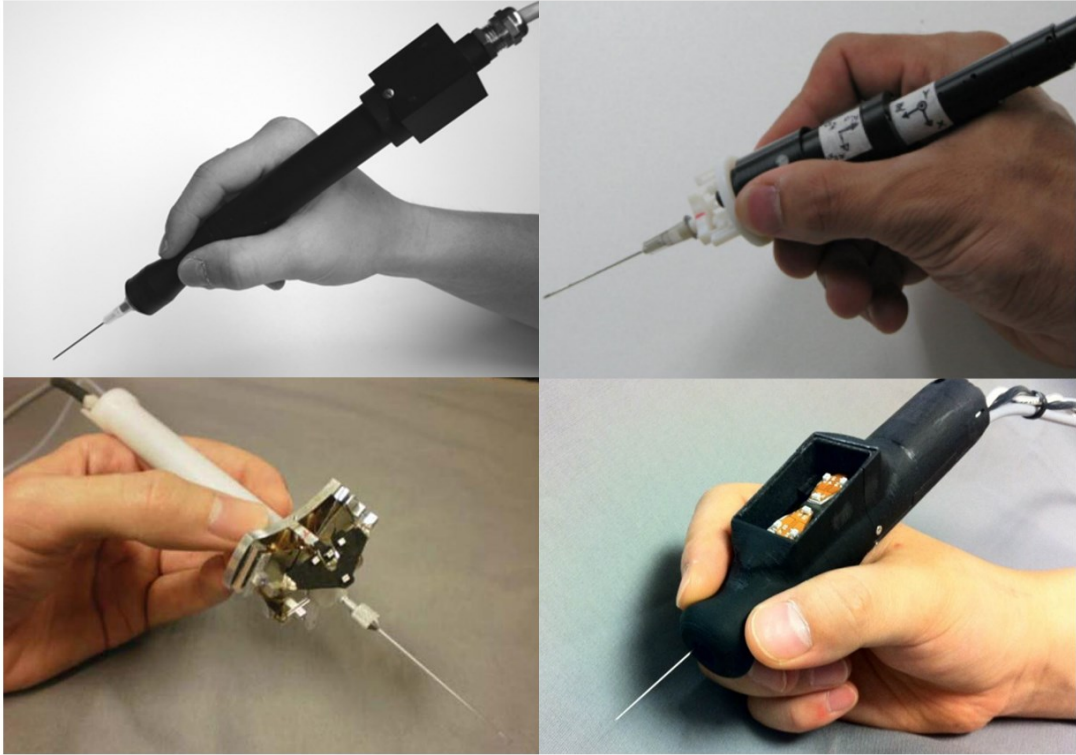


**Figure 2.3: The evolution of tremor suppression hand-held devices: Ang *et al* ©2004 [27] IEEE (top left); Latt *et al* ©2009 IEEE [35] (top right); Cuevas Tabares *et al* ©2010 IEEE [31] (bottom left); Yang *et al* ©2012 IEEE [32] (bottom right).**

Accurate motion sensing is an essential requirement of these tremor suppression systems and a number of technologies have been explored. Inertial sensors represent one such motion-sensing technique that can be fully integrated into the hand-held device itself. Researchers have optimized the placement of the inertial sensors in the pursuit of device miniaturization [35], and furthermore, to maximise sensing resolution [36]. Techniques have been developed to estimate position with inertial sensors [37] but high speed optical tracking techniques [38] provide a superior sensory input as they can accurately localize the absolute 3D position of the tool tip and retain accuracy at lower frequencies [39]. Visual-servoing techniques based on the microscope view have also been integrated into tremor suppression systems [40]. This has allowed semi-autonomous functions to be implemented such as motion scaling of the tool tip and ‘snap-to’ functions that have been used to perform photocoagulation [28].

The actuation design of hand-held tremor compensation devices is another challenge. The key actuator requirements are to have a compact size and low weight, high bandwidth and a position range that must at least match the magnitude of typical human tremor. Actuation mechanism designs are generally compliant-based so as to avoid phenomena such as friction and backlash that are inherent in conventional mechanism designs. Common choices of actuators include piezoelectric elements such as cantilevered bimorphs [39] or stacks that are coupled to amplifying mechanisms [41][42] that allow a slimmer instrument profile. Researchers have investigated other technologies such as ionic polymer metallic composites [43] and voice coil actuators (VCAs), [44] these implementations are summarised in Table 2.2. The latest implementation of Micron uses miniature ultrasonic motors in a 6 DoF Stewart-Gough platform configuration [45][24]. The design allows for an increased workspace and control of the tool orientation for ophthalmic applications in which there is an incision point [24]. Researchers have also integrated intelligent, actuated effectors in to hand-held tremor suppression devices to enhance their functionality [46][47].

### **2.3 Hand-Held Active Guidance Systems**

Image guidance is used routinely in neurosurgical and orthopaedic applications. Hand-held surgical tools are interfaced with preoperative patient data from medical image systems to aid navigation during an intervention. More recently, the robotics community has investigated the incorporation of robotic technology in to such devices so as to provide *active* guidance. Active guidance systems are designed to constrain a surgical tool position towards or away from a pre-defined point, trajectory, or surface. This can allow improved precision of a surgical effector, or improve safety by avoiding anatomically-critical regions. Active guidance has been implemented on Micron using computer vision methods [48]. In this implementation, the surgical tool is constrained to a pre-defined trajectory during manual manipulation. These constraints can be “hard” so that the tool is fully compensated or “soft” so that the tool tip motion is scaled by a pre-set factor. The introduction of such constraints has been shown to improve tool trajectory accuracy in user studies[49]. OCT techniques have also been used to sense the proximity of a surgical tool to an anatomical surface, this can allow physiological tremor to be actively compensated [50].

A similar method of active guidance has also been applied to bone-sculpting applications. The system described by Brisson *et al* [51] and the commercially-available Navio PFS™

(BlueBelt Technologies, Inc, MN, USA) are intended for partial knee replacement procedures (shown in Figure 2.4). These devices are interfaced with image guidance and the rotating burr is retracted when the constraint boundary is approached so as to effectively allow depth control of the cutting tool. A similar approach has also been applied to cranial drilling using pre-operative 3D data to account for local variations in the skull thickness and actively compensate the tool cutting depth [52]. Another hand-held device, the Craniostar, is a motion-controlled mobile robot that is designed to perform accurate craniotomies [53]. This device incorporates a dual wheel drive and an integrated drill; it is held in contact with the cranium by the operator and is steered over the cranial surface. The robot is integrated with a pre-operative image guidance system and its serrated wheels are independently-actuated to allow cooperatively-controlled locomotion about the pre-determined craniotomy trajectory (Figure 2.5).



**Figure 2.4 The Navio PFS<sup>TM</sup> which uses active guidance of the tool tip for resurfacing in partial knee replacement surgery courtesy of BlueBelt Technologies Inc.**



**Figure 2.5** The Craniostar hand-held device (©2009 reproduced from Kane *et al* with permission from Springer) [53].

## **2.4 Hand-Held Devices with Mechatronic Articulation**

The da Vinci™ robotic platform (Intuitive Surgical, CA, USA) is designed to alleviate the problems of MIS, principally by providing wristed instrumentation that returns dexterity to the operating surgeon and an accompanying ergonomic human-robot interface. Surgical device manufacturers have also designed wristed laparoscopic instrumentation to provide additional dexterity over conventional laparoscopic tools. The Radius Surgical System (Tübingen Scientific, Tübingen, Germany), the Roticulator™ and SILS™ instruments (Covidien, MA, USA), the RealHand (Novare Surgical Systems, CA, USA) and the Laparo-Angle™ (Cambridge Endoscopic Instruments, Inc, MA, USA) are all hand-held laparoscopic devices with articulated surgical effectors. These instruments restore dexterity to the surgeon but cannot offer the intuitive user interface that the da Vinci™ provides, although they are considerably lower cost and can be better integrated into surgical procedures. This gap between costly tele-operated robotics systems and the shortcomings of manually-operated

articulated laparoscopic instrumentation is being addressed through the integration of mechatronics into these articulated surgical instruments. The Kymerax™ system (Terumo Medical Corporation, Tokyo, Japan) is a commercially-available articulated hand-held robotic device (Figure 2.6). It uses motor actuation to articulate the tip of a laparoscopic tool with roll, yaw and triggering functions. Motorisation of these functions allows for more intuitive mapping between the user interface and the wristed end-effector. It also allows velocity control of the wristed instruments for precise servoing and the surgical tools can hold their position without requiring a locking mechanism. The Kymerax™ system has thus far been used clinically to perform a laparoscopic hysterectomy [54]. Laparoscopic devices with intelligent, robotic articulation can also be less obtrusive to manipulate, making the surgery more ergonomic to perform. One such articulated device utilizes a pinch-grip interface and allowed suturing tasks to be performed from angles that would not be allowed by conventional instrumentation [55]. Researchers have also investigated user input control strategies with respect to different kinematic configurations of the surgical end-effector in order to facilitate intuitive and precise control of dexterous instrumentation [56]. Another design uses a spherical joint at the instrument handle in conjunction with an actuated trocar port to allow unobtrusive manipulation of the hand-held device [57]. Hand-held articulated robots can also use remote actuation assemblies in which mechanical power is delivered to the device through sheathed tendons [58]. The main advantage to this approach is the considerable weight reduction that is achieved by avoiding direct on-board actuation; however such an approach can also suffer from hysteresis problems [59].



**Figure 2.6 Kymerax™ system (©2011 reproduced from Hackethal *et al* with permission from Springer) [54].**

In addition to wristed designs, articulated hand-held robotic devices can incorporate greater dexterity. The i-Snake<sup>®</sup> is an articulated surgical device with seven actuated DoFs that can be operated in a hand-held configuration and is controlled using a thumbstick (Figure 2.7) [60]. The device uses embedded micromotor actuation that allows for a compact, lightweight design. It incorporates an on-board camera, light source and instrumentation channel. Another articulated, hand-held device used motors mounted in its proximal section to transmit power using linkage mechanisms that provide dexterity of the end-effector [61]. Another hand-held device adopted actuated, steerable needle system using concentric cannulas was developed to facilitate percutaneous intervention [62]. Articulated hand-held devices can also be integrated with image-guidance to allow for localisation of the surgical end-effector. This has been applied to arthroscopy [63] and the concept can also be used to prevent tool collisions during hand-held manipulation [64].



**Figure 2.7 The i-Snake<sup>®</sup> surgical robot in a hand-held embodiment.**

## **2.5 Hand-Held Force Control Systems**

Early medical robots evolved from autonomous industrial robots were designed for use in static, non-deformable environments, hence their early adoption in orthopaedic applications and stereotactic neurosurgery. In soft tissue surgery, the environment is dynamic and deformable, which the surgeon must account for in order to prevent unsafe tool-tissue interaction. This problem is particularly acute for beating heart surgery in which robust robotic control strategies must be adopted to cope with the quasi-periodic motion profiles and the high bandwidth requirement. Yuen *et al* developed a mechatronic force control

system to allow motion tracking of a beating heart, which is otherwise a very challenging task for cardiac surgeons [65][66]. This device uses a combination of distal-tip force sensing and ultrasound-based motion sensing as inputs to a force controller that uses an actuator to maintain a prescribed force setpoint for the surgical tool. Since physiological cardiac motion is typically unidirectional for many surgical applications [67], the use of a 1 DoF force controller is sufficient and particular suited to a hand-held embodiment since the device can be lightweight. Researchers have also investigated various model-based control strategies for hand-held force control devices designed for beating heart surgery. These include a locally-weighted projection regression control strategy [68] and impedance control approach [69]. The introduction of embedded actuation into hand-held force control designs gives rise to significant inertial loads that must be countered by the surgeon, which contaminates haptic feedback. This issue has been addressed through the use of a counter-weight that translates in the opposite direction to the motion compensating actuation mechanism, so as to cancel out the inertial load felt by the surgeon [70].

Researchers have also devised smart hand-held drilling tools that examine the force and torque transients during drilling interventions in order to discriminate the state of tissue [71] (Figure 2.9). These devices use intelligent algorithms that perform automatic control [72], they are nonetheless still under the direct control of the surgeon's hand. These intelligent control schemes can be configured to prevent dangerous penetration of bone during delicate interventions such as cochleostomy, for which the device has been trialled clinically [73][74].

Hand-held force-control robots are expected to play a key role for emerging surgical imaging techniques. High magnification optical probes such as probe-based Confocal Laser Endomicroscopy (pCLE) provide *in vivo* cellular-scale imaging. These “optical biopsies” allow for an *in situ*, real-time diagnosis and can perform mosaicking functions to generate larger maps of tissue morphology. However, since these probes operate at high magnification, motion of the patient or the probe results in inconsistent imaging. Latt *et al* developed a hand-held device incorporating force control for the purpose of maintaining steady contact of a pCLE probe against tissue (Figure 2.8) [75]. This device was demonstrated to maintain consistent imaging of *ex vivo* tissue undergoing simulated

respiratory motion. A modified version of this device was developed for *in vivo* deployment in which it was used to image porcine rectum for transanal endoscopic microsurgery [76]. As well as keeping the probe in contact with moving tissue, the force control scheme ensures that the probe is maintained at a constant force. This is important because different force levels can affect the cell morphology by compressing the tissue [77]. A similar problem occurs in ultrasound imaging, the inconsistent application of force compresses the patient's anatomy that has the potential to alter the diagnosis. Gilbertson *et al* developed a hand-held force control system for maintaining consistent contact of an ultrasound imaging probe [78]. This work also implemented actuator travel limits to prevent saturation of the actuation system, which is an inherent issue to hand-held devices containing embedded actuation. Another feature of this work is the integration of a tri-axial accelerometer to account for the device's orientation and provide additional gravity compensation into the force control algorithm. More recently, force-control mechatronic sub-systems have been integrated in to tremor suppression devices to allow retinal peeling to be performed with less force exertion [79].

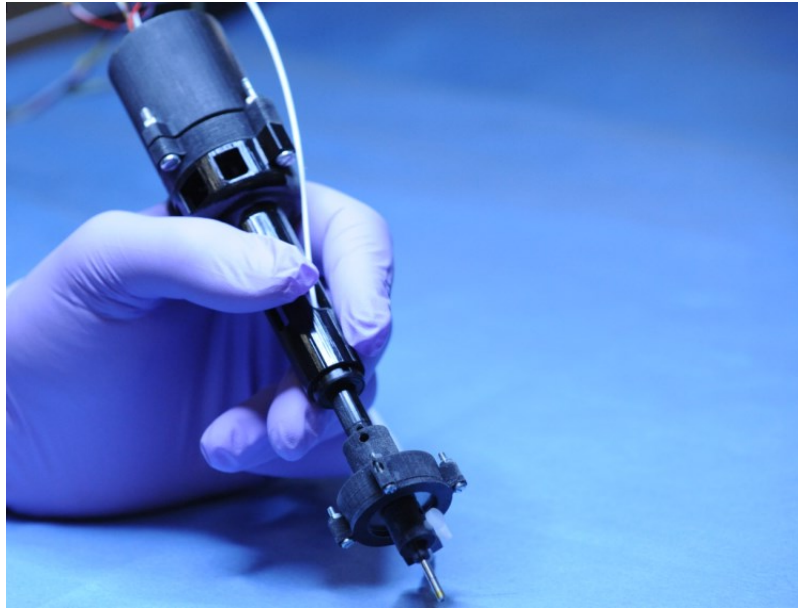


**Figure 2.8** A hand-held mechatronic device developed within the Hamlyn Centre for maintaining steady probe-tissue contact for use with pCLE.



**Figure 2.9** Hand-held sensory guided surgical micro-drill Brett *et al* ©2012 IEEE [71].

**Table 2.2: Four classes of mechatronic hand-held medical devices, features, target clinical applications and exemplars.**

| Class of Device         | Key Features                                                                                                                                                                                                                                                       | Target Clinical Applications                                                                                                              | Example Devices/References                                                                                  |
|-------------------------|--------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------|-------------------------------------------------------------------------------------------------------------------------------------------|-------------------------------------------------------------------------------------------------------------|
| Tremor Suppression      | <ul style="list-style-type: none"> <li>• Active tool stabilisation</li> <li>• High bandwidth motion sensing, filtering and actuation</li> <li>• Multitude of sensing inputs including inertial sensing, optical tracking and computer vision techniques</li> </ul> | <ul style="list-style-type: none"> <li>• Retinal microsurgery</li> <li>• ENT microsurgery</li> <li>• Cell manipulation</li> </ul>         | Micron [39]<br>iTrem [35]                                                                                   |
| Active Guidance Systems | <ul style="list-style-type: none"> <li>• Active tool guidance to constrain tool motion</li> <li>• Can be integrated with image-guidance</li> </ul>                                                                                                                 | <ul style="list-style-type: none"> <li>• Orthopaedics</li> <li>• Microsurgery</li> <li>• Neurosurgery</li> </ul>                          | Navio PFS™<br>SMART forceps [50]<br>Craniostar [53]                                                         |
| Articulated Devices     | <ul style="list-style-type: none"> <li>• Articulated surgical end-effectors</li> <li>• Improved ergonomics over unactuated hand-held tools</li> <li>• Incorporate embedded actuation and intelligent mapping between user interface and end-effector</li> </ul>    | <ul style="list-style-type: none"> <li>• Minimally Invasive Surgery</li> <li>• Single Incision Surgery</li> <li>• Orthopaedics</li> </ul> | Kymerax™<br>Dario <i>et al</i> [63]<br>i-Snake® [60]                                                        |
| Force Control           | <ul style="list-style-type: none"> <li>• Active tool-tissue contact stabilisation</li> <li>• Embedded force sensing in to end-effector or imaging probe</li> <li>• Incorporate intelligent force control schemes</li> </ul>                                        | <ul style="list-style-type: none"> <li>• Cardiothoracic Surgery</li> <li>• Otorhinolaryngology</li> <li>• Medical Imaging</li> </ul>      | Yuen <i>et al</i> [66]<br>Brett <i>et al</i> [71]<br>Latt <i>et al</i> [75]<br>Gilbertson <i>et al</i> [78] |

## 2.6 Human Haptic Perception

Human haptic perception refers to the human sense of touch; it is defined by Gibson as: “*The sensibility of the individual to the world adjacent to his body by use of his body*” [80]. In physiological terms, it is our somatosensory system that is responsible for enabling our haptic perception. Haptic perception is derived from two separate sensory subsystems: cutaneous and kinaesthetic [81]. These sensory sub-systems rely on receptors: physiological structures that generate nervous impulses upon receiving stimuli from the surrounding environment. The main receptors responsible for providing our haptic perception are mechanoreceptors that respond to mechanical stresses and thermoreceptors that respond to temperature. Mechanoreceptors and thermoreceptors embedded within the skin form the cutaneous sensing subsystem whereas mechanoreceptors embedded within the muscles, tendons and hands form kinaesthetic inputs [81]. There are four types of mechanoreceptors, each with a different location, frequency range, receptive field size and rate of adaptation: the temporal response to stimuli from the environment. These are overviewed in Table 2.3 which has been adapted from [82].

A significant body of research has contributed to our fundamental understanding of haptic perception from a behavioural perspective rather than a neuroscience perspective. This approach has sought to address the phenomenology and functionality of haptic perception [81]. An early insight in to the synergistic relation between cutaneous and kinaesthetic sensation was made by Gibson who made explicit the distinction between *active touch* and *passive touch*, or as he put it, the difference between *touching* and *being touched* [83]. Early psychophysical haptic perception experiments had only considered passive touch. These were scenarios in which stimuli was imposed on to the human subject in order to observe a response. Since these experiments failed to elicit the kinaesthetic component of haptic perception, the significance of cutaneous sensation in *active touch* was not previously understood [81].

**Table 2.3: Human mechanoreceptors: maximum sensitivity and function (adapted from [82]).**

| Mechanoreceptor                | Maximum Feature Sensitivity                                                | Primary Functions                                                                                    |
|--------------------------------|----------------------------------------------------------------------------|------------------------------------------------------------------------------------------------------|
| Slowly Adapting Type I (SAI)   | Sustained pressure, maximally sensitive to low frequencies                 | Low frequency vibration detection, coarse texture perception                                         |
| Slowly Adapting Type II (SAII) | Sustained downward pressure, lateral skin stretch, low dynamic sensitivity | Direction of object motion and force due to skin stretch                                             |
| Fast Adapting Type I (FAI)     | Temporal changes in skin deformation (5-40Hz)                              | Low frequency vibration detection, stable precision grasp and manipulation                           |
| Fast Adapting Type I (FAII)    | Temporal changes in skin deformation (40-400Hz)                            | High frequency vibration detection, fine texture perception, stable precision grasp and manipulation |

## 2.7 Human Haptic Acuity

Knowledge of human haptic acuity is an essential pre-requisite in the study of haptic perception and in the design of configurable haptic interfaces. The spatial acuity of human haptic perception has been determined using the two-point-threshold experiment or the point localisation experiment. It has been shown that both methods are highly correlated although the point localisation method consistently yields lower thresholds [84]. Haptic spatial acuity varies across the body but it is highest on the fingertips at approximately 1mm. Furthermore, haptic spatial acuity declines with age, with one study showing a 1% decline per year from the age of 12 to 85 [85]. With respect to temporal acuity, studies have shown that a time period of 5ms can be resolved in between successive tapping of the skin [86]. The spatial resolving ability of the human haptic system is inferior to our vision but superior by comparison to our auditory system. With respect to temporal acuity, the converse is true: our auditory sensory system is superior to our haptic perception and our vision inferior [81].

## 2.8 Weber's Law

Having defined the spatial and temporal resolution of the human haptic perception, it is also important to consider the resolution of human perception in being able to differentiate between stimuli of differing magnitude. This problem was famously addressed by Weber in the context of human weight perception: what is the threshold at which we can discriminate one mass as being heavier than another? The answer is described by Weber's law which states that the Just Noticeable Difference (JND) between two physical stimuli is proportional to the magnitude of the stimuli [87]. In subsequent decades, psychophysical scientists have successfully demonstrated that Weber's law is generalisable to all different kinds of physical stimuli beyond weight.

## 2.9 Haptics in Surgery

Our haptic perception forms an inherent component of how we interact with the world and similarly, haptic perception undoubtedly plays a role in all forms of surgery. Nonetheless, a surgeon's haptic perception can be distorted in a multitude of ways [88]. All surgeons must wear polymer gloves that undoubtedly hinder their cutaneous sensory system. In some forms of open surgery, the surgeon can interact with the patient's anatomy directly which allows them to directly palpate tissue and feel for hidden pathologies or important anatomical structures using their fingertips. However, the majority of the time the surgeon will operate with instrumentation and therefore do not manipulate the patient's anatomy directly. By operating with a surgical instrument, the haptic feedback experienced by the surgeon will differ from direct palpation. To take an illustrative example, if a surgeon handles tissue with grasping forceps, only a global grasping force can be transmitted through the shaft and linkage mechanism. The surgeon's kinaesthetic sense can perceive this force but the surgeon's cutaneous sense is only engaged with the handle of the endoscopic grasper so that no shape or texture information of the grasper-tissue interaction can be perceived. Haptic perception can also be distorted by additional forces that are present but not related to the important tool-tissue interactions. A good example of this problem can be found in MIS. By operating through an incision point, lateral forces that are applied to the endoscopic tools are supported by the incision point which acts as a pivot and transforms lateral forces into torques. Consequently, forces applied to the tip of the endoscopic instrumentation can either be amplified or attenuated by this lever effect. Furthermore, the endoscopic instrumentation is inserted through 'trocar' ports that are designed to form a seal around the instrument shaft.

This sealing effect generates frictional forces as the surgeon translates and rotates the instrumentation which further contaminates haptic perception of the tool tip-tissue interactions. Bholat *et al* compared direct palpation against manipulation with conventional instruments and laparoscopic instruments in correctly discriminating between primitive shapes, textures and stiffness [89]. The authors confirmed that haptic feedback is present during MIS, but that it is nonetheless diminished compared to direct palpation and manipulation with conventional instrument. The study also demonstrated that haptic perception is inferior for inexperienced surgeons, suggesting that more experienced surgeons learn to interpret haptic cues.

The significance of contaminating forces depends on the levels of force that are required during surgery and can be predicted by Weber's law. The greater the magnitude of the contaminating forces, the higher the JND force-threshold becomes and the lower the surgeon's haptic acuity. It follows then, that the surgical context becomes significant to the magnitude of the contaminating force. For example, the forces exerted by surgeons performing microsurgery are generally orders of magnitude lower than those of surgeons performing MIS. In very delicate surgery, such as retinal or microvascular surgery, even the static weight of instrument will be felt by the surgeon and becomes a haptic perception contaminant. Body forces due to the instrument mass and inertial properties will be perceived by the surgeon as the instrument is dynamically manipulated, both intentionally and as a result of physiological tremor. Such contaminating forces may be highly significant in hindering haptic perception during microsurgery but of no consequence in MIS where forces are generally higher.

Gupta *et al* conducted a study to quantify the forces exerted during retinal microsurgery and compared the magnitude of these forces to that which is discernible to retinal micro surgeons [90]. The study was split in to two phases. In the first phase, the subjects performed a retinal surgical procedure on porcine cadaver eyes with a calibrated force-sensing microsurgical pick and the forces exerted were recorded. In the second phase, subjects were blinded and required to hold the same microsurgical pick whilst micro-scale forces were applied to its tip at different magnitudes. Subjects were required to indicate when they could perceive tactile "events" occurring as a result of the mechanical stimuli to the surgical pick. It was shown

that the majority of retinal microsurgery takes place below 7.5mN and that most subjects could not detect the majority of forces below this threshold in the experimental setup described. One conclusion of this work is that the majority of retinal microsurgery is likely to be conducted without the surgeon having perception of the forces exerted by the surgical tool tip on to the tissue. Another study by Harada *et al* examined forceps grasping force in microsurgery [91]. The study found that experienced surgeons applied less force compared to inexperienced surgeons.

### **2.9.1 Haptic Feedback in Master-Slave Surgical Systems**

If haptic perception is “distorted” in these conventional surgical contexts, when a robot is used to operate, intrinsic haptic perception is lost altogether. This is the case for many master-slave robotic systems, including the da Vinci™ system because the surgeon does not directly handle the interventional instruments and the robot has no capacity to reflect the forces exerted back to the surgeon. This deficiency has been a long-standing criticism of the da Vinci™ and other similar master-slave systems. Mohr *et al* studied 148 cardiac surgical interventions that were performed with the da Vinci™ and hypothesised that the absence of tactile feedback may be problematic in allowing surgeons choose an optimum region to perform anastomoses [92]. A master-slave system can provide haptic feedback if the slave system can artificially sense the tool-tissue forces and the master console can reflect these back to the operator through a haptic interface. A few examples have seen clinical deployment, including the NeuroArm (University of Calgary, Canada) and Sensei® X system (Hansen Medical, CA, USA). Many other such systems have been developed within the research community. This has consequently allowed researchers to conduct user performance studies that compare full haptic feedback with no haptic feedback. These studies have provided important insights in to the intrinsic role of haptic feedback across different surgical scenarios.

### **2.9.2 Master-Slave Haptic Feedback and Visual Cues**

When operating master-slave surgical systems that have no capacity for haptic feedback, the surgeon must rely on visual cues, such as tissue deformation in order to infer how much force is being exerted by the surgical instrumentation. Within the haptic perception community, the relationship between vision and haptic perception has been studied in fundamental terms. Klatzky *et al* demonstrated that our haptic perception is more focussed on the material properties of an object, whereas determining the shape of an object is

dominated by our visual perception [93]. Ernst and Banks proposed a methodology for determining the extent to which visual perception and haptic perception dominates when the visual and haptic perceptual systems are combined [94]. They hypothesise that the central nervous system determines dominance through a Maximum Likelihood Estimation (MLE) function. Within a surgical context, studies have demonstrated that visual feedback of tissue deformation can help surgeons infer force application and improve performance [95][96].

The development of force-reflecting master-slave surgical systems has allowed researchers to study the significance of visual cues in inferring haptic perception. Tholey *et al* designed an experiment to explore the relative roles of force-feedback and visual feedback in tissue characterisation using a laparoscopic grasper [97]. They demonstrated in a user study that when both force-feedback and visual feedback are combined, the surgeon's ability to characterise tissue stiffness is superior compare to with only visual feedback or only force-feedback. The authors postulated a second hypothesis that with only force-feedback, surgeons would be able to better characterise tissue stiffness compared to with only vision feedback. This hypothesis was not confirmed, illustrating the significance of visual feedback in haptic perception. Wagner *et al* studied the role of force-feedback during a blunt dissection task [98]. They devised a user study that used a telerobotic system to expose an artery using a video endoscope for visual feedback. They compared cases in which there was no force-feedback and force-feedback with gains of 75% and 150%. Their results demonstrate that users exerted significantly more force when they had no force-feedback but that the rate and precision of dissection were unaffected. In another study, Wagner *et al* examined the impact of haptic feedback in a MIS task with limited visual feedback [99]. The telemanipulation task was designed to simulate an anchor deployment with ultrasound guidance. The goal was to apply a pre-determined optimum force level during the task which considered two tissue samples of differing stiffness. Three cases were considered, pure visual (ultrasound) feedback, visual feedback with force-feedback and visual feedback with vibration-feedback at a pre-defined force level. They confirmed that when manipulating lower stiffness tissue, visual feedback is more effective as the visual tissue deformation is better perceived. They also concluded that the addition of force-feedback does improve user performance as much as providing an indication of a target force level.

### 2.9.3 Sensory Substitution in Master-Slave Surgical Systems

Haptic feedback may also be appropriated through sensory substitution. Researchers have devised means by which direct haptic perception of force can be replaced by displaying force information through another sensory input channel, for example with visual or auditory cues. Jones performed psychophysical studies that showed the effectiveness of human subjects in using visual feedback to exert a constant force [100]. Studies confirmed that users could control both index finger and elbow flexion with greater precision with visual guidance cues, in comparison to pure haptic perception, for a range of forces.

In a clinical context, sensory substitution can provide a means to indicate how much force is being exerted by the slave robot so as to aid the surgeon in limiting the amount of force applied. The Sensei<sup>®</sup> X system is an example of a robotic master-slave system that adopts this principle. A force sensor measures forces experienced by the catheter tip and these can be displayed to the interventional radiologist visual or as vibrotactile feedback. Within the research domain, a study was conducted to evaluate a variety of sensory substitution methods in a suture tensioning task during robotic telemanipulation [101]. Surgeon subjects were required to perform a standard surgical knot tying tasking with a da Vinci<sup>™</sup> system for four conditions, no feedback, visual feedback, auditory feedback and a combination of visual and auditory feedback. The feedback modalities indicated to the surgeon when the ideal suture tension had been reached and the addition of such feedback showed significant improvements in force approximation and consistency. Another study demonstrated that visual feedback of suture tension force demonstrated the greatest improvements for inexperienced surgeons [102]. A further implementation of sensory substitution that has also been implemented on the da Vinci<sup>™</sup> platform, is the Verrotouch system by Kuchenbecker *et al* [103] (Figure 2.1). This approach senses the accelerations that occurs from tool-tissue contact and augments the signal through vibration actuators. The developers of the system have validated the approach *in vivo* and demonstrated a significant proportion of surgical events can be detected by surgeons using this approach [104].

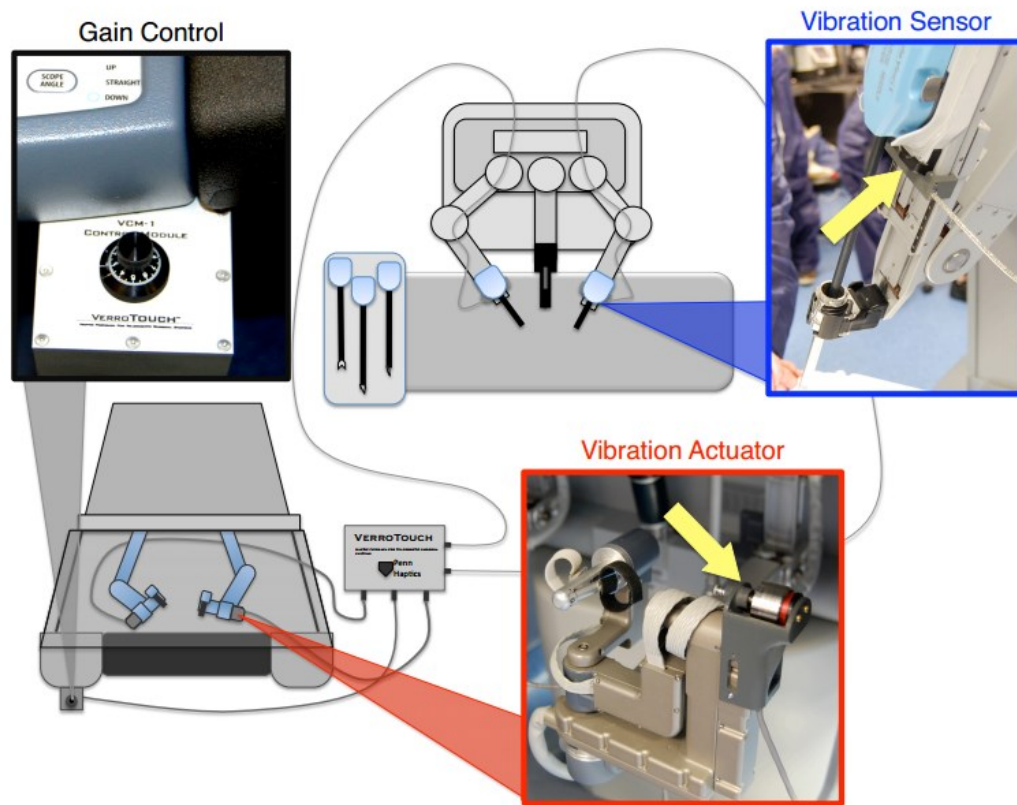


**Figure 2.10** The Verrotouch system that provides vibration-based haptic feedback to the da Vinci™ robot (©2010 reproduced from Kuchenbecker *et al.* with permission from Springer) [103].

#### 2.9.4 Haptic Interaction from Active Constraints

The surgical haptic feedback systems reviewed thus far have all focussed on master-slave architectures. These use real-time sensing of the surgical environment to facilitate haptic feedback to the surgeon through the master control interface. Another means by which haptic feedback can be provided is through “active constraints” [105], also known as “virtual fixtures” [106]. Active constraints are pre-operatively prescribed virtual boundaries that constrain a robot effector through haptic feedback. Within the field of surgical robotics, the principle is known for its role in grounded cooperatively-controlled robotic systems including the Acrobot™ [107] and the subsequent RIO® MAKOpasty® system (Stryker Corp., MI, USA). These systems uses preoperative geometrical data obtained from medical imaging to define a surface constraint that is registered with the patient during surgery. The robotic manipulator is programmed to move freely outside of this constraint so that the

operating surgeon can manually remove bone material using a rotating burr. If the surgeon approaches the constraint boundary, the robot will increasingly impede the surgeon by generating a resistive force. This resistive force-feedback effectively constrains the surgeon from removing excessive bone material and allows them to sculpt accurate geometries. The active constraint approach is advantageous in that it allows the surgeon to experience both the intrinsic haptic feedback from the procedure and the virtual force-feedback. When outside of the constraint region, the robot moves freely so that the surgeon can feel the drill vibration and modulate the amount of force that they wish to exert with the burr. This approach has also been adopted for microsurgical applications using the Steady-Hand robot [108]. In this work, active constraints are used to assist a surgeon in guiding a tool through complex anatomical geometries.

## **2.10 Engineering Design of Haptic Interfaces**

Thus far, a number of haptic feedback systems have been discussed in the context of user performance and haptic perception in surgery. This section of the review intends to describe the technical means by which artificial haptic feedback is made possible. In the broadest sense, a haptic interface is a form of configurable human-machine interface that actively engages the human haptic perception. Haptic interfaces can be designed to stimulate either the human kinaesthetic or cutaneous sensory subsystems or both. Artificial haptic interfaces have been developed over several decades but significant advancements were made in the 1990s when multi DoF haptic interfaces for the kinaesthetic haptic perception first became commercially-available [109]. These haptic interfaces were designed for the human user to actively explore and touch simulated or remote environments. As a surrogate for touching a real environment, the user will manipulate an interface that can, in turn, exert forces back on to the user so as to simulate contact by eliciting the user's kinaesthetic haptic perception and proprioception. The mechatronic design of such haptic devices is challenging. One of the key challenges that has characterised the development of such interfaces is that of *transparency* [109]. An ideal, fully transparent haptic interface would be able to perfectly simulate real-world haptic interaction without the user being conscious that they are not interacting with the real-world but in fact, a simulated environment. One of the first designs to address the challenges of haptic transparency and allow kinaesthetic-based force-feedback was the PHANToM device, developed by Massie and Salisbury [110]. To put the notion of transparency in to more concrete terms, Massie and Salisbury describe two contrasting conditions that illustrate this point. Firstly, "*free space must feel free*" which means that the

user should not perceive any force when moving in free space. This requirement has implications for the mechatronic design of haptic interfaces: quantities such as inertia and friction should be minimised and there should be no unbalanced weight which would break the illusion that the user is interacting with a virtual environment. Secondly, Massie and Salisbury state that “*solid virtual objects must feel stiff*,” for this to be true, sufficient forces and torques are required of the haptic interface in order to effectively resist the user as they contact a virtual rigid object. These transparency requirements necessitate conflicting engineering challenges. Firstly, in order to generate large forces, large (and consequently heavy) actuators are required which would increase inertia and compromise the first transparency requirement. Conventional robotic manipulators would make use of a gear train in conjunction with a lightweight actuator to resolve this problem, but such an implementation would introduce friction, inertia and backlash which all serve to degrade the users’ sensation of “*free space feeling free*”. The PHANToM design used a number of strategies to optimise for transparency. Pre-tensioned tendon-capstan assemblies provide friction-free, low inertia torque amplification of the motors. This allows the use of relatively low weight motors that can still exert high torques on to the counter-balanced linkage mechanism that the user manipulates. This PHANToM design was commercialised by Sensable Technologies (now Geomagic, MA, USA) and it has been optimised over subsequent decades and adapted in to other designs. Whilst the design has evolved, many of the original design features are utilised in commercially-available haptic interfaces today including the use of pre-tensioned tendon-driven capstan assemblies. The Quanser HD<sup>2</sup> system (Quanser, Ontario, Canada) uses two coupled manipulators of a similar architecture to the PHANToM device in a parallel manipulator configuration. The W6D device (Entact Robotics, Ontario, Canada) also uses a parallel-linkage architecture. These configurations are capable of generating comparatively higher loads and recreating stiffer collisions but at the expense of workspace limitations and singularity problems. The Sigma, Omega and Delta systems (Force Dimension, Nyon, Switzerland) use a delta-based parallel kinematic architecture that can generate high forces without singularity issues. These devices can also incorporate active gravity compensation so as to improve transparency by making the electromechanical assembly feel weightless to the user.

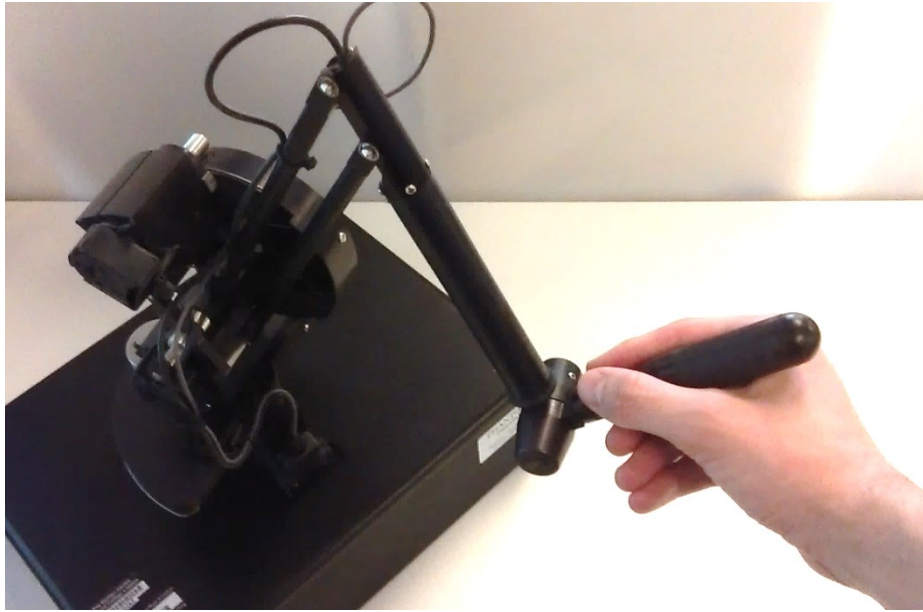


**Figure 2.11** A commercial PHANTOM device, originally developed by Massie and Salisbury [110].



**Figure 2.12** The Omega.3 device that uses a Delta manipulator (image courtesy of Force Dimension, 2015).

Another significant challenge in the design of transparent haptic interfaces is a computational one. Television can accurately simulate real-world human visual experience because of the temporal resolution limitations in the human visual sensory system. The human brain can interpolate between individual frames so that a series of discrete images received by the human visual sensory system at frequencies in the order of 30Hz will appear continuous and indistinguishable from human perception of the real world. Unfortunately for haptic interface designers, the human haptic perceptual system has a more acute temporal resolution which warrants haptic interfaces to have update rates in the region of 1kHz [111]. This has computational implications for the control systems of haptic interfaces which must resolve the forces being exerted on to the user within timescales in the order of magnitude of a millisecond. Whilst modern computers have gained significant computational power since the development of the original kinaesthetic haptic interfaces, computing the forces that a haptic interface must generate is still a significant challenge. It is especially challenging when considering haptic interactions with environments that require non-linear calculations, such as deformable soft tissue surfaces. For perfect transparency in a master-slave configuration, the slave manipulator must be capable of conducting perfect environment force sensing and dynamic modelling with minimal latency in the data transmission. It is for these reasons that force-reflecting master-slave systems can potentially become unstable, a potentially catastrophic event in robot-assisted surgery. This has motivated researchers to consider the sensory substitution techniques reviewed earlier as a haptic rendering method for robotic surgery since the approach is inherently robust to control instability.

## **2.11 Ungrounded Haptic Interfaces**

The haptic interfaces discussed thus far, have all been mechanically-grounded: they are anchored to a grounding frame or attached to a heavy mass that can support the reaction forces generated by the interface itself. As an alternative to this approach, haptic feedback can be implemented in mechanically-ungrounded devices. Incidentally, this type of haptic interface is ubiquitous the world over: vibration-based haptic feedback is commonly integrated in to mobile phones, computing devices and gaming consoles. These devices commonly use electric motors to rotate an off-axis mass that generates a vibration [112]. Researchers have extended this concept to provide tactile feedback in to mobile devices using an array of piezoelectric cantilevers that locally stretch the skin of the user's fingertip [113]. Arasan *et al* developed a haptic stylus with vibrotactile feedback which could create the illusion of a tactile flowing effecting along the axis of the stylus or as a rotation about the

stylus axis [114]. Multiple vibration-based generators are integrated in to the device and a control system modulates when they are activated which alters the user's perception of the vibration source and direction. These devices, whilst limited in their ability to fully render haptic interaction forces, are low cost and have a theoretically unlimited workspace. There are no mechanical linkages that are required by grounded haptic interfaces which constrain both the reachable and dexterous workspace.

A number of researchers have developed ungrounded hand-held haptic interface that use inertial members within the device itself to generate loads that experienced by the user. The TorqueBAR [115] is a two handled bar that is held by the user, the device incorporates a mass that can translate along the bar, so as to alter its own centre of mass in real-time. The quasi-static torque generated in this device is as a result of gravity acting on the mass at a distance from the bar handle which is held by the user. Reaction torques can also be generated from dynamic sources such as rotary motors, whereby angular acceleration and deceleration of a rotating mass transmits reaction torques to the user. The Gyrocube uses three flywheels in orthogonal orientations to permit torque generation in all three axes [116]. An alternative approach is to exploit the gyroscope effect by actively rotating high-speed flywheels in an axis orthogonal to its axis of rotation. The Gyro Moment Display makes uses a flywheel mounted on to an actuated 2 DoF gimbal assembly [117]. The iTorqU device also work on a similar principle [118]. Later developments of this device also resulted in improved transparency, the gimbal assemblies were programmed to actively compensate for the users pose so as to nullify the gyroscopic effect when no torques were intended to be exerted on to the user [119]. Another gyroscope-based haptic interface was specifically designed to be integrated in to a tablet and the researchers discussed how such feedback could simulate virtual changes in weight, viscosity, momentum and assist in path guidance [120]. The approaches discussed thus far have used an actuatable solid mass to generate reaction forces. Researchers have also investigated the use of fluids, and specifically air-jets to generate reaction forces as a result in the change of momentum in the air flow [121]. Magnetic fields too have been used to effectively manipulate ungrounded hand-held devices. The Maglev 200™ (Butterfly Haptics LCC, Pittsburgh, USA) is a magnetic levitation system that provides 6 DoF of haptic feedback to the user.



**Figure 2.13** The iTorqU 2.1 device that uses gyroscopes to provide ungrounded haptic feedback, Winfree *et al* ©2010 IEEE [119].

Another class of mechanically-ungrounded hand-held haptic interface are physically mounted to the user so that they themselves provide a ground through which force can be transmitted. A good example of this concept is the CyberGrasp device (Cyber Glove Systems LCC, San Jose, USA). The CyberGrasp is a mechanically-ungrounded haptic feedback glove that is mounted to the user's hand and provides force-feedback to each individual finger which can allow the user to perform virtual grasping tasks. Studies investigated the use of the CyberGrasp in performing teleoperation tasks [122][123]. The researchers demonstrated that the system could allow users to accurately discriminate between different virtual object sizes but that user stiffness discrimination was more challenging for users. This user-grounded haptic feedback principle has also been applied to stylus-based devices. Researchers devised a 3 DoF force-feedback stylus which is attached to the user's hand [124]. In later work this device was integrated with an optical tracking

system so as to allow virtual haptic interactions with a virtual environment through the stylus [125]. Cutkosky *et al* investigated the problem of using mechanically-ungrounded haptic interfaces in rendering forces beyond those which are internal to the hand [126]. The researchers performed a series of boundary detection experiments in which users had to determine when a virtual boundary had been contacted with both grounded feedback and ungrounded force application to the fingertip. They found that whilst grounded haptic feedback was superior in portraying the physical experience of contacting a rigid surface, ungrounded haptic feedback was still effective in the users' ability to detect the boundary. They also found that visual feedback aided boundary detection.



**Figure 2.14: The Cybergrasp system that provides ungrounded haptic feedback, courtesy Cyber Glove Systems. Image from Aiple *et al* ©2013 IEEE [123].**

## **2.12 Hand-Held Surgical Devices with Haptic Feedback**

In master-slave and cooperatively-controlled systems, haptic sensing is a necessity in order for any haptic feedback to be provided either as a rendered force or through sensory substitution. Hand-held mechatronic devices already allow haptic feedback that is equivalent to existing passive instrumentation since they are directly manipulated by the surgeon. Consequently, the role of haptic feedback in hand-held devices is to enhance the existing

tool-tissue interactions or to provide the surgeon with additional haptic information. One underlying goal is to present the surgeon with the important haptic information required to improve an intervention by diminishing contaminating haptic interactions. For example, in laparoscopic surgery, the goal is to extract the important tool-tissue interactions from the trocar frictional forces. In microsurgery, the goal is for the surgeon to differentiate the delicate tool-tissue interaction forces from the weight and inertial forces felt from manipulating the surgical tool.

### **2.12.1 Haptic Sensing and Feedback in Hand-Held Instruments for MIS**

Bicchi *et al* developed a laparoscopic grasper with force sensors that could measure grasping force, this information could then be visually relayed to the operating surgeon so that they could infer tissue stiffness [127]. Hu *et al* developed a similar device that could accurately characterise soft tissues [128]. Hanna *et al* developed a force-sensing laparoscopic grasper that could provide visual-based sensory-substitution to the surgeon for clinical use [129][130]. Whereas these previously described devices examined sensory-substitution for global force-feedback in laparoscopic instrumentation, Ottermo *et al* investigated tactile sensing in a laparoscopic tool with visual feedback of the integrated tactile array [131]. This sensing methodology was also used in conjunction with a tactile pin display [132] based on a prior concept postulated by Howe *et al* [133]. In this design, the operating surgeon could place their finger over the tactile display on the hand-held device so as to discriminate the shape of the object being grasped by the distal sensor. Yao *et al* developed a tactile magnification system, the MicroTactus, which was developed to aid the detection of lesions in arthroscopic intervention [134]. This device works by sensing the tool-tissue interactions with an accelerometer that is integrated into the proximal region of the arthroscopic hook tool. A magnetic actuator is then used to amplify these signals to provide vibrotactile feedback to the surgeon. This method of vibrotactile feedback is intrinsically fail-safe because a failure in the control system would not cause erroneous motion of the surgical tool. Furthermore, the surgical tool remains unmodified allowing the device to be sterilisable and have low manufacturing costs.

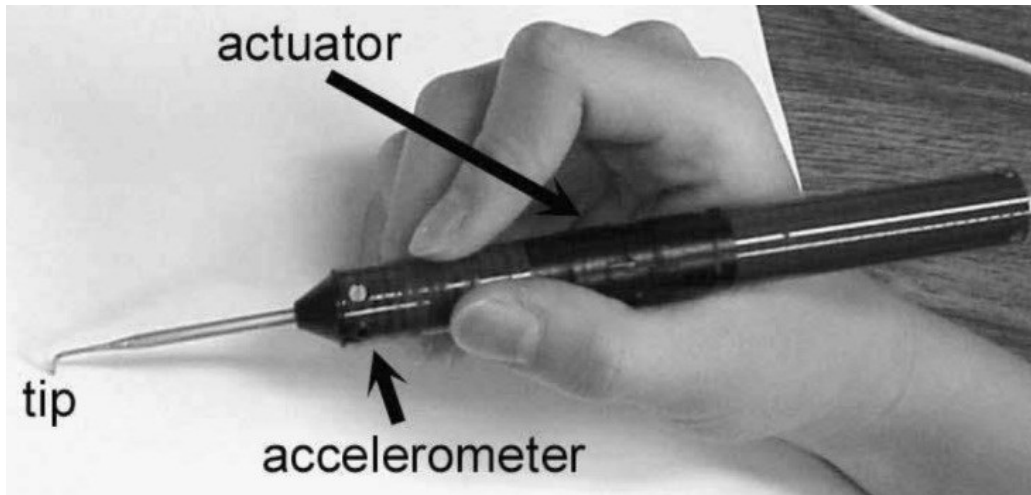


**Figure 2.15: The MicroTactus (©2005 reproduced from Yao *et al* with permission from Springer) [134].**

### **2.12.2 Haptic Sensing and Feedback in Hand-Held Instruments for Microsurgery**

It has been shown that the magnitude of force interactions in microsurgical applications can be extremely low and often beyond human haptic perception. Tool-tissue forces in ophthalmic surgery, for example can often be below 7.5mN, which is below the threshold at which some surgeons can physically perceive force [90]. It is particularly crucial in very delicate surgery, such as microsurgery (i.e. neurosurgery, ophthalmic surgery) that iatrogenic injury does not occur as trauma can be acute and permanent. This has prompted researchers to develop sophisticated sensing systems that can operate at the micro-scale with high sensitivity. Berkelman *et al* developed a triaxial, micro-scale force sensing system for use in microsurgery. The design uses a double four-beam cross configuration which forms the base of a microsurgical instrument shaft [135] [136]. This configuration allows quantification of both the lateral loads applied to the instrument tip as well as loads that exerted along the axis of the surgical tool shaft. In a normal cross-beam configuration there is an inherent anisotropy in sensitivity that is a result of the surgical tool acting as a cantilever which leads to lateral loads generating higher strains compared to axial loads. The use of a double cross-beam configuration addresses this issue by increasing the stiffness of the instrument in response to lateral loads. This force sensor assembly was integrated in to a hand-held design and used to capture force data during *in vitro* surgical task studies by Jagtap and Riviere [137].

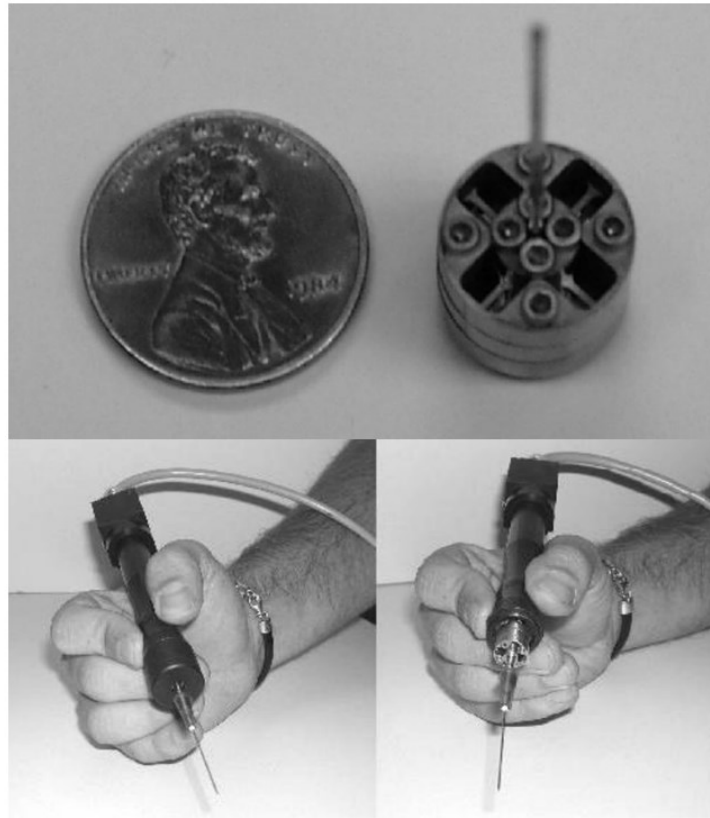


**Figure 2.16: Micro-scale force sensing hand-held instrument using a cross-beam sensing assembly; the original fabrication Berkelman *et al* ©2003 IEEE [136] (top) and integration in to a hand-held device Jagtap and Riviere ©2004 IEEE [137] (bottom).**

One finding from the work of Jagtap and Riviere was that the measured forces were higher compared to other similar studies. This was attributed to the lateral forces experienced as the tool shaft is passed through the incision point in the sclera. This highlights a problem with the proximal-based force-sensor configuration, which is the inability of the sensing system to distinguish between loads that are applied at the instrument tip and at other arbitrary locations along the instrument shaft. To overcome this challenge, researchers have integrated force-sensing assemblies in to the shaft of the microsurgical tool itself using optical fibres [138]. This approach allows for a sub milli-Newton force sensing resolution and for increased sensitivity to loading at the instrument tip, where the important force interactions occur. The design uses three optical fibres which are bonded to a micro-machined shaft at 120° increments so that the entire assembly deflects as one cantilevered beam. The total assembly was 0.5mm in diameter and used 0.16mm diameter fibres. Fibre Bragg gratings (FBG) are machined in to the optical fibres. These gratings have a characteristic wavelength

known as the Bragg wavelength which alters as the fibre deforms due to bending of the tool shaft and temperature changes. The authors reported an algorithm for temperature compensation of the sensing assembly.

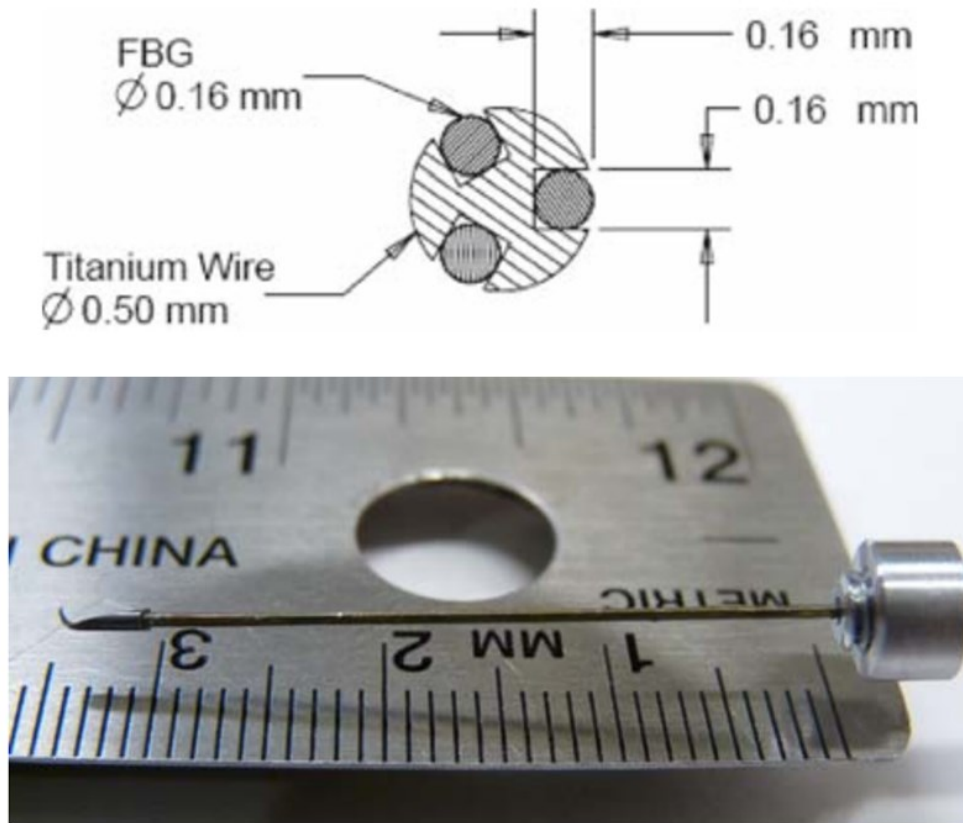


**Figure 2.17:** A FBG sensor assembly for microsurgery which has subsequently been integrated in to hand-held designs; top image shows a cross section and the three sensing fibres. Bottom image shows the integrated assembly. (©2009 reproduced from Iordachita *et al* with permission from Springer) [138].

Whilst the design of such force sensing assemblies is technically challenging, the sensors themselves are immune to electrical noise and have good sterilisation and biocompatibility properties. One interesting implementation of such micro-scale sensing technology is in conjunction with auditory-based sensory-substitution [139]. The micro-scale force sensing system is used to generate auditory cues as to how much force is being exerted during delicate manipulation tasks. The system uses audible beeps to indicate force levels and modulates the beep frequency to infer force magnitude to the operator. User studies showed that this method was effective in reducing force levels during simulated retinal peeling tasks.

FBG-based micro-scale force sensors have been integrated in to Micron and demonstrated to improve user performance when auditory feedback was used to indicate a critical force threshold [140]. FBG-based sensors have been implemented in to microforceps designs that are used to perform retinal peeling. These designs have been applied to the grounded, cooperatively-controlled Steady-Hand robot [141] as well as in ungrounded, hand-held configurations [142].

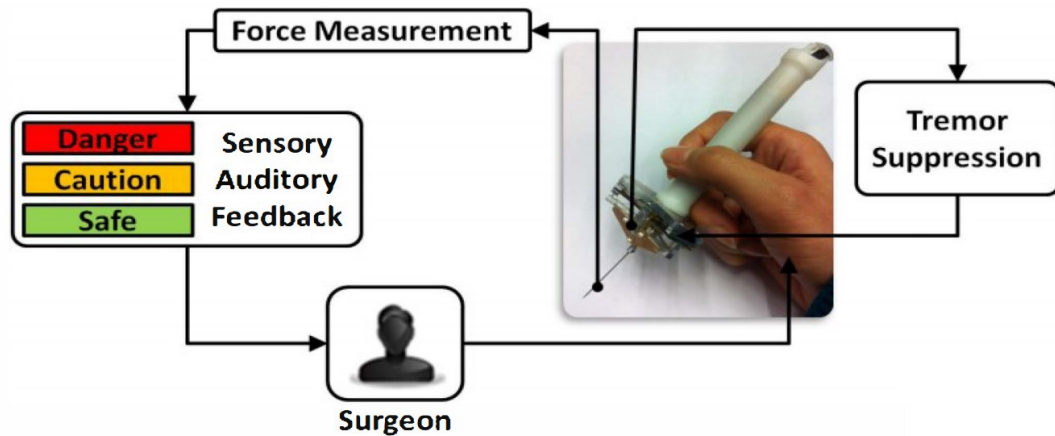


**Figure 2.18:** Integration of FBG force sensing elements in to a tremor suppression device, Gonenc *et al* ©2012 IEEE [140].

3 DoF micro force sensing instrument designs using FBG sensors have been investigated [143]. By machining compliant flexures in to the tool shaft, the magnitude of deformation due to axial loading is increased and thus the sensor sensitivity is improved. This work also addresses the challenges in obtaining a robust calibration for a triaxial sensor assembly.

### 2.12.3 Force Magnifying Hand-Held Tools

Recently, researchers have proposed force-magnifying hand-held devices for improving haptic perception in surgery. Stetten *et al* proposed a force-magnifying hand-held stylus device that does not require affixing to the ground [144]. This device works through sensing the tool-tissue forces and magnifying this force using embedded actuation within the device itself, a schematic of this is presented in Figure 2.19. An anchoring brace attached to the operator's wrist can support the reaction force generated by the hand-held device. The device

has been evolved further to incorporate improved force sensing and bi-directional force-feedback [145].

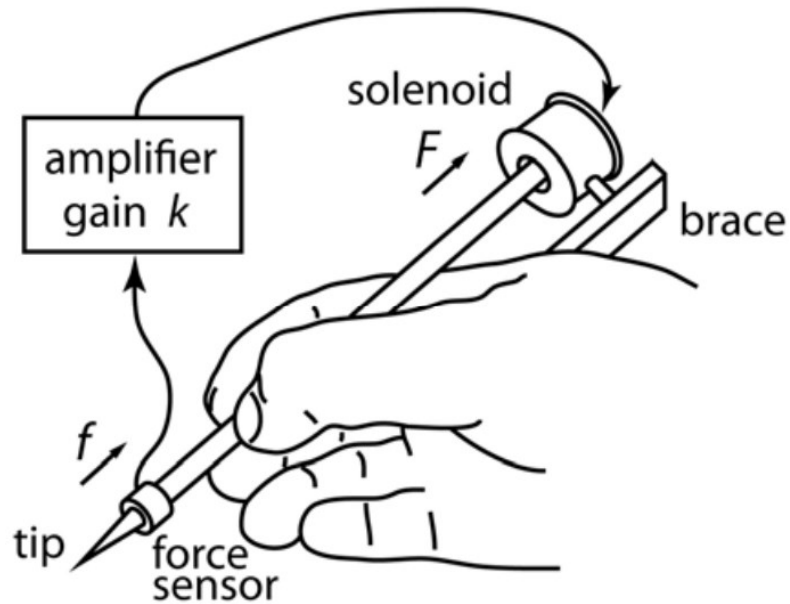


Figure 2.19: The mechatronic force magnifying concept (©2011 reproduced from Stetten *et al* with permission from Springer) [144].



Figure 2.20: The evolved force magnifying instrument originally developed by Stetten *et al* [144] (©2013 reproduced from Lee *et al* with permission from Springer) [145].

## 2.13 Conclusions

This chapter has reviewed the background literature pertaining to the hand-held haptic feedback strategies that will be proposed in this thesis. Firstly, a history of surgical robotics was presented to provide the context to the class of hand-held smart surgical device that is a foundation to this thesis. The state-of-the-art in hand-held mechatronic tools was discussed to provide context and to illustrate the various sensing, actuation and control strategies that researchers have adopted. The next section of the review focussed on human haptic perception and the relevance of haptics to surgical interventions. This was followed with a presentation of recent engineering efforts that have been made towards the design of electromechanical interfaces that can stimulate the human sense of touch and allow simulation of real and virtual environments. The review further focussed on instruments that can perform these functions in a hand-held, ungrounded configuration. Finally, ungrounded haptic augmentation systems specifically designed for surgery were presented.

It is evident from the literature reviewed in this chapter that careful force application is crucial in surgery. In delicate microsurgical procedures, the surgeon must carefully manage the forces exerted during an intervention because excessive force can cause irreversible trauma to the patient. It is vital that the surgeon is well informed with respect to how much force they are exerting. In other instances, it is important that the surgeon avoids critical anatomical regions altogether. Whilst robotic technology can assist a surgeon's haptic perception and prevent undesirable tool-tissue interactions; the cost, size and set up time of large, grounded robotic systems is prohibitive to the widespread clinical uptake of robotic technology. This thesis will explore different opportunities for simple, low cost, hand-held solutions that can provide different forms of haptic augmentation to the operating surgeon.

# Chapter 3: Ungrounded Force Amplification in a Hand-Held Device<sup>†</sup>

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<sup>†</sup>Content from this chapter was published as:

**A New Hand-Held Force-Amplifying Device for Micromanipulation.**

Payne C.J., Tun Latt W. and Yang G-Z. IEEE International Conference on Robotics and Automation, St Paul, USA. 2012. pp1147-1152.

This chapter introduces an ungrounded hand-held surgical device with the ability to amplify force through an electromechanical system integrated within the device itself. Chapter 2 reviewed human haptic perception and surgical mechatronic systems that incorporate haptic feedback. Psychophysical studies have demonstrated the limitations of human tactile sensing in surgery, particularly in delicate microsurgical interventions where tool-tissue forces can be at the micro-Newton scale [90]. Many researchers have thus hypothesised that poor force perception can lead to surgical error. For example, researchers have deduced the force magnitude at which retinal tearing can occur during retinal peeling interventions [18]. A number of haptic augmentation systems that can amplify tool-tissue forces have been proposed to improve force perception and improve safety. The master-slave and cooperatively-controlled architectures described in Chapter 2 can provide a means of force amplification because these systems are physically anchored to the ground and can thus support the reaction loads. Conversely, force amplification strategies for *ungrounded* surgical devices is a relatively under studied field. The hand-held force magnifying device proposed by Stetten *et al* is the first reported effort that provides a means of ungrounded force magnification of tool-tissue interactions [144]. This device is a hand-held stylus that is coupled to an electromechanical actuator which is, in turn, mechanically anchored to the user's hand. A force sensor is used to measure the axial force experienced by the tool tip and the on-board actuator then produces a force in proportion to this tool tip force, so as to create an amplifying effect. The on-board actuator generates equal and opposite reaction forces to those intended for the purpose of creating the amplification effect which are supported by the bracing anchored to the user's hand.

The haptic feedback strategy proposed in this chapter also provides a force-amplifying function through an ungrounded architecture. A distinct novelty of the proposed strategy compared to that of the work by Stetten *et al*, is that the force-feedback is provided to the user's index fingertip and the corresponding reaction force generated by the device is transmitted through the user's grip of the device casing. This allows the device to be held in a less constrained configuration and forgoes the need for a bracing or anchoring mechanism. The primary contribution of this chapter is the proposal of a hand-held device that can amplify tool-tissue forces to the surgeon's fingertip and validation that this improves human haptic perception of forces that cannot otherwise be felt.

### 3.1 Force-Amplifying Device

The instrument developed is configured like a stylus so that the operator grips the casing and places their finger on a slider that can translate back and forth. A central bore in the instrument allows a surgical tool to be passed through and a clamp allows the instrument to be affixed to the aforementioned slider that contacts the operator's finger. The slider and instrument can then translate together relative to the stylus casing. A force sensor mounted at the instrument tip senses the contact force between the instrument tip and the environment. An actuator within the stylus casing then exerts forces on to the slider that are proportional to the forces exerted at the instrument tip in a closed-loop control scheme where a sensor measuring the force exerted on to the operator's fingertip is used to close the control loop. The reaction forces generated by the actuator are then transmitted to the operator's grip of the stylus so that the haptic feedback is completely ungrounded. This arrangement is presented in Figure 3.1.

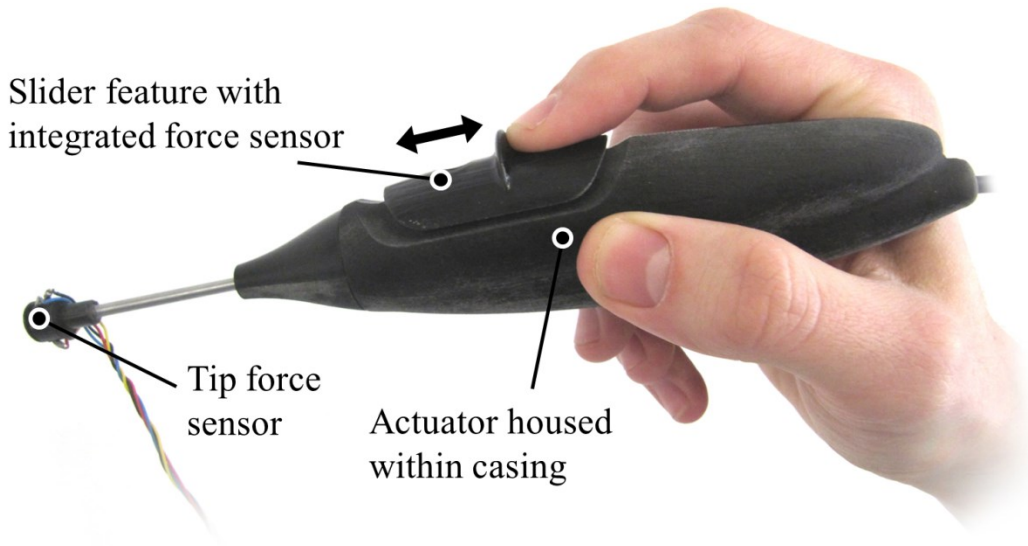


**Figure 3.1:** A photograph of the hand-held, force-amplifying device showing the key features.

The force sensor at the instrument tip measures the contact force which shall be denoted as ( $F_t$ ). The actuator within the casing generates a force, ( $F_m$ ) which is directly proportional to  $F_t$ . The operator can set the factor by which  $F_m$  is proportional to  $F_t$ . The summation of  $F_t$  and  $F_m$  is the force that is exerted on to the operator's finger, ( $F_f$ ) so that:

$$F_t + F_m = F_f \quad 3.1$$

This is illustrated graphically in Figure 3.2. The force  $F_f$  is measured by a sensor embedded in the slider that the operator's finger is in contact with. The ratio of the  $F_f$  and  $F_t$  is the scaling factor ( $s$ ) by which the tip force is being amplified:

$$s = \frac{F_f}{F_t} \quad 3.2$$

This force arrangement is shown graphically in Figure 3.2. The slider has a ridge feature that transmits the load on to the operator's fingertip in a compressive, rather than shearing manner.

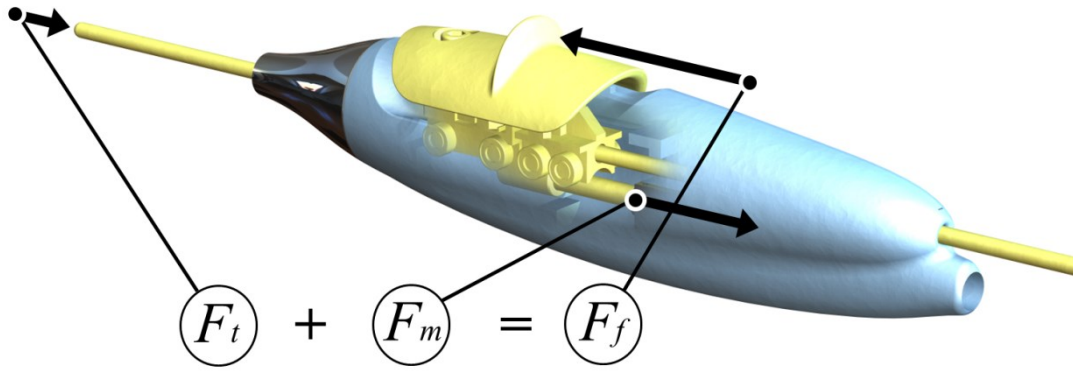


**Figure 3.2:** A ghosted CAD rendering of the hand-held, force-amplifying device labelling the forces  $F_t$ ,  $F_m$  and  $F_f$ . The components rendered in yellow can translate relative to the stylus casing.

### 3.1.1 Mechanical Design

The main function of the force-amplifying device is to increase the user's sensitivity to force and to reduce the minimum force threshold that the user can perceive. It is therefore important that the user's perception of the tool-tissue contact force is not contaminated by unwanted forces, namely friction, inertia, mechanical noise (vibration) or discontinuities that arise from backlash. The mechanical design and choice of actuator are therefore very important in catering to these requirements. A three-phase electromagnetic linear motor

(LM) (LM0830-015-01, Faulhaber) was chosen to provide the additional force required for amplification. The LM can provide high forces (1.03N continuous, 2.72N peak) relative to its size (8×8×30mm) and weight (15g), making it suitable for a compact hand-held device. Another reason for the selection of the LM is that it has just one single moving component: a magnetic shaft which translates within the bore of the motor housing, making the motor completely free of backlash and with minimal friction. The LM has integrated Hall-effect sensors to provide position feedback and permit motion control. Figure 3.3 and Figure 3.4 illustrate the engineering features of the force-amplifying device.

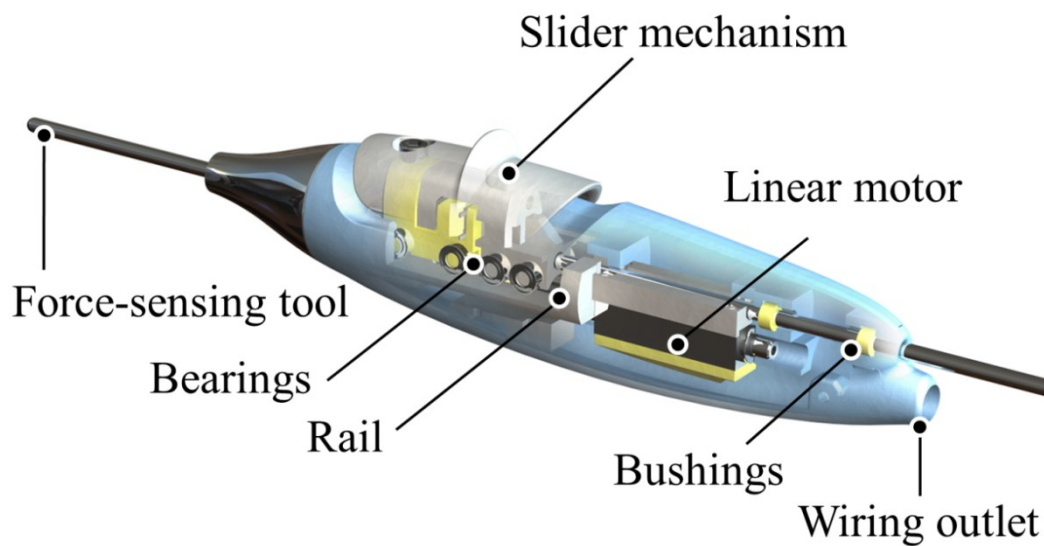


**Figure 3.3:** Shows a ghosted view of the force-amplifying device highlighting the key engineering features.

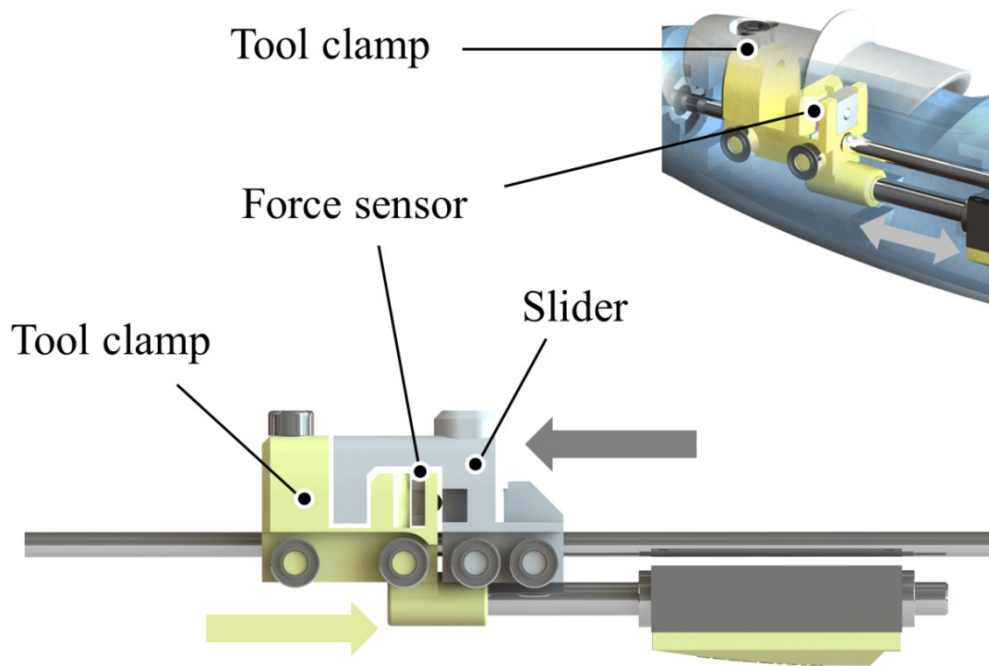


**Figure 3.4:** Shows a ghosted view of the force-amplifying device force sensing assembly and how load is transmitted to the force sensor that senses the fingertip force  $F_f$ .

The vectors of the forces acting on the sliding assembly;  $F_t$ ,  $F_m$  and  $F_f$  are not collinear (this is illustrated in Figure 3.4) which leads to a moment being generated as the device is loaded during operation. In order to avoid jamming of the sliding assembly due to this moment, the slider assembly translates on bearing rollers that are seated within a rail so as to minimise friction. The rail is mechanically clamped to the casing shell that is held by the user's grip. The slider assembly is comprised of two components: 1) the tool/probe clamp, to which the tool/probe is affixed and 2) the slider feature that is in contact with the user's fingertip. The measurement of  $F_f$  is made through a force sensor that is placed between these two components. The routing of the motor and force sensor wiring is made so that there is no mechanical impedance to the slider assembly motion. The force sensor wiring is also routed within the device so as to avoid the vicinity of the magnetic field generated by the LM, which would otherwise induce noise into the force measurements. The device (excluding external wiring) weighs 80g.

### 3.1.2 Control Implementation

A closed-loop, Proportional-Integral-Derivative (PID)-based force control strategy is used to control force amplifying device. The force controller is designed to ensure that the force exerted on to the operator's finger is always in proportion to the input force sensed at the tool tip. The force measurement of the tool/probe tip,  $F_t$  forms the input to the control loop, it is multiplied by a user-defined scaling factor and so becomes the desired force  $F_d$ . The measured force exerted on the user's fingertip,  $F_f$  provides the feedback to complete the loop. The error ( $E$ ) between the desired force and measured force is fed into a PID control algorithm so that the controller attempts to maintain a zero error. The analogue voltage output ( $V$ ) from the real-time embedded industrial controller, the CompactRIO (cRIO-9014, National Instruments Corp) is subsequently fed in to the LM motion controller (MCLM 3006S, Faulhaber). This motion controller controls the position of the LM shaft based on the input voltage using a PID control scheme. The integrated Hall-effect sensors of the LM are used to measure position of the shaft ( $P$ ) for the closed-loop position control. The slider assembly is actuated by the LM and the user's finger acts as a spring in generating the measured reaction force between the sliding assembly and the user's finger. A high level block diagram of this control scheme is shown in Figure 3.5.

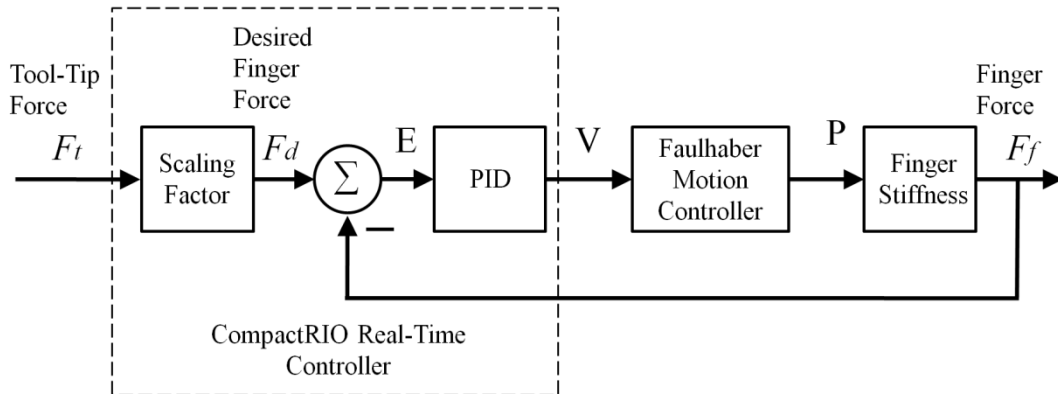


**Figure 3.5: A block diagram showing the closed-loop force control of the scaling factor for the proposed hand-held device.**

The PID control algorithm was implemented on a real-time embedded industrial controller, the CompactRIO (cRIO-9014, National Instruments Corp). Real-time controllers ensure that

the relevant calculations to be performed by the controller are guaranteed to be executed within a determined timeframe. Real-time controllers are typically used in medical mechatronic systems to ensure robustness by performing critical calculations within a distinct time period. It is also important that software calculations are executed at high speed in haptic feedback applications because of the relatively high temporal resolution of human haptic perception. Prior literature has reported update rates in the order of 1kHz as being a minimum requirement for smooth haptic rendering [109]. The CompactRIO system is comprised of a real-time controller (cRIO-9014), a field-programmable gate array (FPGA) backplane with reconfigurable input/output (I/O) modules. The FPGA is capable of rapidly acquiring and transmitting the input/output lines respectively. It can perform low-level operations at a nominal rate of 30kHz. The FPGA is connected to the real-time controller through an internal, high speed Peripheral Component Interconnect (PCI) bus. The real-time controller is connected to a host personal computer (PC) through an Ethernet connection. The control hardware is configured using the LabVIEW programming environment. A program was written on to the FPGA to acquire and digitise the relevant analogue input channels from the sensing lines. The program then passes the digitised signal to the real-time controller. A second program is deployed on the real-time controller which incorporates two parallel control loops. A deterministic loop running at 1kHz is implemented so as to receive the digitised inputs, perform the PID control algorithm and send an output back to the FPGA. A non-deterministic loop allows data to be passed to the host PC for logging during experimentation. The digitised sensing signals are passed from the deterministic loop to the non-deterministic loop using a global variable function. The non-deterministic loop runs at 100Hz. A third program, implemented on the host PC allows the down-sampled sensor signals to be continuously passed from the real-time controller through the Ethernet connection to the PC, where the values are logged. For the purposes of troubleshooting, the sensing signals are also visualised on the host PC on a waveform chart. The control software implementation is illustrated in Figure 3.6.

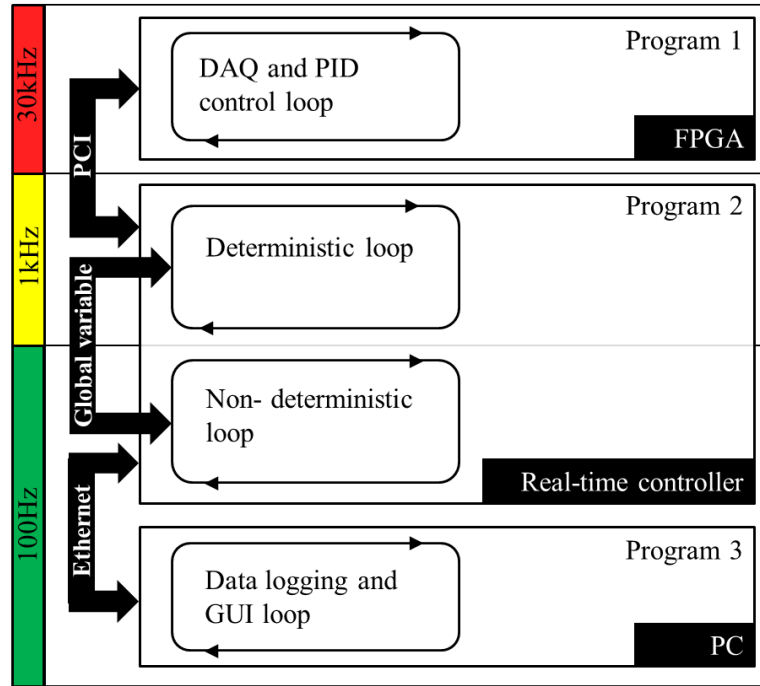


**Figure 3.6: Diagram showing the embedded controller architecture, the software is implemented on three separate programs with four separate while loops running concurrently at different rates.**

### 3.1.3 Force Sensing and Signal Processing

For accurate force sensing at the tool/probe tip, Piezoresistive-based force sensors (FSS1500NS, Honeywell) are employed. Operational amplifiers are used to amplify the sensor measurements. The amplified signal is then filtered using a low-pass, first-order RC filter having a cutoff frequency of 20 Hz to reduce high-frequency noise, including the 50 Hz noise due to nearby power sources. It is expected that a 20 Hz cut off frequency is above the typical operating frequency required during micromanipulation and therefore should not affect the performance of the device due to a loss of information. The force sensors and Hall-effect sensor lines are independently shielded from the high current-carrying LM power lines so as to avoid electromagnetic interference.

## 3.2 Force amplification Bench Tests

In order to assess the intrinsic sensitivity of the force-amplifying device, a detailed bench test was performed. The experiment evaluates the performance of the device by observing the force sensor measurements of  $F_t$  and  $F_f$ . In this experiment, the device was driven into a spring at various desired scaling factors ranging from 1.5-15. The sensors were calibrated against a reference F/T sensor, (Nano17, ATI Industrial Automation, Inc., USA). Figure 3.7 and Figure 3.8 show the relationship between  $F_t$  and  $F_f$  for a range of scaling factors, linear regression models using the least squares method for the measured scaling factors are also shown in these plots and are numerically summarized in Table 3.1.

**Table 3.1: Linear regression models for the measured scaling factors using the least squares method.**

| Desired Scaling Factor | Regression Ratio | Residual Force (N) |
|------------------------|------------------|--------------------|
| 1.5                    | 1.6              | 0.010              |
| 2                      | 2.1              | 0.026              |
| 3                      | 3.0              | 0.056              |
| 4                      | 4.0              | 0.004              |
| 5                      | 4.7              | 0.003              |
| 8                      | 8.4              | 0.013              |
| 12                     | 11.4             | 0.038              |
| 15                     | 12.0             | 0.059              |

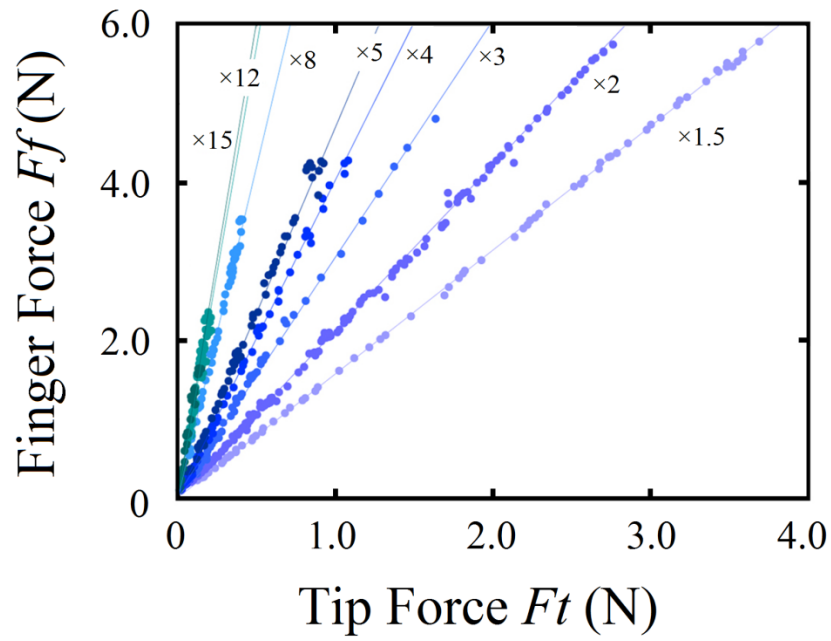


Figure 3.7: The relationship between  $F_t$  and  $F_f$  for scaling factors of 1.5-15 over a tool tip force range of 0.0N to 4.0N. Linear regression models based on this data are also shown.

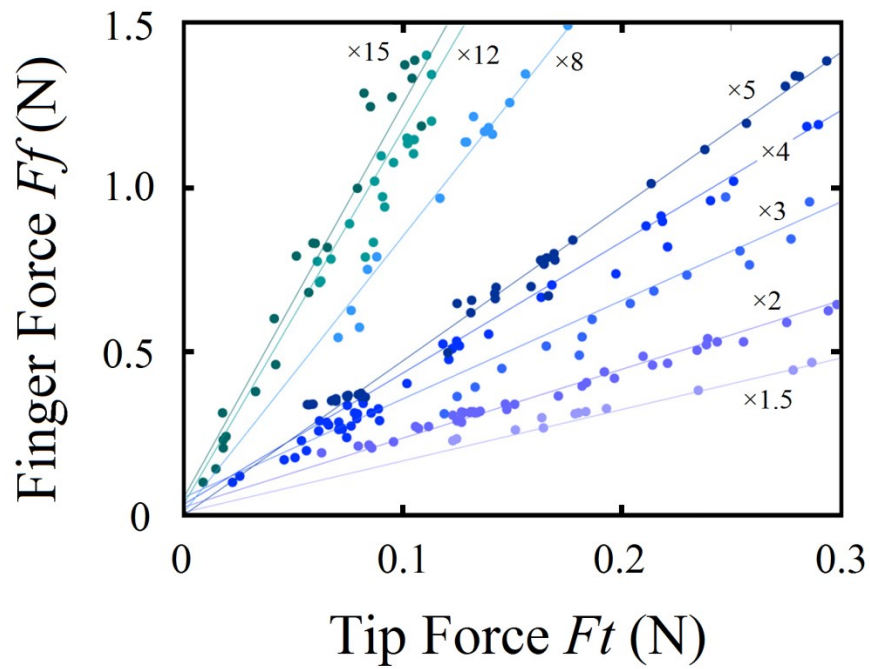


Figure 3.8: The relationship between  $F_t$  and  $F_f$  for scaling factors of 1.5-15 over a tool tip force range of 0.0N to 0.3N. Linear regression models based on this data are also shown.

The results confirm that the closed-loop control scheme performs as required. The measured scaling factors of the force-amplifying device remain constant up until the LM stalls at around 2.7N when the motor controller current limits are reached; beyond this point, there is a discontinuity in the  $F_t - F_f$  relationship, this is not shown in Figure 3.7. Additionally, the maximum  $F_f$  values observed are lower for higher scaling factors (best seen in Figure 3.7); this is because the motor provides a greater proportion of  $F_f$  so that the peak values tend towards the motor stall force. The regression models presented in Figure 3.7 and Figure 3.8 are based only on the linear region of the  $F_t - F_f$  relationship. At higher scaling factors, the control system is more susceptible to overshoot from the sensor noise which leads to a reduction in scaling factor consistency. At higher scaling factors, the actuation of the slider takes place over a shorter period of time, leading to fewer force measurements being recorded of the  $F_t - F_f$  relationship. This leads to distortion in the calculated regression models, most evident in the 15 scaling factor regression model where the residual value is highest (0.0585N) and the regression ratio is subsequently lower than expected (12.0).

### 3.3 Force Threshold Detection User Study

For assessing the practical value of the platform, a user study was conducted. The study evaluates the effectiveness of the force-amplifying device in improving human force perception. The study assessed the threshold at which users could detect the tool tip contact for three cases. Two cases consider use of the device, *with* and *without* the force amplification activated. Additionally, a control case was performed using a syringe to compare the performance of the device against standard clinical equipment, as per the protocol reported by Stetten *et al* [144]. The syringe used in the control trials weighed 7.8g. These experiments were conducted in a randomised fashion to prevent biasing. Seven users (5 male, 2 female) were recruited for these trials and all gave their informed consent to take part in the study. All of the users described themselves as being right-handed.

The experimental setup is shown in Figure 3.9. The users were requested to hold the force-amplifying device stationary in their dominant (right hand), with the tool tip facing downwards. A platform, consisting of a calibrated force-torque (F/T) sensor (Nano17, ATI Industrial Automation, Inc., USA) and low-stiffness spring was translated vertically upwards so as to make contact with the tip of the device. The users also held a push button in their non-dominant hand and were requested to immediately depress the push button at the instant

they could feel the presence of the contact force. For each user, 3 measurements were taken for all 3 experiments. The F/T sensor output was acquired in to the CompactRIO through a data acquisition (DAQ) card (NI-6221, National Instruments Corp., USA). The acquired data was then made available to the CompactRIO real-time controller via a DAQ device (NI-USB6008, National Instruments Corp., USA). The push button signal was also acquired by the CompactRIO and the channels were both logged by the host PC. Figure 3.11 shows a block diagram of the hardware configuration in this experiment. Once the user depressed the push button, the corresponding force value acquired from the F/T sensor was recorded as the force detection threshold for each trial.

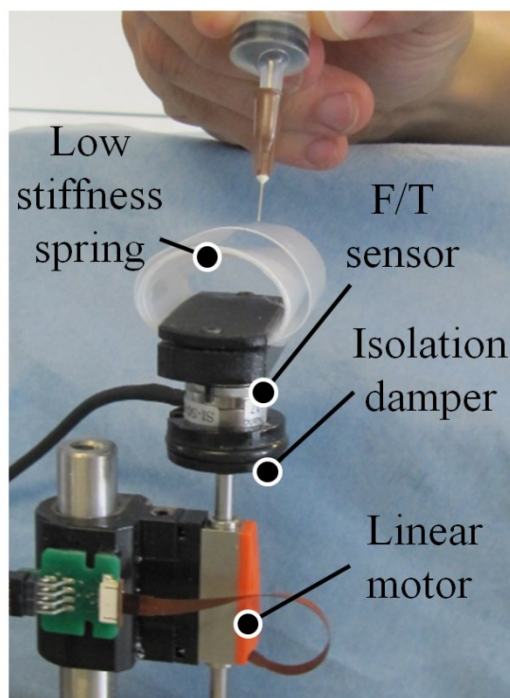


**Figure 3.9:** Experiment setup for the force threshold detection user study with the syringe.

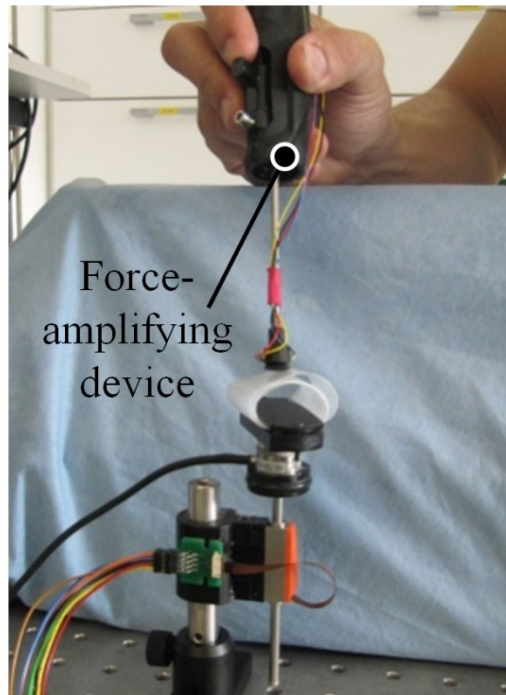


**Figure 3.10:** Experiment setup for the force threshold detection user study with the force-amplifying device.

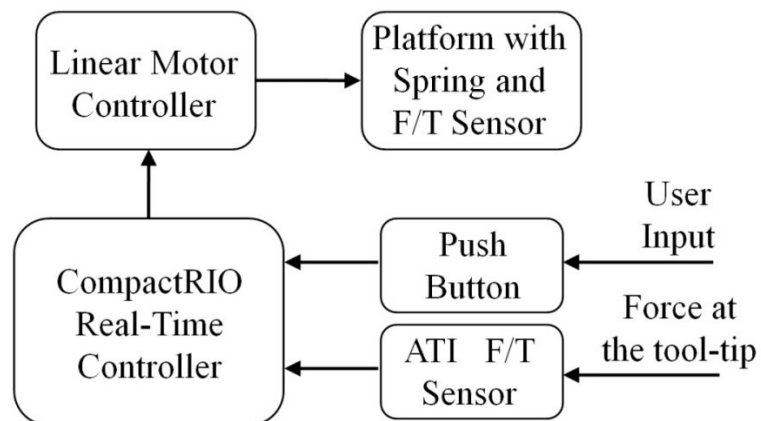


**Figure 3.11:** Block diagram showing the hardware and data acquisition equipment used in the experimental setup.

The use of a translating platform in this experiment allowed the forces exerted on to the device/syringe to be displacement-controlled, through the stiffness of the spring. A low stiffness spring was used so as to minimise the error in measuring the user's minimum force threshold due to erroneous motion of the user's hand that can be as large as a few hundred microns [63]. Furthermore, translating the F/T sensor-spring platform into the device/syringe (rather than the requesting the user to probe a stationary F/T sensor) avoided any effects of the subject's proprioception sense. Otherwise, it is hypothesised that the users could learn the location of the spring and intuitively exert less force over the course of multiple trials. The F/T sensor and spring were translated using an electromagnetic three-phase LM (LM0830-040-01, Faulhaber) which was chosen as it can produce smooth motions with low mechanical noise. Further efforts to reduce the effects of noise were made through the use of a first order low-pass filter on the F/T sensor as well as a nitrile isolation damper which was seated between the translating platform and F/T sensor. Additionally, the experiment took place on a vibration-isolated optical table. The LM was ramped to a constant velocity of 1.75mm/s so as to avoid inertial measurements of the platform assembly during force measurements. The spring was formed from a polyethylene strip that was wrapped into a coil so as to form multiple layers. The layering of the spring allowed graduated loading of the device/syringe tip and avoided users making contact with the platform assembly which is much stiffer and would lead to erroneous measurements being made. Efforts were made to remove the effects of other sensory cues experienced by the user during the user study: users were blindfolded and wore isolation headphones to remove visual and auditory cues.

A scaling factor setpoint of 15 was used for the trials in which force amplification was active. This scaling factor was chosen as it was the highest scaling factor evaluated in the bench test and it was hypothesised that a higher scaling factor would improve force perception more significantly. The sensors were all zeroed prior to each trial. In the trials using the device when force amplification was not active, the users were requested to hold the device in the same manner as they did for the trials when force amplification was active: with the user's finger placed on the slider. The results of the experiment are presented in Figure 3.12, Table 3.2 and Table 3.3.

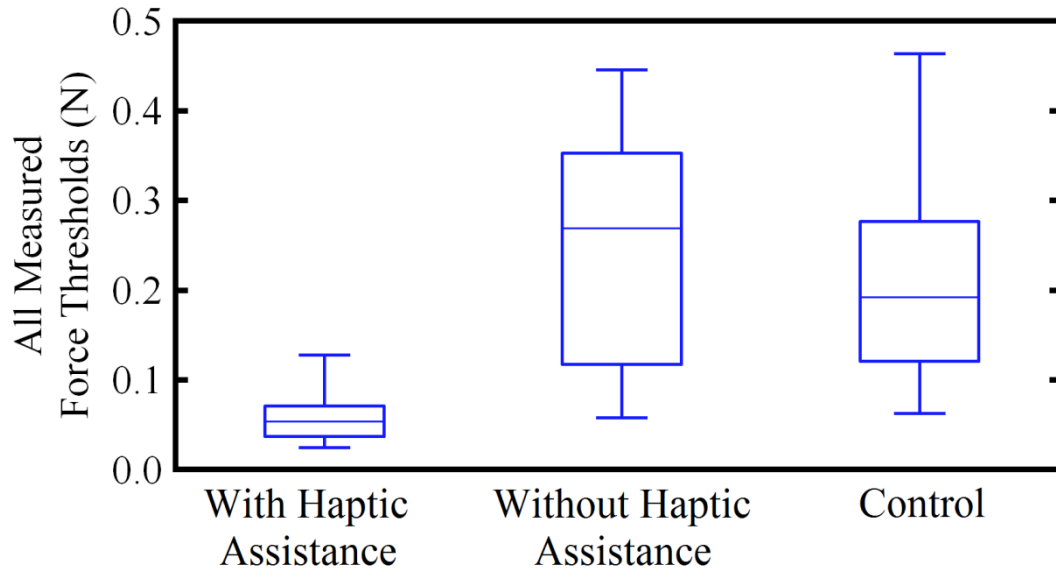


**Figure 3.12:** Box plot showing the distribution of measured force detection thresholds for the seven users across the three conditions examined during the user study. Boxes indicate the lower quartile, upper quartile and median, whiskers represent the minimum and maximum values.

**Table 3.2:** Key force detection thresholds statistics for the user study.

|                                             | <i>With</i><br>Amplification | <i>Without</i><br>Amplification | Control Study |
|---------------------------------------------|------------------------------|---------------------------------|---------------|
| Minimum force threshold (N)                 | 0.023                        | 0.056                           | 0.061         |
| Maximum force threshold (N)                 | 0.129                        | 0.447                           | 0.465         |
| Median force threshold (N)                  | 0.054                        | 0.271                           | 0.193         |
| Interquartile range of force thresholds (N) | 0.034                        | 0.236                           | 0.156         |

**Table 3.3:** Statistical comparison tests for the three experimental conditions in the force threshold user study.

| Comparison of force threshold distributions | p-value (Mann-Whitney U) |
|---------------------------------------------|--------------------------|
| With v without                              | <0.001                   |
| With v control                              | <0.001                   |
| Without v control                           | 0.4734                   |

### 3.4 Discussion and Conclusions

The results of the user study demonstrate that the hand-held force-amplifying device can reduce the threshold at which users could perceive contact force of the tool tip. The force-amplifying device, when trialled *with* force amplification active, improves human force perception compared to the two other cases: use of the device *without* force amplification and the control case in which users handled a lightweight syringe.

When the device was trialled *with* force amplification, the median force detection threshold for all of the users was reduced by a factor of 5 when compared to the case when the device was trialled *without* force amplification. The minimum reported force detection threshold was comparatively reduced by 59% and the maximum reported threshold was also comparatively reduced by 71% when force amplification was active. In comparison to the control case, when the device was trialled *with* force amplification, the median force detection threshold for all of the users was reduced by a factor of 3.6. There was a 40.3% increase in the median force detection threshold between the case of the device being trialled *without* force amplification and the control case. Mann-Whitney U statistical comparison tests were performed on all of the force detection thresholds measured for the three cases considered in this user study. The force detection thresholds that were observed when the device was trialled *with* force amplification were significantly reduced in comparison to when the device was trialled *without* force amplification ( $p < 0.001$ ) and in comparison to the control case ( $p < 0.001$ ). There was no statistically significant difference between the observed force detection thresholds for the case of the device being trialled *without* force amplification and the control case ( $p = 0.4734$ ). The force amplifying device was also demonstrated to improve the consistency with which a minimum force detection threshold can be perceived: the interquartile range of force detection threshold for the *with* force amplification case was 34mN compared to 236mN and 156mN in the *without* force amplification and control cases, respectively.

The reduction in the median force detection threshold observed in the case *with* force amplification compared to the other cases was lower than the pre-set scaling factor of 15. It is hypothesised that experimental errors are a significant cause of this. The force detection thresholds observed across the entire study would have been higher than the actual perceived

force threshold as a result of the time delay between the user perceiving tool contact and physically depressing the push button. During this latent period, the force exerted on to the low stiffness spring would have increased as the platform would have translated and compressed the spring further. It is likely that this error would have differed from user to user but would remain consistent across the different cases examined. This error will have therefore disproportionately increased the observed force detection thresholds for the case *with* force amplification relative to the *without* force amplification and control cases.

The mechanical response of the force-amplifying device when used *with* force amplification will have introduced some additional latency between the tool tip making contact and the user perceiving contact. Inertial, gravitational and frictional loads experienced by the slider/roller mechanism and tool shaft of the force-amplifying device would contribute to this mechanical response time and will have artificially increased the observed force detection thresholds due to the spring compression effect previously cited. Anecdotally, Stetten *et al* also observed that the perceived force amplification was consistently lower than the actual magnification in their psychophysical studies and they cite mechanical factors, including as weight a probable cause of this. The loss of performance due to additional weight and other contaminating forces is in accordance with Weber's law. Another possible explanation for the discrepancy between the pre-set scaling factor and observed device performance is due to the different tactile stimulation modes through which the user would perceive tool contact. When the users trialled the force-amplifying device *with* force amplification they would experience a compressive tactile sensation at the tip of their index finger. When handling the syringe, the tactile stimulation occurs in a shearing mode and is distributed across all points of grasping contact.

A further source of error is the erroneous physiological motion of the device caused by the user, including tremor and hand drift, would have resulted in a change of deflection in the spring and therefore give an error ( $\pm$ ) in the force detection threshold measurements. Attempts at mitigating these errors were made by actuating the platform with a low velocity ramp and through use of a low-stiffness spring. The time delay error previously mentioned will have varied from user to user and added further uncertainties in to the results. These errors are systematic across the entire study and do not bias the results. Nonetheless, for all

of the reasons previously cited, it is probable that the force-amplifying device improves tool contact perception to a greater extent than has been concluded in this experiment.

This chapter has presented an ungrounded hand-held force-amplifying augmenting device designed to improve user perception of tool contact forces by providing tactile feedback to the user's fingertip. The bench tests reported in this chapter demonstrate the device's ability to intrinsically amplify force, allowing scaling factors of up to 15, greater than that achieved by comparable devices being developed by researchers [144][145]. The user study confirmed that user perception of tool contact is unequivocally improved; the study also demonstrated improvements in the consistency with which tool contact can be perceived by users. The experiments performed nonetheless highlight a number of improvements that are subsequently addressed in Chapters 4 and 6.

The experiments performed highlight intrinsic limitations of the developed device. A significant limitation was the force sensor which has a limited resolution in the sub-threshold range for human tactile perception. Whilst scaling factors of 15 were reported, the device was primarily evaluated at force ranges at the upper limits of microsurgery where this type of augmenting device is likely to be at its most useful. The integration of high resolution, micro-scale force sensors in to hand-held mechatronic surgical tools is thus investigated in Chapter 4. The work in this chapter also highlights the importance of minimising weight and optimising device ergonomics which could serve to degrade rather than improve user performance. This point is especially true in the targeting of delicate microsurgical interventions. The user study also highlights the potential importance of analysing aspects such as the device bandwidth which are also considered in Chapter 4.

# **Chapter 4: Ungrounded Hand-Held Microsurgical Forceps with Force-Feedback for Micromanipulation<sup>†</sup>**

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<sup>†</sup> Content from this chapter was published as:

**Hand-Held Microsurgical Forceps with Force-Feedback for Micromanipulation.**

Payne C.J., Rafii-Tari H., Marcus H. and Yang G-Z. IEEE International Conference on Robotics and Automation, Hong Kong, 2014. pp284-289.

Chapter 3 investigated a force-amplifying prototype device that could improve human perception of tool contact force. This chapter evolves the work of Chapter 3, by considering the more specific case of force-feedback for hand-held microsurgical forceps. Forceps are routinely used in microsurgical interventions across many clinical domains as varied as neurosurgery, plastic surgery, ophthalmic surgery, otorhinolaryngology and microvascular surgery. Hand-held microsurgical forceps are utilised by surgeons for the purposes of grasping and manipulating delicate tissue or for manipulating sutures and micro needles. Whilst microsurgical forceps designs vary across the different disciplines, most designs feature a pair of elastic cantilevers that are grasped by the surgeon. Spring-loaded forceps configurations allow the surgeon to ergonomically and precisely position the tips of the forceps. However, the surgeon's intrinsic haptic perception of the forces being exerted at the forceps tips is diminished as a result of the elastic spring return force. The force required to overcome this spring return in the forceps can be many times greater than the force exerted at the instrument tips, especially when delicate tissue is handled. One of the theoretical novelties explored in this chapter is thus the use of tool-tissue force-feedback as means of bypassing a contaminating force. Conceptually, researchers have examined this problem before, for example, in the context of epidural needle insertion [146]. In this particular case, the clinician often relies on tactile cues from subtle changes in the needle insertion forces so that they can determine when an effective puncture has occurred. However, these tactile cues are not always clearly felt by the clinician because the shear forces exerted on to the needle contaminate the important puncturing forces that are being transmitted to the needle tip. By augmenting the needle tip forces above the needle shearing forces, researchers have demonstrated improvements in puncture detection. This chapter poses the same problem in the context of hand-held microsurgical forceps. Researchers have previously devised hand-held forceps that incorporate piezoelectric sensing and actuation [147] and modified designs for skill assessment [91] but the question of human tip force perception has not been studied.

This chapter also considers a number of engineering advancements to the preceding work reported in Chapter 3. One key point identified in Chapter 3 was the need for tool tip force sensing at low force levels in the range below intrinsic human haptic perception. In recent years, researchers have focussed on micro-scale force sensing forceps designs, particularly for ophthalmological applications. One popular technique makes use of optical fibre sensors to provide high sensitivity force sensing in ophthalmic instruments. Fibre Bragg grating

(FBG) sensors with sub milli-Newton accuracy have been integrated into surgical tools for retinal microsurgery [138]. This work has primarily focussed on sensing the deflection of the forceps shaft so as to infer the lateral traction forces that are exerted by the keyhole ophthalmic instruments. Subsequent design iterations of the force sensing system have considered the axial force [148] and more recently researchers have integrated these sensors in to tremor suppression devices and with motorised forceps grasping [46][47]. Despite these advances, there has been little investigation of the hand-held elastic cantilever type forceps of interest in this chapter. In terms of sensing methodologies, micro-electromechanical systems (MEMS)-based sensors have been developed for force-feedback systems [136], force-sensing ophthalmic instruments [137] and in micro-gripper designs [149]. Strain gauge-based sensing has also been adopted for micro-gripping applications [150][151]. This chapter examines the use of silicon strain gauges to obtain high resolution force sensing at sub-threshold levels. Two forceps design iterations are proposed and evaluated in this chapter. Unique micro-scale force sensor designs are theoretically analysed and experimentally validated. The dynamic response of mechatronic hand-held microsurgical forceps is also considered in this chapter. A user study demonstrates the impact of the force-feedback concept on human haptic perception.

#### **4.1 Hand-Held Force-Feedback Forceps: Prototype I**

The force-feedback forceps are designed according to standard microsurgical forceps whereby the operating surgeon must compress opposed elastic cantilevers in order to close the forceps. The device is presented in Figure 4.1. A force sensor is integrated into one tip of the forceps in order to allow measurements of the forceps grasping force ( $F_g$ ). An actuator is then used to exert amplified forces on to the operator's fingertip. The reaction forces generated by the actuator are transmitted through the operator's grip of the force-feedback forceps. Crucially, the actuator is mounted perpendicular to the grasping direction of the forceps elastic cantilevers. This design feature ensures the operating surgeon can distinguish between the amplified-force-feedback from the forceps tip sensor and the forces required to compress the spring levers of the forceps by exploiting the ability of the human fingertip to resolve forces spatially. A second force sensor, mounted between the actuator and the operator's fingertip allows measurements of the force exerted on the operator ( $F_f$ ) which permits closed-loop force control to provide amplified force-feedback. As was the case in Chapter 3, the ratio between the  $F_g$  and  $F_f$  is set by a scaling factor ( $s$ ) so that:

$$s = \frac{F_f}{F_g}$$

4.1

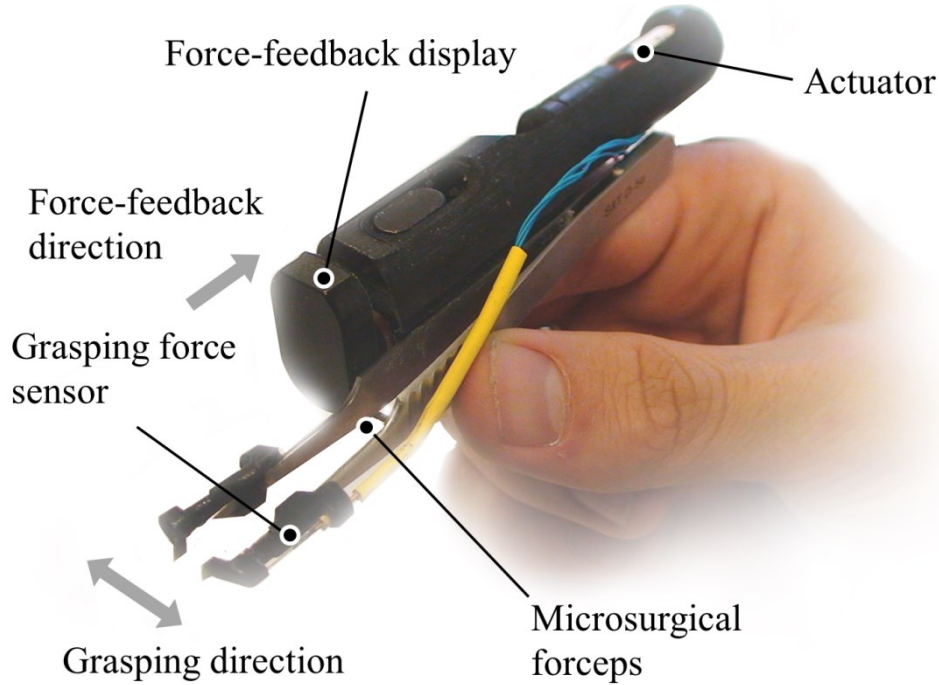


**Figure 4.1: A photograph of the hand-held force-feedback forceps prototype I.**

#### **4.1.1 Grasping Force Sensor Development: Prototype I**

A critical aspect of force-feedback systems for microsurgical applications is the tool-tissue force-sensing. Commercially-available, fully-integrated force-sensors are typically large and designed to measure forces at the newton-scale. With such sensors, force measurements at the milli-Newton scale require significant amplification and are therefore prone to noise interference, this was a challenge experienced in Chapter 3. Another key consideration is the physical size of the sensors which must remain compact for practical clinical deployment. In consideration of these aspects, a silicon strain gauge-based force sensor (AE801, Kronex, Oakland, USA) was chosen to make measurements of the forceps grasping forces. The sensor is configured as a silicon cantilever beam which is sensitive to bending deflections. The sensor is compact (1.8mm diameter  $\times$  10mm total length) and can measure a maximum force of 120mN, leading to good sensitivity and a high signal-to-noise ratio at milli-Newton

force levels. The silicon strain gauge sensor exhibits good linearity ( $\pm 0.25\%$  FSO) and low hysteresis ( $\pm 0.1\%$  FSO). Further specifications of the strain gauge sensor are provided in Appendix I. Another advantage of strain gauge sensors is their relatively low cost and that their operation does not necessitate additional hardware, such as the integrators required for FBG sensors. A disadvantage of the AE801 is its fragility; the sensor is only rated to accept a 120% overload (144mN) and is also fragile to other out-of-plane and torsional loads, making it impractical for direct deployment in a surgical device. To address this issue, the AE801 is integrated into an assembly that utilises an elastic beam in parallel to the sensor beam so as to support a proportion of the total grasping load. The sensor assembly is presented in Figure 4.2 and Figure 4.3.

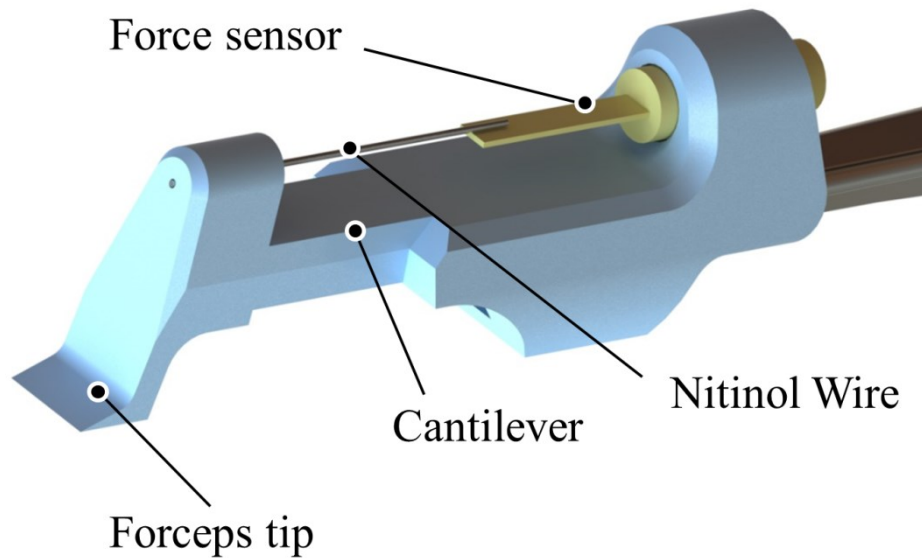


**Figure 4.2:** CAD illustration of the grasping force sensor assembly used in prototype I.

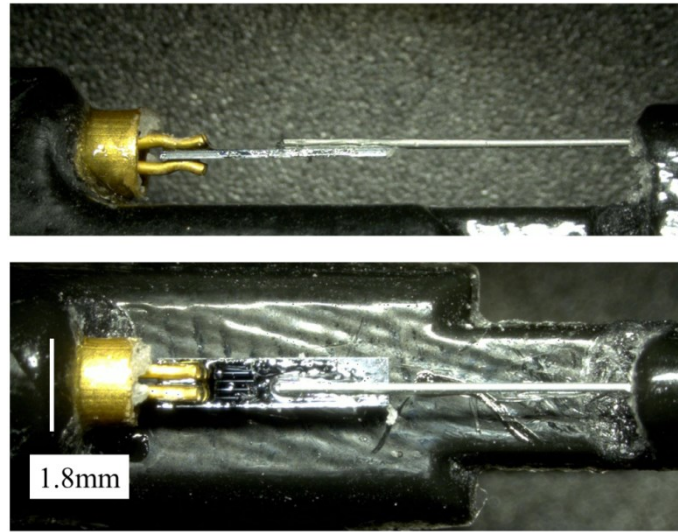


**Figure 4.3: Photographs of the force sensor assembly used in prototype I.**

Since the AE801 is sensitive to displacements at the  $\mu\text{m}$  scale, the sensor is coupled to the forceps tip using a 0.17mm nitinol wire which acts as a low stiffness spring so as to prevent significant undesirable pre-loading during assembly. It also avoids the necessity for high precision manufacturing which cannot be achieved from the rapid prototyping methods used to manufacture the prototype I sensor assembly. The sensor was designed through empirical testing, with the aim of achieving a minimum sensing range of 0-300mN so as to be sensitive enough at low force levels whilst remaining robust to higher loading. This was done through a process of tuning the stiffness of the parallel cantilever and the stiffness of the coupling spring through geometrical alterations to the design. The AE801 was rigidly bonded to the forceps tip assembly using a cyanoacrylate adhesive to minimise hysteresis effects.

#### **4.1.2 Mechatronics Design: Prototype I**

A base chassis and the sensor assembly previously described are mounted to standard, stainless steel microsurgical forceps (S&T D-5A, Dumont Surgical, Gaithersburg, USA) that are commonly used for grasping tasks in micro-anastomosis. The forceps tips have a total grasping stroke of 11mm. The force-feedback forceps are designed to operate as close to conventional forceps as possible with minimal interference from the additional components. The tight spatial constraints imposed by the short, angled tips of the forceps and the grasping location of the operator's fingertip make it impractical to mount a force-feedback actuator ahead of the fingertip location. It is therefore mounted at the proximal region of the forceps,

off the axis of the forceps levers so as to allow space for the operator's finger. A voice coil actuator (VCA) (LVCN-013-013-02, Moticont) was used to provide the force-feedback. The forces generated by the VCA are transmitted through a low friction slider mechanism which is constrained to purely axial motion through an anti-rotation keyway. The slider is coupled to a plate which makes contact with the operator's fingertip through the feedback force sensor (8438 5005, Burster). The total weight of the device is 48g. These design features are illustrated in Figure 4.4.

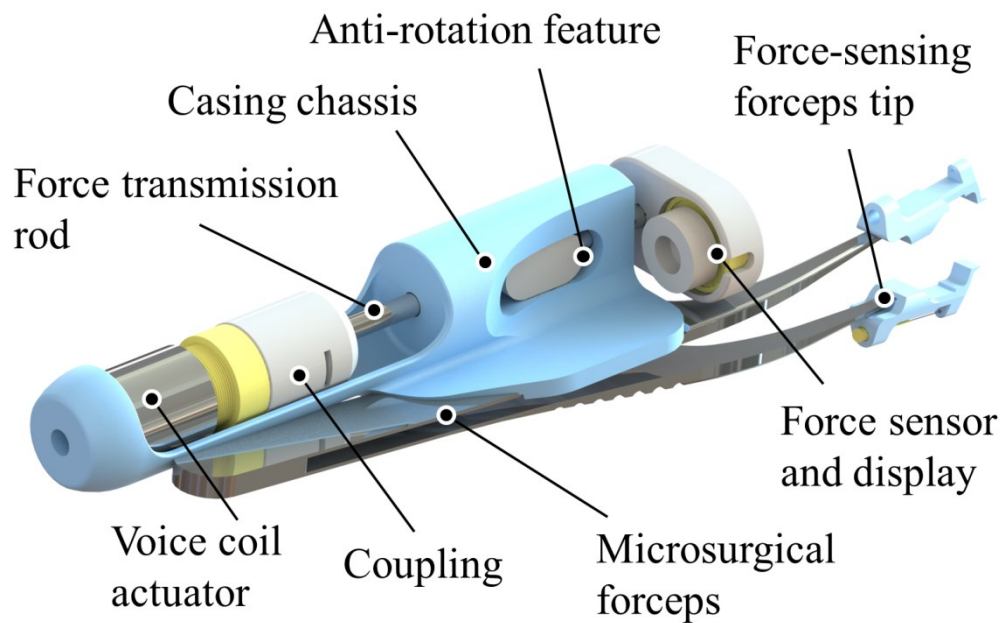


**Figure 4.4:** Shows the key engineering features of the microsururgical force-feedback prototype I.

The addition of the mounting chassis and VCA to the elastic forceps forms a cantilever mass-spring system. Therefore, the chassis is mounted on the opposite forceps lever to that of the lever containing the force sensor. This avoids control instability arising through excitation due to the mechanical coupling of the sensor and actuator.

#### **4.1.3 Control Implementation: Prototype I**

The control scheme for the force-feedback forceps is implemented on a multi-threaded, real-time controller (CompactRIO, cRIO-9022, National Instruments Corp.). The same

architecture proposed in Chapter 3 was used in this implementation (see Figure 3.6). Both the grasping force sensor and feedback force sensor input voltages are acquired by the controller following amplification through separate Wheatstone half-bridge arrangements before being digitised and converted to milli-Newtons by the controller. The forceps grasping force measurement  $F_g$  is then multiplied by the pre-set scaling factor before forming the setpoint to a PID control loop. The fingertip force measurement  $F_f$  provides the feedback in order to facilitate closed-loop control. The output from the PID algorithm is converted to a voltage by the CompactRIO which is then received by a linear amplifier (LSC 30/2 linear 4Q Servoamplifier, Maxon) which supplies power to the VCA.

#### **4.1.4 Grasping Force Sensor Calibration: Prototype I**

A calibration experiment was conducted to evaluate the forceps tip sensor assembly prototype with respect to its linearity and repeatability. This is important in order to validate the force-amplifying capabilities of the overall system. The silicon strain gauge sensor is sensitive to light fluctuations, so the sensor assembly was shrouded and the lighting conditions were maintained constant to avoid incorrect measurements during the experiment. The sensor assembly was loaded against a force-torque (F/T) sensor (Nano17, ATI Industrial Automation, Inc., USA) whilst the force output from the F/T sensor and the voltage output from the forceps tip sensor were acquired. The Nano17 has a resolution of 3.125mN for the calibration adopted in this study which imposes a limit on the minimum force that can be quantified using this calibration method. Three loading and unloading cycles were conducted over a range of 300mN and are shown in Figure 4.5. A line of best fit using the least squares method is fitted through this data. Table 4.1 provides a summary of the regression ratios and coefficients of determination for the individual loading and unloading curves.

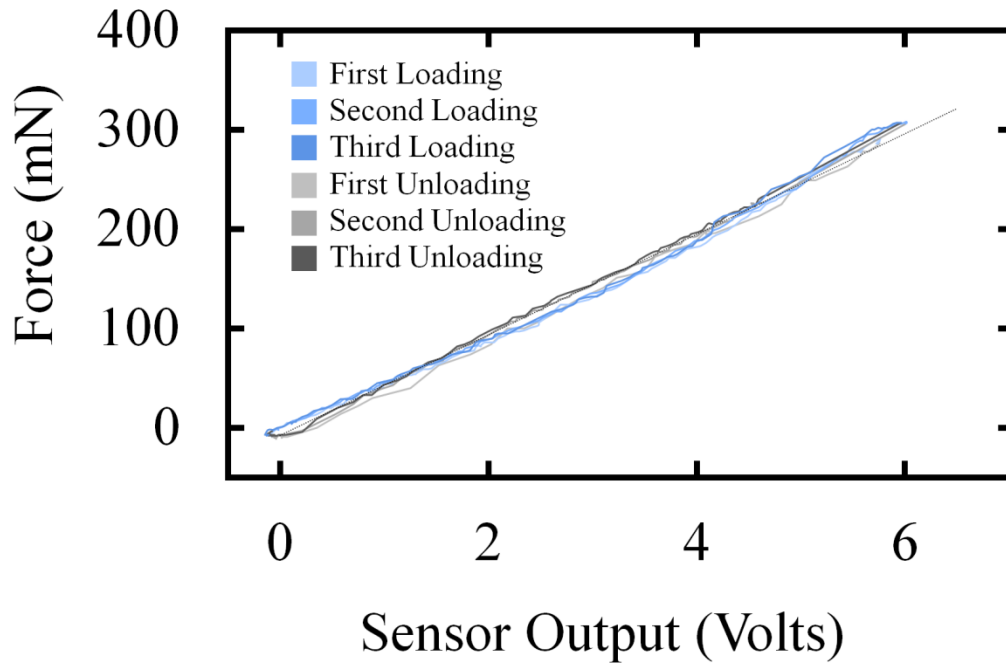


Figure 4.5: Calibration experiment showing the voltage output for a force input range of 0-300mN for three successive loading cycles of the force sensor prototype I, a linear regression for the entire data set is included on the plot.

Table 4.1: The linear regression models for the measured grasping force sensor assembly calibration for prototype 1.

|                     | Regression Ratio (mN/V) | Coefficient of Determination $R^2$ |
|---------------------|-------------------------|------------------------------------|
| First loading       | 50.94                   | 0.999                              |
| First unloading     | 50.69                   | 0.996                              |
| Second loading      | 50.77                   | 0.999                              |
| Second unloading    | 51.29                   | 0.994                              |
| Third loading       | 50.54                   | 0.999                              |
| Third unloading     | 50.46                   | 0.993                              |
| Median              | 50.73                   | 0.998                              |
| Interquartile range | 0.32                    | 0.00                               |

The results presented in Figure 4.5 and Table 4.1 indicate that the sensor assembly produces a repeatable and linear response over a range of 0-300mN. It can be observed that there is a slight hysteresis effect between loading and unloading cycles which is most likely due to the polymer-based material from which the force-sensor assembly is constructed from.

#### 4.1.5 Force amplification Bench Tests: Prototype I

The force-feedback forceps were evaluated in a bench test designed to evaluate the force-amplifying abilities of the system. Measurements of the forceps grasping force  $F_g$  and the fingertip force  $F_f$  were made as the device was loaded and unloaded for a series of scaling factors of 3, 6, 8, 15, 30, 50 and 80. The force calibration described in the previous section was adopted for the  $F_g$  measurements. The force-feedback sensor that makes measurements of  $F_f$  was calibrated against a precision weight. A spring was used to support the reaction forces generated by the VCA whilst the forceps tip sensor was loaded and unloaded. The results are presented in Figure 4.6 and Table 4.2. Lines of best fit have been added to the measured data for the range of scaling factors using the least squares method.

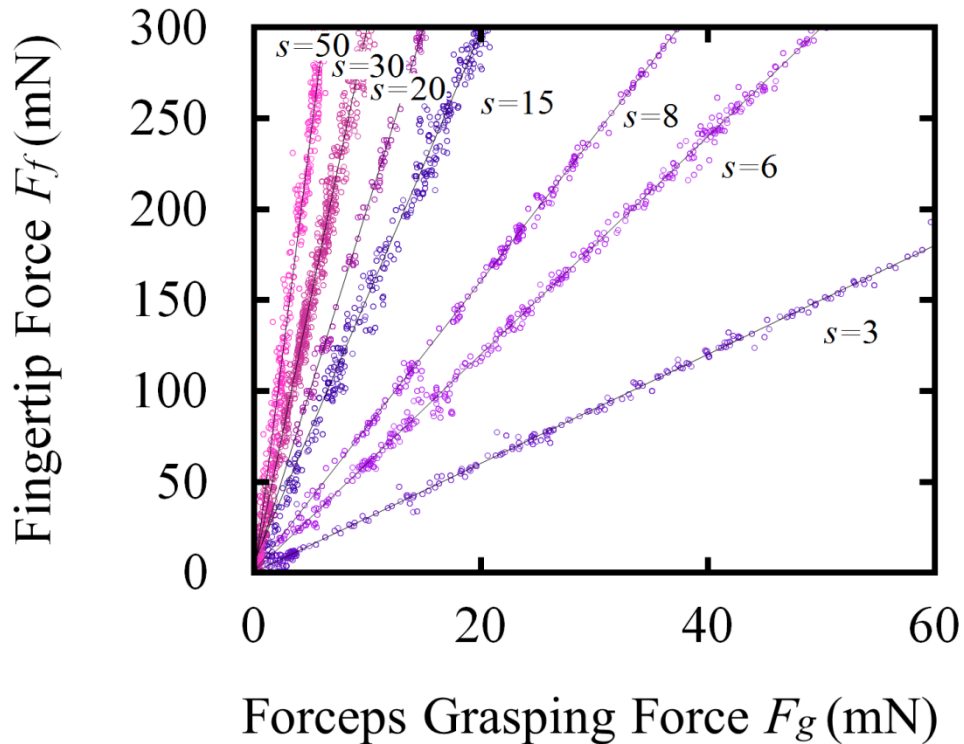


Figure 4.6: Plot showing the force-amplifying abilities of the force-feedback forceps for scaling factors of 3, 6, 8, 15, 30, 50 and 80 for prototype I.

**Table 4.2: Linear regression models for the measured scaling factors using the least squares method for prototype I.**

| Desired Scaling Factor | Regression Ratio | Residual Force (mN) |
|------------------------|------------------|---------------------|
| 3                      | 3.0              | 0.18                |
| 6                      | 6.0              | 0.28                |
| 8                      | 8.0              | 0.69                |
| 15                     | 14.6             | 5.71                |
| 20                     | 19.9             | 2.06                |
| 30                     | 30.0             | 1.43                |
| 50                     | 48.9             | -2.65               |

These results illustrate the force-amplifying abilities of the force-feedback forceps. The data captured only considers the linear region in which force amplification occurs, neglecting the discontinuities introduced by the VCA saturating at its peak power consumption. The force-feedback forceps can retain controlled force amplification at all measured scaling factors. Whilst the VCA can generate forces greater than the 300mN range, this is the most significant region of interest as it represents the perceptible force levels displayed to the fingertip (in the order of  $\times 10^2$  mN as reported in Chapter 3) against the otherwise imperceptible force levels exerted by the grasping forceps ( $< 10$  mN [90]).

## 4.2 Hand-Held Force-Feedback Forceps: Prototype II

Following the development of a first hand-held microsurgical forceps prototype, a second prototype was devised and evaluated. This second prototype was developed to address some of the shortcomings seen in the first design. Rather than retrofitting hardware to existing forceps designs, a completely customised design was produced in order to meet the ergonomic and miniaturisation requirements of such a device. Whereas in the first prototype, the actuation is mounted on to one side of the forceps, in the second prototype, the actuator is mounted in between the forceps cantilevers. This architecture has ergonomic benefits as the weight is symmetrically distributed and the device bulkiness is reduced. Furthermore, in the proposed arrangement, the actuator is mechanically isolated from the elastic cantilevers so as to completely avoid the potential for control instability from coupling between the sensor and actuator. The second prototype also adopts an open-loop force-feedback control scheme which avoids the necessity for a feedback force sensor to measure  $F_f$ . This design choice was made to enable miniaturisation of the device and to improve the system stability. A T-profiled structure transmits the loads generated by the actuator on to the operator's fingertip

in the perpendicular direction to the grasping force. As in the previous design incarnation, this method of force-feedback ensures that the operator can distinguish the rendered force-feedback from the elastic cantilevers spring-return of the forceps. The total weight of the device is 22g, less than half the weight of prototype I. The device is shown in Figure 4.7.

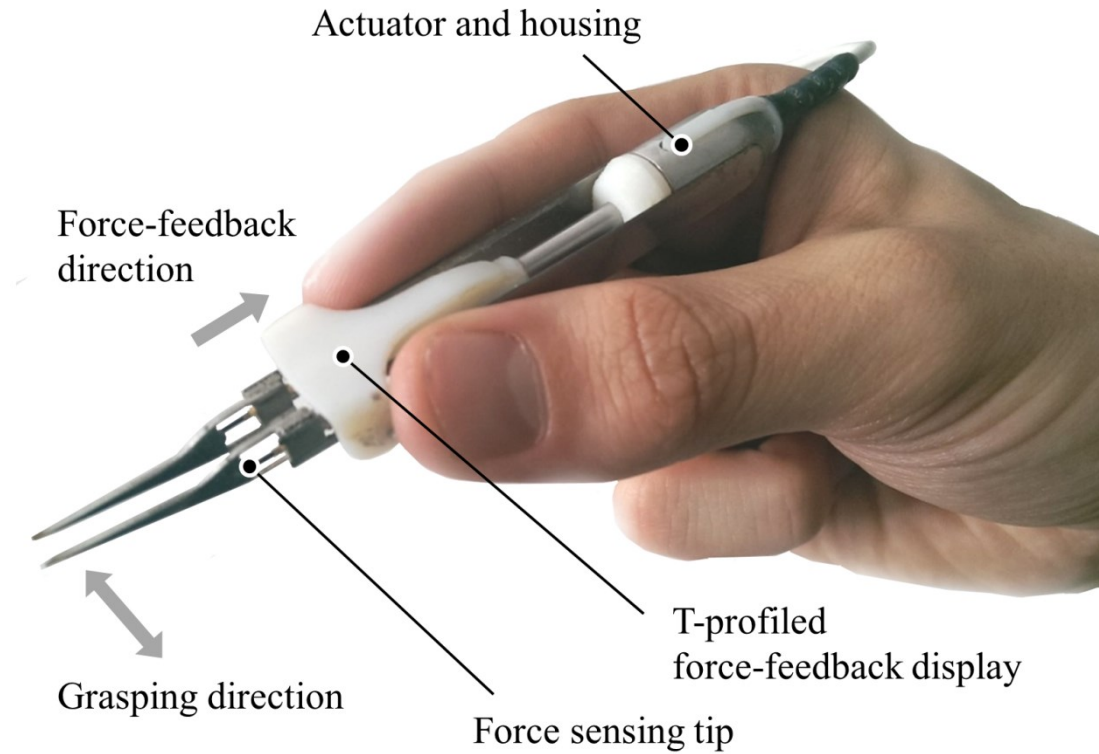


**Figure 4.7: Photograph of the microsurgical force-feedback forceps prototype II.**

#### **4.2.1 Grasping Force Sensor Development: Prototype II**

A new grasping force sensor design assembly was required for the second incarnation of the force-feedback forceps. The key requirement was to produce a *miniaturised* forceps tip design akin to existing microsurgical forceps with improved rigidity and robustness. The tips are curved through an angle of  $15^\circ$  so as to allow the surgeon to hold forceps at a natural angle, as per existing designs. The same design goal of a 0-300mN sensing range was pursued in order to demonstrate effective force sensing at sub-threshold force levels. The new design proposed the same silicon strain gauge sensor (AE801, Kronex, Oakland, USA) as adopted in the first prototype. The same design architecture was also proposed in which the silicon sensor beam is coupled to the forceps tip section. This allows a proportion of the

grasping load to be supported by additional cantilever features which increases the maximum grasping force load that can be applied by the forceps. The proposed forceps tip assembly design is presented in Figure 4.8.

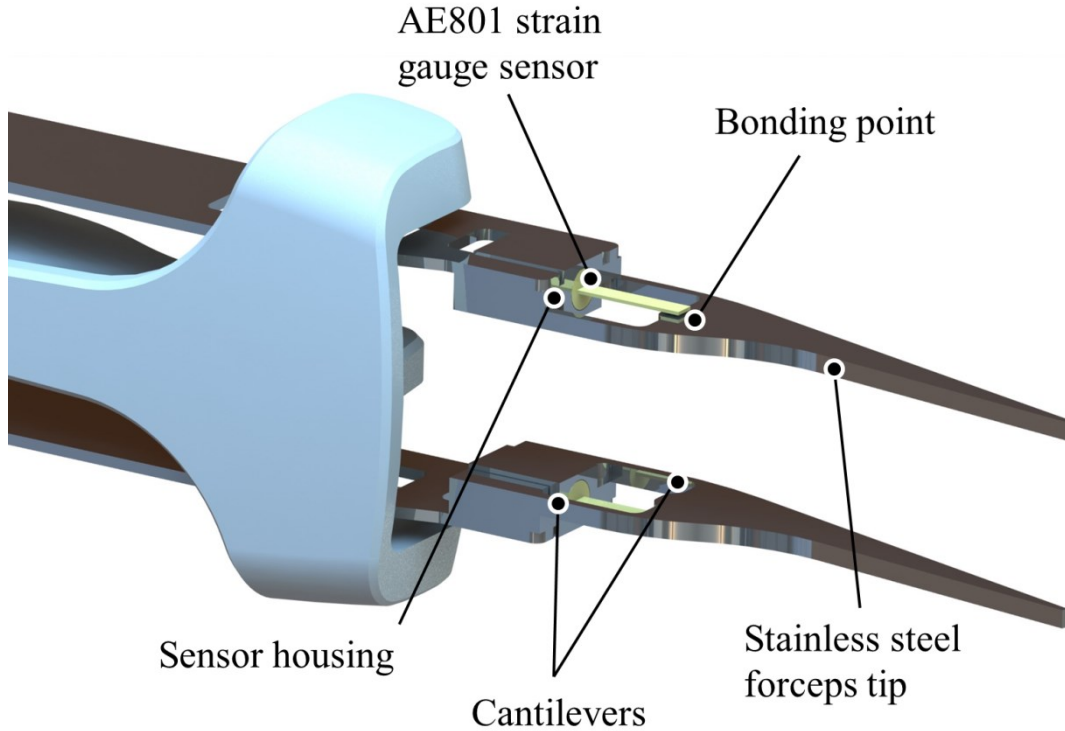


**Figure 4.8:** Detailed view of a CAD rendering showing the forceps tip assembly for prototype II.

As with prototype I, it is key that the forceps tip force sensing assembly can sense load at a sub-threshold scale (0-300mN) but that is also sufficiently robust to withstand higher loading too. For example, Harada *et al* observed grasping forceps of up to 5N during simulated microsurgical procedures [91]. The previous incarnation of the grasping force sensor assembly was designed empirically through iterative tuning of the mechanical cantilever configuration using rapid prototyping. Prototype II was developed using FE simulations that were implemented to predict the proportion of strain transmitted to the force sensor which has a peak cantilever load carrying-capability of 144mN (Appendix I). Many parameters of the forceps tip design can be altered to achieve a compromise between sensor sensitivity and robustness. However the chosen parameter to vary was the profile thickness which has a

significantly greater bearing on the mechanical rigidity of the cantilever features in the forceps tip profile design.

Four FE simulations were performed to assess the effectiveness of the forceps tip assembly designs with respect to robustness and sensitivity. An initial simulation was performed to assess the intrinsic mechanical parameters of the AE801 force sensor as a reference. Subsequently, three forceps tips designs of the same profile but with plate thicknesses of 0.5mm, 0.7mm and 0.9mm were simulated for comparison. These FE simulations were formulated and solved using Ansys (Ansys Inc, Cecil Township, PA, USA). The AE801 sensor was modelled as a beam of 4mm in length which is the distance between the mounting head base and the sensor tip (see Appendix I). For the first simulation, the sensor geometry was imported in Initial Graphics Exchange Specification (IGES) format. Two hard points were attributed to the imported geometry so as to allow Ansys to seed coincident nodes during the meshing phase. These hard points were located at the centre of the resistor location (inset 0.6mm from the mounting head point) so as to capture the average strain experienced by the entire resistor element. An automated meshing procedure was used to create a mesh comprised of 541 tetrahedral, 10-node elements (SOLID187).

The AE801 sensor beam element is manufactured from single crystal N-type silicon and thus, its mechanical stiffness is dependent on the silicon crystal orientation. The silicon sensor beam was thus modelled as a linear, orthotropic material using the elasticity matrix provided by Hopcroft *et al* [152], see equation 4.2 (all values in GPa). The symbols  $\sigma$  and  $\epsilon$  relate to stress and strain respectively, the subscript denotes the direction. The corresponding elasticity values used in the FE simulation are given in Table 4.3.

$$\begin{bmatrix} \sigma_{xx} \\ \sigma_{yy} \\ \sigma_{zz} \\ \sigma_{yz} \\ \sigma_{zx} \\ \sigma_{xy} \end{bmatrix} = \begin{bmatrix} 194.5 & 35.7 & 64.1 & 0 & 0 & 0 \\ 35.7 & 165.7 & 64.1 & 0 & 0 & 0 \\ 64.1 & 64.1 & 165.7 & 0 & 0 & 0 \\ 0 & 0 & 0 & 79.6 & 0 & 0 \\ 0 & 0 & 0 & 0 & 79.6 & 0 \\ 0 & 0 & 0 & 0 & 0 & 50.9 \end{bmatrix} \begin{bmatrix} \epsilon_{xx} \\ \epsilon_{yy} \\ \epsilon_{zz} \\ \epsilon_{yz} \\ \epsilon_{zx} \\ \epsilon_{xy} \end{bmatrix} \quad 4.2$$

**Table 4.3: The elasticity values for orthotropic silicon of a standard (100) wafer [151].**

| Elasticity Parameter | Parameter Symbol | Value   |
|----------------------|------------------|---------|
| Young's modulus (x)  | $E_x$            | 169 GPa |
| Young's modulus (y)  | $E_y$            | 169 GPa |
| Young's modulus (z)  | $E_z$            | 130 GPa |
| Poisson's ratio (yz) | $\nu_{yz}$       | 0.360   |
| Poisson's ratio (zx) | $\nu_{zx}$       | 0.280   |
| Poisson's ratio (xy) | $\nu_{xy}$       | 0.064   |
| Shear modulus (yz)   | $G_{yz}$         | 79.6GPa |
| Shear modulus (zx)   | $G_{zx}$         | 79.6GPa |
| Shear modulus (xy)   | $G_{xy}$         | 50.9GPa |

The AE801 sensor elements are cut from a “100 wafer” of monocrystalline silicon: an explanation of how the crystal orientation was determined for the AE801 is provided in Appendix II. A local coordinate system was attached to the sensing element according to the crystal orientation of the AE801 sensor, denoted in Figure 4.9, and the orthotropic elasticity values reported in Table 4.3 were assigned to the elements representing the sensor beam.

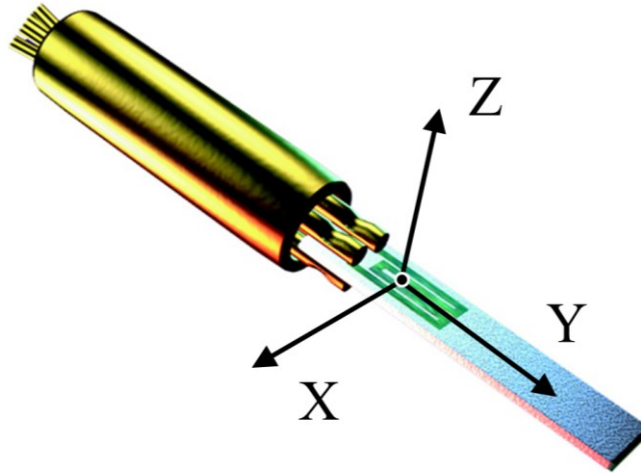


**Figure 4.9: The coordinate system assigned to the AE801 sensing element for the orthotropic elastic material properties assigned to the mesh model. Image provided, with permission from Kronex Technologies Corporation (Oakland, USA).**

Zero displacement boundary conditions were applied to the base of the sensor beam to approximate the sensor mounting head. A 144mN load was applied to a node at the cantilever tip which represents the peak loading condition described by the datasheet.

For the additional analyses, the sensor geometry and forceps tip profile designs were imported in IGES format for all three cases. For each analysis, the same meshing procedure was performed on both the sensor and forceps tip profile geometry, the number of elements generated for each FE model is reported in Table 4.4. The silicon strain gauge was assigned the same orthotropic material parameters and coordinate system as previously described. Each forceps tip profile was modelled as a linear, isotropic material of Young's modulus 193GPa and Poisson's ratio 0.3 in order to represent stainless steel. Zero displacement boundary conditions were applied to the base of the sensor beam and the forceps tip base. The AE801 sensor beam and forceps tip component were rigidly coupled together to represent a bonded joint. A 300mN load was then applied to a node at the forceps tip to represent  $F_g$ , the grasping force. These boundary conditions are presented in Figure 4.10. All of the simulations were performed in 10 sub steps and the Y-direction strain ( $\epsilon_y$ ) at the hard point nodes were recorded for each step. The peak deflection (displacement vector sum) for each analysis was also recorded.

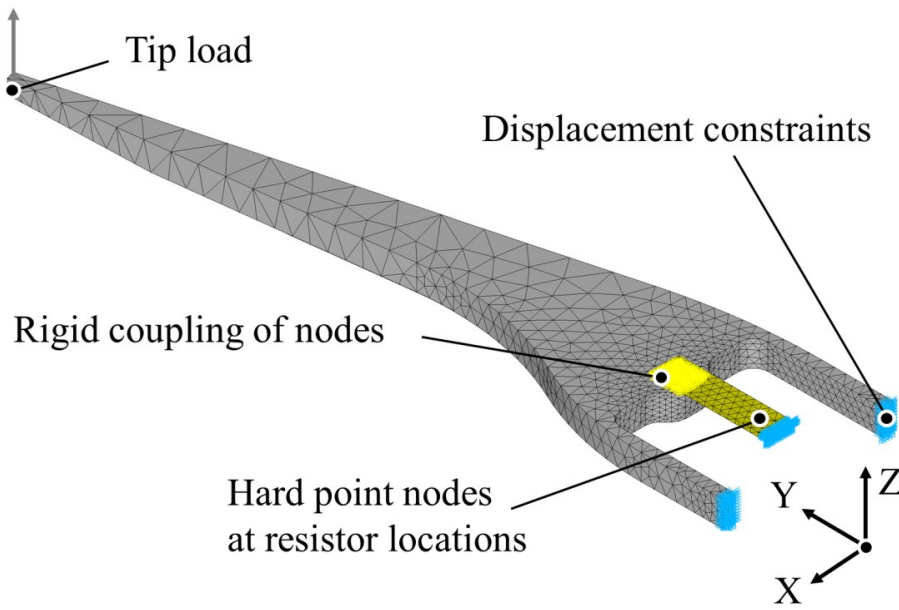


**Figure 4.10: The FE mesh and boundary conditions for the forceps tip force sensor assembly.**

For the initial FE simulation of the AE801 sensor alone, a peak deflection of  $65.6\mu\text{m}$  was reported which is consistent with the  $60\text{-}80\mu\text{m}$  figures quoted in the AE801 datasheet (refer to Appendix I). The strain at the both resistor locations was found to be  $600\mu\text{strain}$  and  $-600\mu\text{strain}$ . These peak strain magnitudes were used to determine safety factors and failure loads for the three subsequent simulations of the forceps tip assemblies. The peak deflections, resistor strains, safety factors (at  $300\text{mN}$  load) and failure loads are reported for the three forceps tip profiles in Table 4.4. An example contour plot of the displacement vector sum for the  $0.9\text{mm}$  case is provided in Figure 4.11. Example contour plots of the strain  $\varepsilon_y$  for  $0.9\text{mm}$  case are provided in Figure 4.12 and Figure 4.13. Figure 4.14 plots the relationship between  $\varepsilon_y$  and forceps tip load.

**Table 4.4: Results from the FE simulations for the 0.5mm, 0.7mm and 0.9mm thickness tip profiles.\*Based on the higher compressive strain for each case.**

| Parameter                                                         | 0.9mm | 0.7mm | 0.5mm  |
|-------------------------------------------------------------------|-------|-------|--------|
| Number of elements in FE model                                    | 2844  | 2464  | 2254   |
| Peak displacement vector sum ( $\mu\text{m}$ )                    | 34.4  | 72.3  | 194.5  |
| $\varepsilon_y$ of resistor in tension ( $\mu\text{strain}$ )     | 22.0  | 52.2  | 125.2  |
| $\varepsilon_y$ of resistor in compression ( $\mu\text{strain}$ ) | -47.0 | -79.3 | -140.9 |
| Total strain difference ( $\mu\text{strain}$ )                    | 69.0  | 131.5 | 266.1  |
| Safety factor at 300mN tip load*                                  | 12.8  | 7.6   | 4.3    |
| Grasping force failure load* (N)                                  | 3.8   | 2.3   | 1.3    |

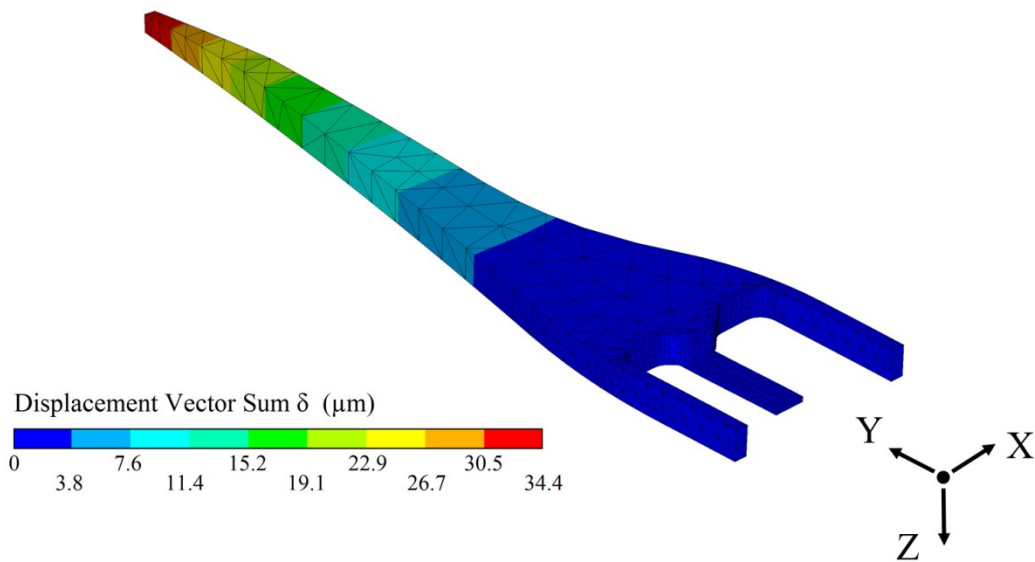


**Figure 4.11: The displacement vector sum of the force sensing tip assembly (0.9mm thick model) in response to a 300mN tip load.**

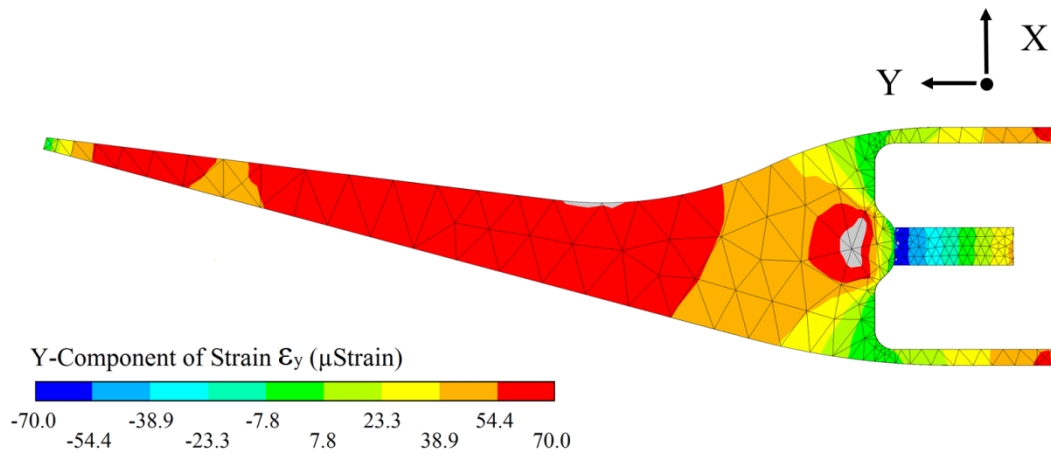


**Figure 4.12:** The Y-component of strain of the force sensing tip assembly (0.9mm thick model) in response to a 300mN tip load (view of the inner side that is undergoing predominantly tensile strain).

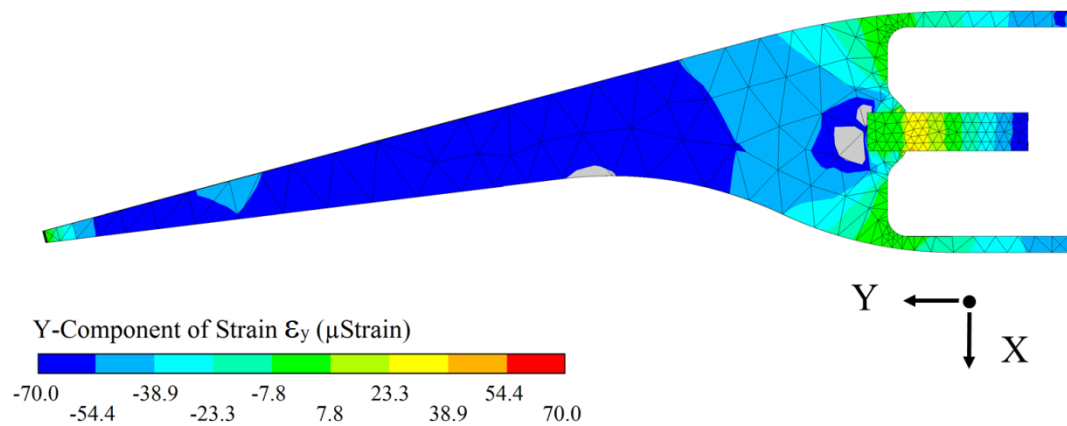


**Figure 4.13:** The Y-component of strain of the force sensing tip assembly (0.9mm thick model) in response to a 300mN tip load (view of the outer side that is undergoing predominantly compressive strain).

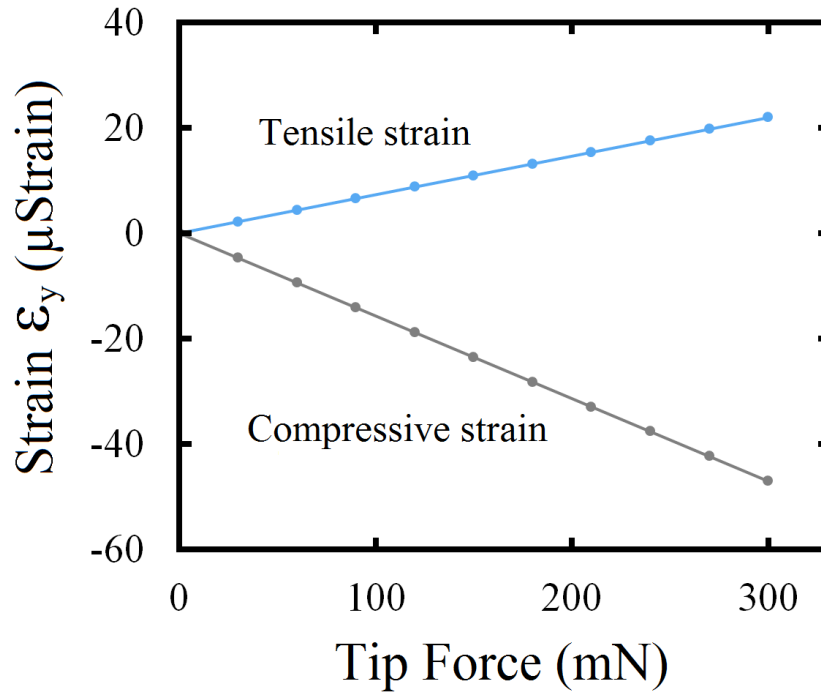


**Figure 4.14:** The predicted strains at the resistor locations with respect to tip load for the forceps tip force sensor assembly (0.9mm thick model).

The forceps tip assembly FE simulations demonstrate a consistently lower magnitude tensile strain compared to the compressive strain for each case (best observed in Figure 4.14). This is a result of the tip load not being applied about the neutral plane of the sensor beam. The end load is instead offset by the thickness of the forceps tip component. This asymmetrical loading is greater for the thicker forceps tips, in which the disparity in bending strain between resistor locations is greater. The response is however predicted to be linear and ultimately does not affect the sensor performance because it is the difference in strain that can be sensed through the Wheatstone half-bridge.

The safety factor calculations compared the resistor strain of the AE801 sensor at peak loading with the highest (compressive) resistor strain for each forceps tip assembly. All of the proposed designs were shown to operate within a safety margin for a 300mN tip load. The lowest safety factor (for the 0.5mm thick forceps tip) was 4.3. A calculation was made to extrapolate the forceps tip load that would induce a compressive strain in the force sensor equivalent to the peak strain that the AE801 can tolerate. The results favour the 0.9mm thick

forceps tip which can theoretically permit grasping forces of up to 3.8N before failure. The 0.9mm thick design was subsequently selected for prototyping: it demonstrates a total strain difference of 69.0  $\mu$ strain which corresponds to a 3.8m $\Omega$  -5.3m $\Omega$  change in resistance which can be sensed through the use of a Wheatstone bridge and amplified. The results also show the assembly to be sufficiently stiff: the tip deflection for 0.9mm case is 34.4 $\mu$ m which is negligible in comparison to the deflection of the forceps. There were a number of limitations associated with the FE simulations performed. The boundary conditions were simplifications of the real-world scenario, the AE801 sensor elements are mounted in to a header assembly which is not perfectly rigid. Similarly, the coupling of the silicon sensor element to the stainless steel forceps tip would be made using an adhesive which would have non-rigid mechanical properties that weren't simulated in this analysis. These limitations, whilst not likely to be significant, were further justification for the choice to prototype the stiffer, more robust forceps tip assembly from the 0.9mm thick stainless steel sheet metal.

Following the FE simulations, a prototype assembly was fabricated; Figure 4.15 and Figure 4.16 show details of this assembly. This tip section was fabricated from stainless steel, in accordance with existing microsurgical designs and also with the goal of producing less hysteresis with cyclical loading. Laser micromachining was used to form the tip profiles. The AE801 silicon beam gauge is mounted within a stainless steel housing that was machined using computer numerical control (CNC) techniques. These housings are welded to the elastic cantilevers of the forceps. The AE801 sensor was bonded to the steel housing and to the steel forceps tip component using a cyanoacrylate adhesive.

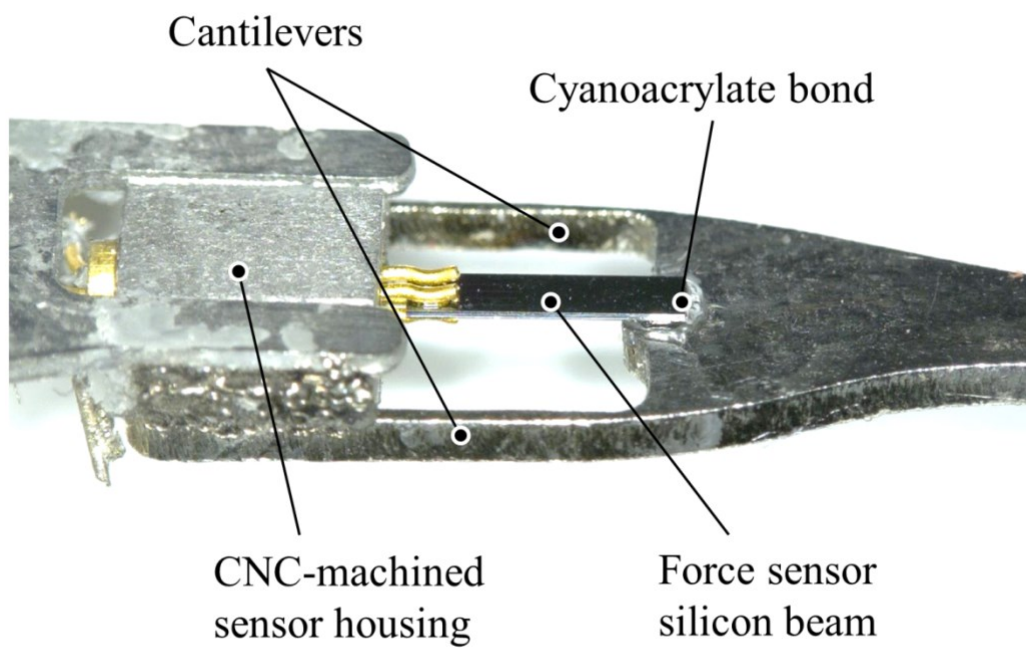


**Figure 4.15:** Detail of the forceps force sensor assembly for the force-feedback microsurgical forceps prototype II.



**Figure 4.16:** Detail of the forceps force sensor assembly for the second force-feedback microsurgical forceps prototype II.

#### **4.2.2 Mechatronics Design: Prototype II**

The forceps design is comprised of two elastic cantilevers that are bonded at their proximal region so that they can be clasped together in the usual manner that forceps are manipulated. The forceps tips have a total grasping stroke of 7.4mm. The elastic cantilevers of the forceps are manufactured from 0.5mm thick stainless steel using a photo etching process. At the proximal region of the elastic cantilevers is a CNC-machined stainless steel housing for the force-feedback actuator, a VCA (LVCM-010-013-01, Moticont). The VCA is 9.5mm in diameter, it has a length of 20.7mm and can provide up to 0.28N of continuous force. This actuator was chosen for its compact size and low weight (6.3g). The continuous force delivery is comparatively low in comparison to prototype I and the force-amplifying device proposed in Chapter 3, however it can produce loads that are easily perceived and the lower reaction forces generated avoid recoiling of the device during hand-held operation. The mounting of the VCA in between the elastic cantilevers ensures the device is balanced and provides a means for effective force-transmission of the force-feedback display. The force generated by the VCA is transmitted through a shaft that is supported by a low friction bearing and is coupled to a T-profiled force-feedback display which makes contact with the operator's fingertip. This force-feedback display is seated in slots that are formed in the elastic cantilevers which constrain its motion to be purely axial. The forceps tip assemblies described previously are welded at the distal region of elastic cantilevers and were manufactured so that the tips made contact as the forceps are squeezed by the operator. Only a single forceps tip force sensor was adopted in this prototype. The force sensor and VCA wires were routed through the housing and in to a cable assembly with strain relief. The device itself is 125mm in length. These design features are illustrated in Figure 4.17

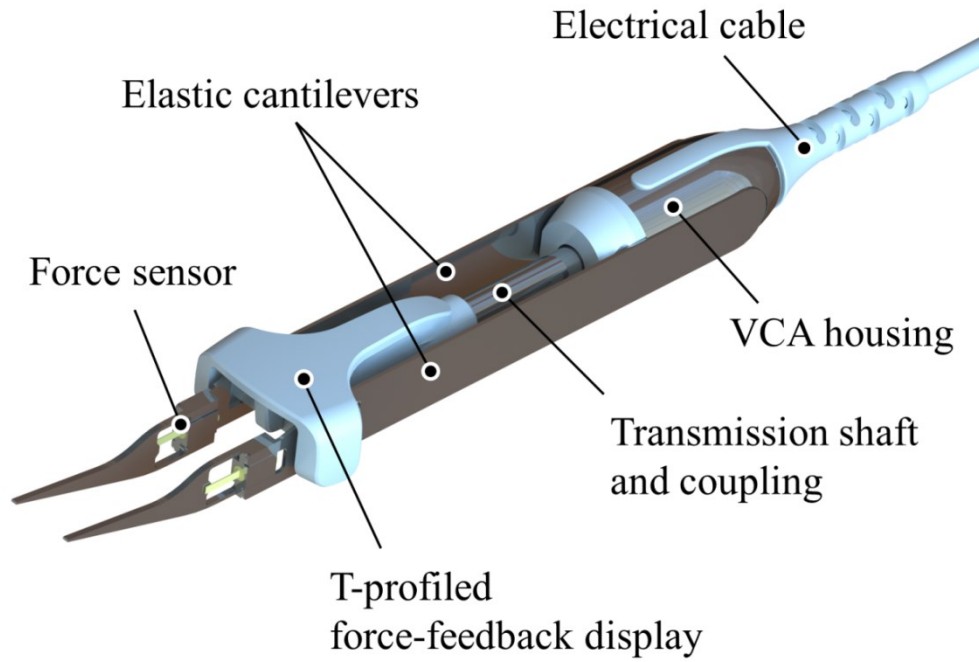


**Figure 4.17: CAD rendering of the force-feedback forceps prototype II.**

### 4.2.3 Control Implementation: Prototype II

An open-loop control scheme was implemented for the prototype II forceps on a multi-threaded, real-time controller (CompactRIO, cRIO-9022, National Instruments Corp.) that operates on a 1kHz loop. This control scheme exploits the linear relationship between the current flowing through the VCA ( $I_{VCA}$ ) and the corresponding force generated ( $F_{VCA}$ ). A force constant ( $k_{VCA}$ ) relates these quantities and an empirical determination of  $k_{VCA}$  for the VCA used in prototype II is provided in Appendix III. The grasping force sensor input voltage is acquired by the controller following amplification through a Wheatstone half-bridge before being converted to Newtons by a program written on the FPGA which makes use of the force sensor calibration constant previously determined. A desired feedback force  $F_f$  is calculated by multiplying the input force by the pre-set scaling factor. The FPGA program converts this force in to an input current through the VCA force constant determined in Appendix III. An input voltage is then determined using Ohm's law and the VCA coil resistance which is also empirically determined in Appendix III. This voltage is supplied by a linear amplifier (LSC 30/2 linear 4Q Servoamplifier, Maxon) which provides power to the VCA.

#### 4.2.4 Grasping Force Sensor Calibration: Prototype II

A calibration experiment was conducted to evaluate the second forceps tip sensor assembly with respect to linearity and repeatability. The previous calibration experiment used to evaluate the first prototype device adopted a F/T sensor (Nano17, ATI Industrial Automation, Inc., USA) to perform this task. To improve upon this calibration, a high sensitivity precision balance (HCB123, Highland™, UK) was used to obtain a force-voltage characterisation. The scales have a resolution of  $\pm 0.001\text{g}$  ( $\pm 9.81\mu\text{N}$ ) and repeatability of  $\pm 0.003\text{g}$  ( $\pm 29.4\mu\text{N}$ ). The precision balance was connected to a host PC through an RS-232 interface. The prototype force sensor signal was connected to a Wheatstone bridge, amplified and acquired by a real-time controller (CompactRIO, cRIO-9022, National Instruments Corp.). A LabVIEW program passed the digitised force sensor signal to the host PC and synchronised this signal with the data obtained from the balance output.

The sensing assembly was rigidly clamped to a linear stage so that it could be traversed on to the precision balance at a perpendicular angle. A mount was affixed to the balance that allowed load to be transmitted to the tip of the forceps force sensor assembly. The balance and force sensor were zeroed prior to calibration. The sensor assembly was shrouded and the lighting conditions were maintained constant to avoid erroneous sensors measurements. Three loading and unloading cycles were performed over a range of 300mN and are shown in Figure 4.18. Table 4.5 provides a summary of the regression ratios and coefficients of determination for the individual loading and unloading curves.

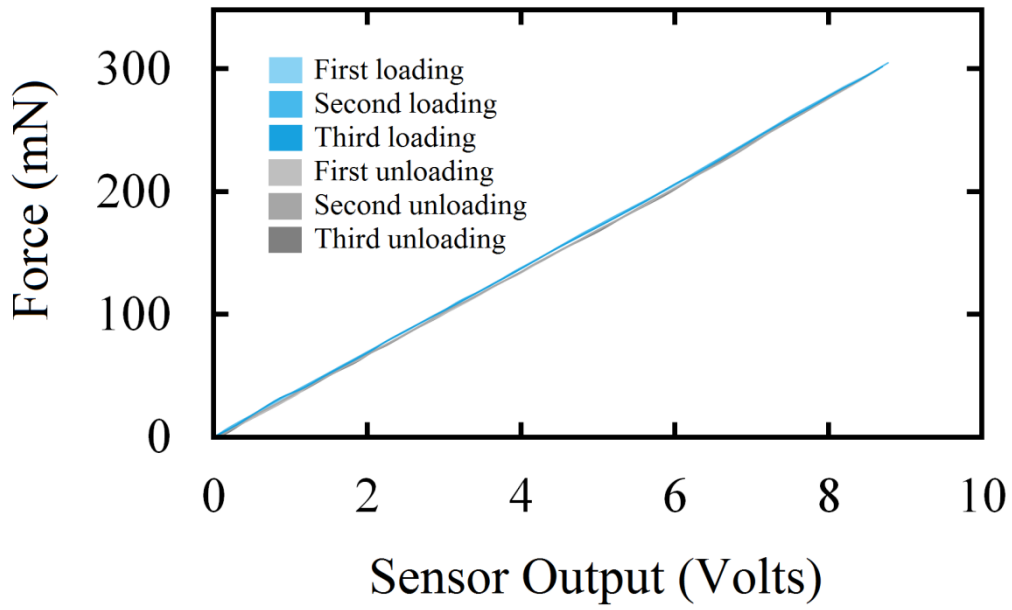


**Figure 4.18: Force sensor calibration plot for the prototype II.**

**Table 4.5: The linear regression models for the measured grasping force sensor assembly calibration for prototype II.**

|                     | Regression Ratio (mN/V) | Coefficient of Determination $R^2$ |
|---------------------|-------------------------|------------------------------------|
| First loading       | 34.62                   | 1.000                              |
| First unloading     | 34.57                   | 1.000                              |
| Second loading      | 34.59                   | 1.000                              |
| Second unloading    | 34.41                   | 1.000                              |
| Third loading       | 34.51                   | 1.000                              |
| Third unloading     | 34.51                   | 1.000                              |
| Median              | 34.54                   | 1.000                              |
| Interquartile range | 0.080                   | 0.000                              |

The results from the sensor calibration illustrate a considerable improvement in force-sensing fidelity compared to the first prototype. The sensor produces a repeatable and linear response over a range of 0-300mN with a median response of 34.54mN/V over the 300mN

loading range. The interquartile range between the observed regression ratios is 0.08mN/V which is 25% of the repeatability uncertainty observed in the first prototype calibration (0.32mN/V). Furthermore, the coefficient of determinations for each loading/unloading curve was found to be highly consistent, demonstrating excellent linearity and an improvement on the previous calibration. These improvements can be attributed to both the enhanced sensor construction and the calibration methodology. The use of stainless steel construction and rigid bonding of the force sensor is responsible for the improvements in hysteresis. There was considerably less noise observed compared to the calibration of the first prototype which is as a result of the higher resolution precision balance.

#### 4.2.5 Force amplification Bench Tests: Prototype II

The prototype II force-feedback forceps were evaluated in a bench test designed to evaluate the force-feedback abilities of the system. The prototype was rigidly affixed to a vibration-isolated optical table and the grasping force sensor was zeroed. A motorised linear stage was used to apply a gradual loading and unloading ramp to the force-sensing tip assembly at a constant velocity. The reaction forces generated by the VCA during the loading and unloading procedure were supported by a spring. Measurements of both the grasping force  $F_g$  and the voltage supplied to the VCA were made for each loading and unloading cycle. The voltage supplied to the VCA was converted to a theoretical current using Ohm's law and the VCA coil resistance ( $2.4\Omega$ , see Appendix III). The current supplied to the VCA is directly proportional to the force generated via the VCA force constant  $k_{VCA}$  (0.32N/A, see Appendix III) and thus a theoretical feedback force  $F_f$  was calculated for each loading/unloading run. The procedure was performed for seven scaling factors: 3, 6, 8, 15, 30 and 80. These results are presented in Figure 4.19 and Table 4.6.

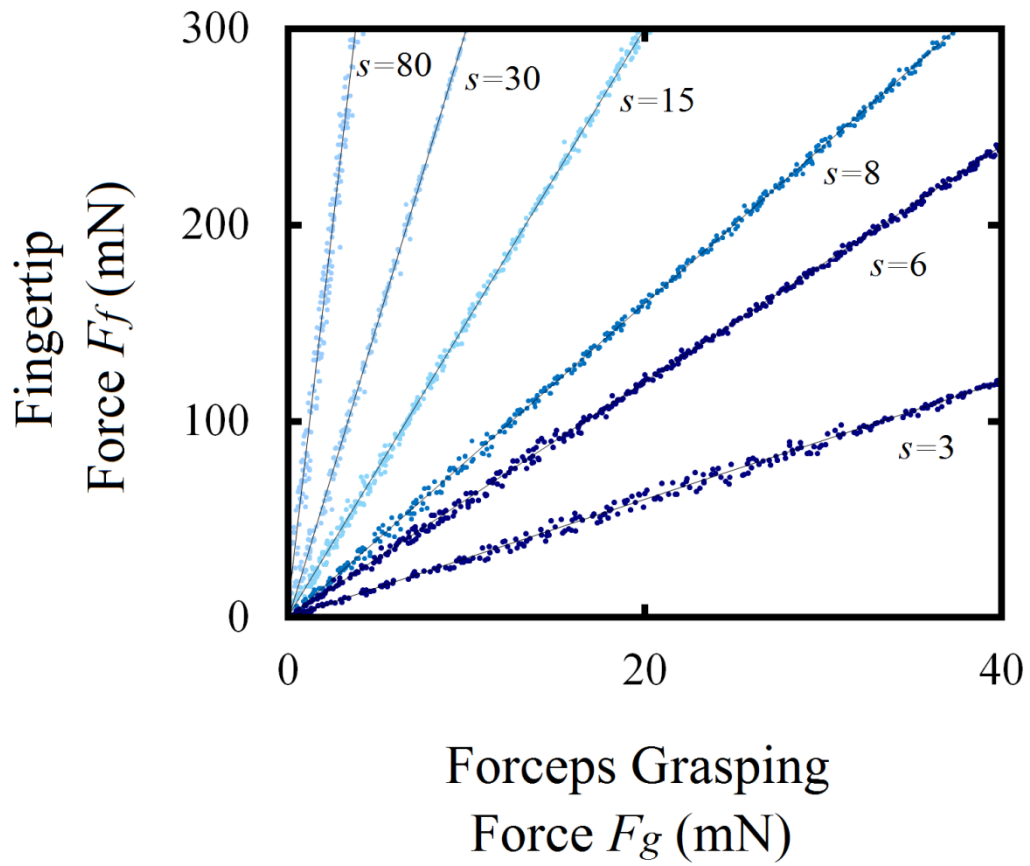


Figure 4.19: Plot showing the force-amplifying abilities of the force-feedback forceps for scaling factors through a range of 3, 6, 8, 15, 30 and 80.

Table 4.6: Linear regression models for the measured scaling factors using the least squares method.

| Desired Scaling Factor | Regression Ratio | Residual Force (mN) |
|------------------------|------------------|---------------------|
| 3                      | 3.0              | 0.2                 |
| 6                      | 6.0              | 0.2                 |
| 8                      | 8.0              | -0.3                |
| 15                     | 15.0             | 0.4                 |
| 30                     | 30.1             | -0.9                |
| 80                     | 77.8             | 5.5                 |

The results of this study demonstrate the force-amplifying abilities of the prototype II force-feedback forceps. Although not directly comparable to the equivalent experiment of prototype I, the improved sensor design ensures that the voltage supplied to the VCA is less noisy and that distinct scaling factors can be prescribed to the control system. The open-loop nature of the control implementation means that the actual force-amplifying ability of the device is likely to be less than what is reported in Figure 4.19 and Table 4.6 because there is no feedback sensor to compensate environmental errors such as friction. Nonetheless, the results obtained from the VCA force calibration provided in Appendix III show a reliable correlation between VCA coil current and force output.

#### **4.2.6 Dynamic Response Bench Test: Prototype II**

In order to assess the dynamic response of the actuation system and force-feedback mechanism, a bench test was performed. The force-feedback forceps prototype was rigidly affixed to a vibration-isolated optical table. The actuation assembly was coupled to a soft spring, designed to approximate the mechanical properties of a human fingertip. The spring was manufactured from a composite rubber-polymer (TangoBlack, VeroWhite) assembly using 3D prototyping (Objet260 Connex1, Stratasys, Israel). The soft spring was coupled to a force sensor (8438 5005, Burster) so as to facilitate a direct measurement of  $F_f$ . This sensor was acquired in to the real-time controller (CompactRIO, cRIO-9022, National Instruments Corp.) and calibrated against a precision mass. The VCA was then sinusoidally excited through a range of 17 frequencies in an incremental fashion from 0.1Hz-100Hz. The input voltage provided to the VCA corresponded to a 0 - 0.3N amplitude based on calibration against the force sensor used in this experiment. The output voltage signal provided to the VCA and the acquired force sensing signal were synchronised on the FPGA, buffered and passed to the host PC using a first-in, first-out (FIFO) system. This allowed a data acquisition rate of 1kHz to be obtained without causing a data overflow problem. A fast data acquisition was necessary in order to effectively capture the high frequency system response without experiencing aliasing of the signals. Following acquisition, the data was processed to obtain the peak-to-peak magnitude of theoretical input force provided by the VCA and the measured output force (the maximum peak-to-peak amplitude for each frequency increment). A plot showing the magnitude response as a function of frequency is provided in Figure 4.20. Tabulated data of this experiment is also provided in Table 4.7.

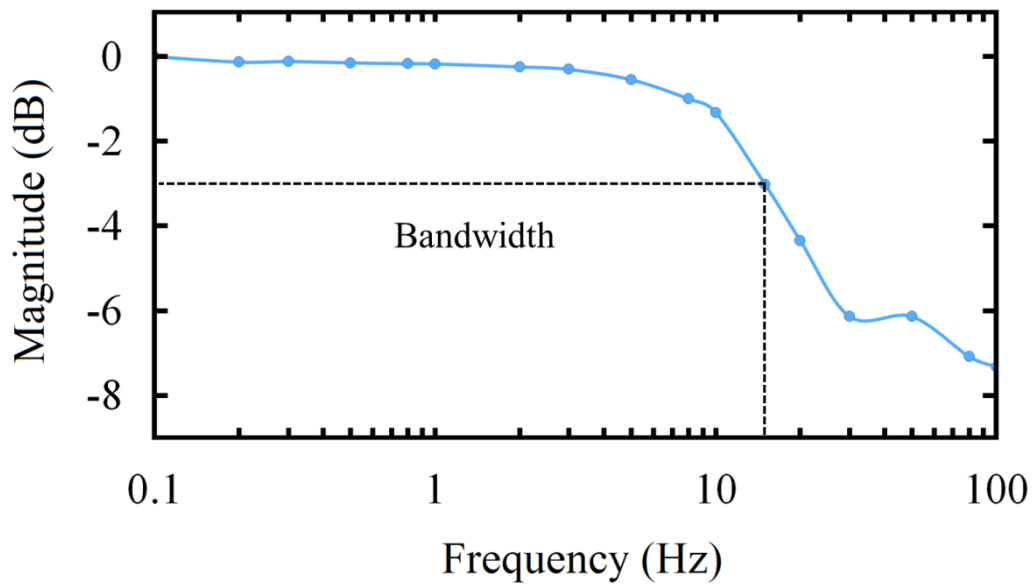


**Figure 4.20: The dynamic response of the force-feedback forceps.**

**Table 4.7: Tabulated data for the dynamic response bench test.**

| Frequency (Hz) | Peak-to-Peak Force Measured (N) | dB    |
|----------------|---------------------------------|-------|
| 0.1            | 0.300                           | 0.00  |
| 0.2            | 0.291                           | -0.13 |
| 0.3            | 0.292                           | -0.11 |
| 0.5            | 0.290                           | -0.15 |
| 0.8            | 0.289                           | -0.17 |
| 1.0            | 0.288                           | -0.17 |
| 2.0            | 0.284                           | -0.24 |
| 3.0            | 0.280                           | -0.30 |
| 5.0            | 0.265                           | -0.55 |
| 8.0            | 0.238                           | -1.00 |
| 10.0           | 0.221                           | -1.32 |
| 15.0           | 0.150                           | -3.02 |
| 20.0           | 0.110                           | -4.34 |
| 30.0           | 0.073                           | -6.14 |
| 50.0           | 0.073                           | -6.14 |
| 80.0           | 0.059                           | -7.09 |
| 100.0          | 0.056                           | -7.33 |

These results demonstrate that the actuation mechanism of the prototype II force-feedback forceps can generate forces over the frequency range that would be expected in hand-held manipulation tasks. The open-loop bandwidth is approximately 15Hz, with the VCA force output dropping by -3dB at this excitation frequency.

### 4.3 Force Detection Threshold User Study

A user study was conducted in order to evaluate the effectiveness of the force-feedback forceps in a grasping task with the hypothesis that force-feedback reduces the force threshold at which a user can detect grasping of an object. The prototype I force-feedback forceps were used in this study. The experiment compares this force-feedback forceps prototype against standard microsurgical forceps (S&T D-5A, Dumont Surgical, Gaithersburg, USA), the identical model to the forceps that were integrated into prototype I. These conventional forceps weigh 15g. Eight users (5 male and 3 female) with an age range of 24 to 33 years were recruited for this study and gave their informed consent to partake. All users were right-handed and none reported prior clinical experience. The experimental setup is shown in Figure 4.21 and Figure 4.22.

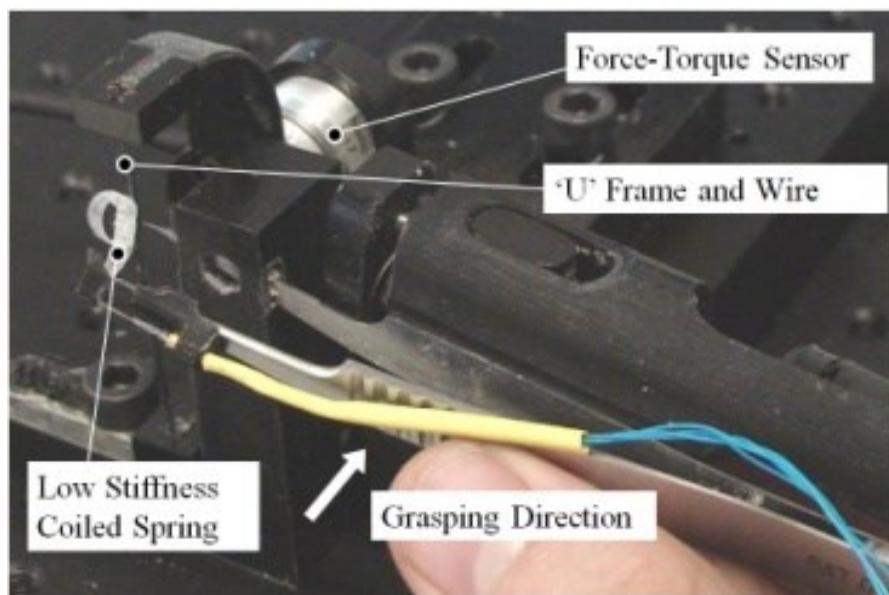


**Figure 4.21:** The experimental setup shown with the force-feedback forceps prototype; the grasping plane is denoted by the white arrow.

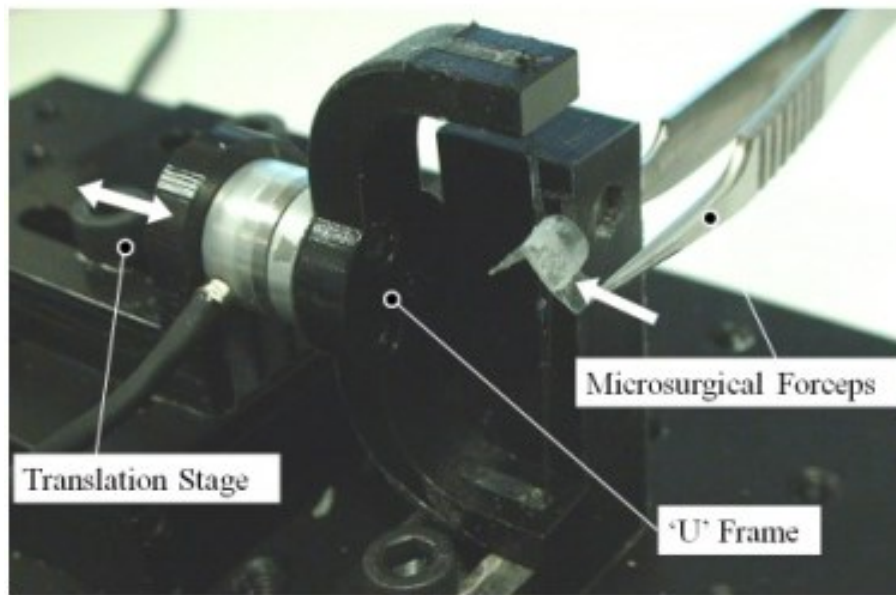


**Figure 4.22:** The experimental setup shown with the force-feedback forceps prototype; the grasping plane is denoted by the white arrow. The translation stage was used to randomise the grasping distance.

The forceps were arranged such that one elastic cantilever of the forceps was grounded whilst the other cantilever was free to be grasped by the users. A wire and low stiffness spring were mounted to a 'U' profiled frame so as to be orientated longitudinal to the grasping plane of the forceps and positioned so that contact would be made upon grasping of the forceps by the users. The 'U' profiled frame was coupled to a grounded F/T sensor (Nano17, ATI Industrial Automation, Inc., USA) so that an independent measurement of the forceps grasping force could be made. This arrangement was adopted to ensure the measurement of a bilateral grasping force, otherwise users could potentially detect contact resulting from only a single lever of the forceps. Since it was expected that the force-feedback forceps would be capable of determining considerably lower force thresholds, a low stiffness spring fabricated from polyurethane was placed in series with the wire so that the forceps tip would always make initial contact with the low stiffness spring. A spring of low stiffness was used, firstly to be representative of the soft tissue encountered in microsurgery and secondly, in order to minimise errors as a result of the reaction time between the users making contact and perceiving contact with the spring. Once the soft spring had been fully compressed the forceps would be resisted purely by the wire which was much stiffer, so as to exert up to around 2N at maximum compression of the forceps. This ensured that users would eventually perceive a force in addition to the spring return on

the forceps. The tips of the forceps for both cases were mounted so as to contact at the same point along the wire to ensure that it would give the same stiffness response for both experimental set-ups.

The 'U' profiled frame and F/T sensor were mounted on a translation stage that could translate in the direction of the forceps grasping plane. This allowed the position of the wire and low stiffness spring assembly (in the plane of the grasping direction) to be randomly altered to different positions over a range of 3mm offset from the tip of the forceps. This was done in an effort to reduce the user's proprioception sense which could help them learn when they would make grasping contact over the course of a trial. Users were prevented from observing the grasping task so as to rely solely on their sense of touch in detecting the grasping force.

The force data was acquired by the same multi-threaded, real-time controller (CompactRIO, cRIO-9022, National Instruments Corp.) that was used for the control of the instrument. The force data was captured at 50Hz and passed to a host PC. Once a user detected the force, they were requested to immediately stop grasping and depress a push button indicating that they had made contact. This push button generated a binary value which was also logged by the controller and synchronized with the force data. The local maxima of the force measurement, prior to the threshold logging, was taken to be the force detection threshold. The F/T sensor and forceps tip force sensor were zeroed before each user trial. The force-feedback forceps were shrouded and the lighting conditions were maintained constant to avoid incorrect sensor measurements. The experiment trials were repeated five times for both the conventional and force-feedback forceps. The order in which the trials were conducted was also randomised to prevent biasing of the results. A scaling factor of 30 was used in the trials in which the force-feedback forceps were used. The Mann-Whitney U test was performed to compare the force detection thresholds obtained with the force-feedback forceps and with conventional forceps, with a value of  $p < 0.05$  considered statistically significant. The results of the experiment are presented as a box plot in Figure 4.23 and a summary of the results are provided in Table 4.8.

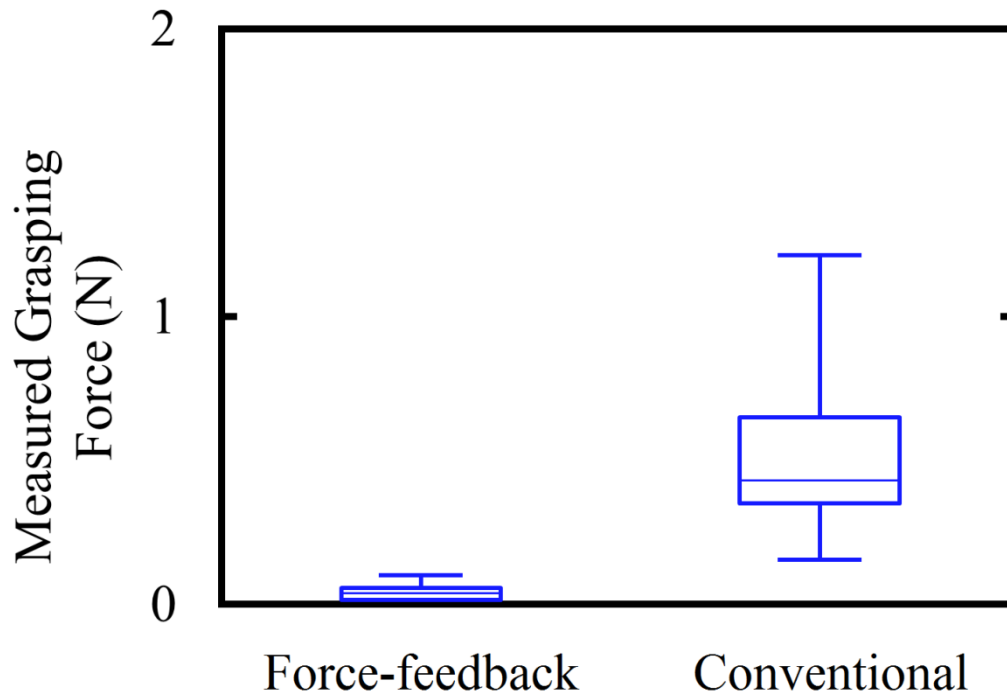


**Figure 4.23:** Box plot showing the distribution of measured grasping force thresholds for the 8 users in which force-feedback and conventional forceps were used in the experiment. Boxes indicate the lower quartile, upper quartile and median, whiskers represent the minimum and maximum values.

**Table 4.8:** Numerical statistical results from the force-detection user study.

|                                             | Force-feedback | Conventional |
|---------------------------------------------|----------------|--------------|
| Minimum force threshold (N)                 | 0.009          | 0.149        |
| Maximum force threshold (N)                 | 0.110          | 1.221        |
| Median force threshold (N)                  | 0.034          | 0.431        |
| Interquartile range of force thresholds (N) | 0.056          | 0.650        |

## 4.4 Discussion and Conclusions

The results of this study show that the inclusion of amplified force-feedback into microsurgical forceps significantly improves user perception of force, confirming the original hypothesis. The total distribution of force detection thresholds for both cases was

found to be statistically significant ( $p < 0.001$ ). Both the median and maximum force thresholds are approximately an order of magnitude higher when conventional forceps are used over the force-feedback prototype. This result can broadly be attributed to the spring return mechanism of the forceps, which hinder perception of the grasping forces. The force required to depress and completely close the tips of the spring-loaded forceps was measured to be 1.7N using the F/T sensor. Even for the grasping displacements that took place during the experiment, it is unsurprising that users could not detect the forces generated by the low stiffness spring that were considerably lower than the baseline forces required to depress the forceps. This is in accordance with the human perceptual response that is described by Weber's law.

All force thresholds measured from the experiment using the force-feedback forceps were made before the soft-spring was fully compressed compared to use of the conventional forceps in which users always fully compressed the soft-spring and made contact with the much stiffer wire before perceiving the grasping force. This explains the band separating the highest threshold measured with the force-feedback forceps and the lowest threshold measured with the conventional forceps. The distribution of force thresholds measured with conventional forceps was also greater than with the force-feedback device, with standard deviations of 256mN and 28mN respectively. This too can be attributed to both the 'filtering' effect of the low stiffness spring that ensured users could perceive force before making contact with the higher stiffness wire as well as randomization of the grasping distance which allowed variation in the total amount of force that could be exerted on to the wire.

The use of a higher stiffness wire will have led to errors in the force threshold measurements in the experiment using conventional forceps as a result of the user reaction times. This is because a large force would be generated over a small displacement which would give the users less time to respond to perceiving of the grasping force. Moreover, variations in the speed with which users closed the forceps will have also contributed to the variations in the force thresholds measured. There were also variations in the force threshold perceived between different users which can also be attributed to factors such as variations in the grasping speed as well as the intrinsic perception abilities of different users. However, there is considerably less inter-user variation for the experiment that evaluated the force-feedback

forceps, which is also in accordance with Weber's law. It is true that had the users been allowed to complete the study with visual feedback, they would have inevitably been able to perceive grasping contact at much lower force thresholds. However, in bimanual surgical tasks such as micro anastomosis, the surgeon cannot rely on visual feedback throughout if they are to avoid becoming cognitively overburdened.

This chapter has presented two novel microsurgical forceps prototypes that provide amplified force-feedback in the presence of a significant contaminating force. The method of haptic feedback presented has illustrated the importance of reliable and direct sensing of the surgical interaction forces as a necessity in bypassing other contaminating forces, such as weight, inertia and specifically in the case of microsurgical forceps, the elastic return force. The use of a customised, high resolution force sensor assembly has allowed ungrounded force scaling factors of up to 80 to be demonstrated. This represents a significant technical improvement upon the force sensing capabilities of the device reported in chapter 3. The first steps towards device miniaturisation were also made in this Chapter through the use of miniature actuators, sensing elements and customised fabrication. The force-feedback forceps concept was evaluated in a user study which validated that human subjects can detect grasping force at a lower threshold with force-feedback. Such a device has potential clinical applications in neurosurgery and vascular microsurgery when delicate anatomical structures must be handled by the operating surgeon. Excessive application of force can lead to catastrophic trauma to the patient, for example in the handling of eloquent brain tissue, aneurysms, and during vascular micro anastomosis.

Thus far, Chapters 3 and 4 have demonstrated that hand-held mechatronic force amplifying devices can improve human tactile perception of sub-threshold forces. However these haptic feedback strategies do not provide the operating surgeon with quantitative information relating to the intervention they are performing. Furthermore, the devices trialled so far have also yet to be deployed in a clinically-relevant context. Chapter 5 will address these outstanding issues through the demonstration of ungrounded vibrotactile force-threshold feedback in a hand-held device.

# **Chapter 5: Ungrounded Haptic Feedback using Force Limits for Neurosurgical Microdissection<sup>†</sup>**

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<sup>†</sup> Content from this chapter was published as:

**A Smart Haptic Hand-Held Device for Neurosurgical Microdissection**

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The previous chapters have investigated force amplification in hand-held devices and have demonstrated the use of this strategy in improving human tactile perception of tool-tissue contact. This concept has been postulated for deployment in clinical applications where tool-tissue forces are below what can be perceived by the surgeon. Examples of such applications include ophthalmic surgery and microvascular surgery. It is as yet, unclear as to whether such a force-feedback strategy might benefit clinical applications in which the surgical manipulation forces are supra-threshold. Furthermore, whilst the strategies examined in Chapters 3 and 4 have shown improvements in human perception, they cannot inform the surgeon as to when iatrogenic injury might occur, which is a significant goal of such haptic augmentation systems. A key contribution of this chapter is the implementation of haptic feedback in a hand-held device to warn the surgeon when a force threshold has been reached. This chapter also introduces the clinical exemplar of microneurosurgery as a means of illustrating this principle.

The performance of microneurosurgery requires extensive dexterity, precision and delicate force application, as well as a clear mental picture of the underlying anatomy. Speaking of cranial pathology, the eminent British surgeon Sir Percivall Pott declared in 1760 that they were “...*accompanied with more uncertainty, create more anxiety to the surgeon, and more hazard to the patient, than almost any other ill, to which the human frame is liable*” [153]. Although the outcome of patients undergoing cranial surgery has dramatically improved in the subsequent centuries, operations on the brain carry an inherent risk of significant morbidity and mortality. The human brain has a highly complex three-dimensional structure, is exquisitely prone to injury, and has very limited capacity for regeneration. It follows that surgeons performing microneurosurgery must learn to be both precise and delicate. The learning curve for microneurosurgery is long and arduous. Neurosurgeons garner their operative expertise through nearly a decade of training during residency and fellowship, and continue to refine their technique during independent practice. In recent years, limits to the hours worked by residents have reduced the caseload of surgical trainees, with surveys demonstrating a perception of worsened education and patient safety [154][155]. Improvements in the quality of training, including the use of simulation-based rehearsal, may mitigate the effects of resident duty hour restrictions [156]. Alongside these advances in resident education, the technological development of new surgical devices may play a role in reducing the learning curve, particularly for challenging procedures [157].

Among the most important skill residents must acquire during their microsurgical training is the judicious application of force when handling neurovascular tissue; sufficient force must be applied to successfully carry out microsurgical manoeuvres and economise movements, but undue force can result in tissue injury. Despite a great deal of anecdotal experience, quantitative data on the forces exerted during microsurgery remains sparse. Nonetheless, recent studies have confirmed that the range of forces exerted during microneurosurgery (typically less than 1 N) is an order of magnitude less than for general surgical procedures [158] [159]. Moreover, novices exert considerably more force than experts during microneurosurgery, and excessive forces are more likely to result in iatrogenic injury [91] [159].

The inherent subjectivity of human force perception makes it difficult for the less experienced surgeon to know when they are exerting force levels that could cause iatrogenic injury, and excess forces may only be recognised once irreversible injury has already occurred. Moreover, a multitude of factors can affect human force magnitude perception [160]. For example, in one psychophysical study, researchers showed that human subjects exert greater force with a rigid probe compared to when directly applying force with their fingertip [161]. One solution to this problem is to provide an indication to the surgeon as to when a force limit has been exceeded. Researchers have explored the use of auditory feedback to indicate force thresholds in microsurgical tasks where the force levels are below human tactile perception [139][140]. Vibrotactile force threshold indication has also been explored in a master-slave arrangement designed to enhance ultrasound-guided procedures in which visual feedback was diminished [99].

This chapter presents a force limiting hand-held haptic device for neurosurgical dissection. The device can indicate to the surgeon when a force threshold has been exceeded by providing vibrotactile feedback. The device design, engineering analyses and bench experiments are presented. In order to validate the vibrotactile-based force-threshold limiting concept, two psychophysical user studies were conducted to demonstrate how such an approach might limit the amount of force exerted by a surgeon. Furthermore, an *ex vivo*

cadaveric study was performed to demonstrate the potential clinical value of such an approach.

## **5.1 Hand-Held Smart Haptic Dissector**

The range of instruments used in cranial microsurgery has remained relatively unchanged since Yaşargil popularised such approaches [162]. Most microsurgical procedures entail a combination of sharp and blunt dissection using a series of dissectors and hooks. To this end, a blunt dissector was modified to incorporate sensors capable of measuring lateral forces applied to its tip. An on-board actuator then generates a subtle, but easily perceptible vibration when a force threshold has been exceeded. The device is presented in Figure 5.1.



**Figure 5.1: Photograph of the force-threshold vibrotactile feedback hand-held device.**

A blunt surgical dissector (Yaşargil FD304R, Braun Aesculap) was axially affixed to a profiled stainless steel shaft. Wire-based electro-discharge machining (EDM) was used to create two orthogonal beam profiles into the stainless steel shaft, as shown in Figure 5.2. Strain gauges (N11MA212011, RS Components) were bonded to either side of both beam

sections so as to enable independent measurements of the beam bending strains in each orthogonal direction.

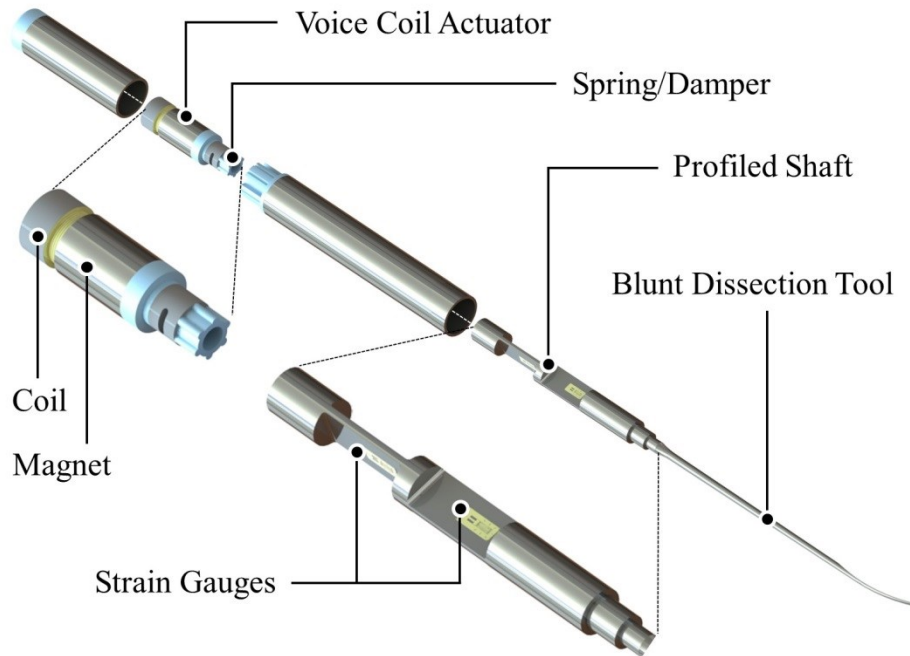


**Figure 5.2: CAD rendering of the force-threshold vibrotactile feedback hand-held device with exploded view illustrating the various sub-components.**

### 5.1.1 Finite Element Analyses of Strain Gauge Assembly

The dissector shaft is locally profined where the strain gauges are bonded in order to increase the local strain and improve sensing sensitivity. However, by locally thinning a region of the instrument shaft, its lateral rigidity is reduced which alters the ergonomics of the dissecting tool. Consequently, there is a trade-off between obtaining good sensor sensitivity without compromising the instrument rigidity significantly beyond the devices currently used in clinical practice. A number of analyses were performed in order to assess the trade-off prior to manufacture of the profined shaft component. The FE models were generated in order to predict the mechanical response to a perpendicular load at the dissector tip. The two parameters recorded were the total shaft deflection and the axial strain in the most proximal beam profile, these parameters permit an assessment of instrument rigidity and sensitivity respectively. Only loading in the bending plane of the proximal beam section was studied

because this represents the loading condition that would induce the greatest lateral deflection of the dissector. A coordinate system orientation for the dissecting tool was defined and is provided in Figure 5.3. The Z-axis is co-aligned with the tool axis and the X-axis is orientated in the plane of the dissector tool tip curvature. The most proximal beam section on the profiled shaft is then aligned to be perpendicular with the X-axis.

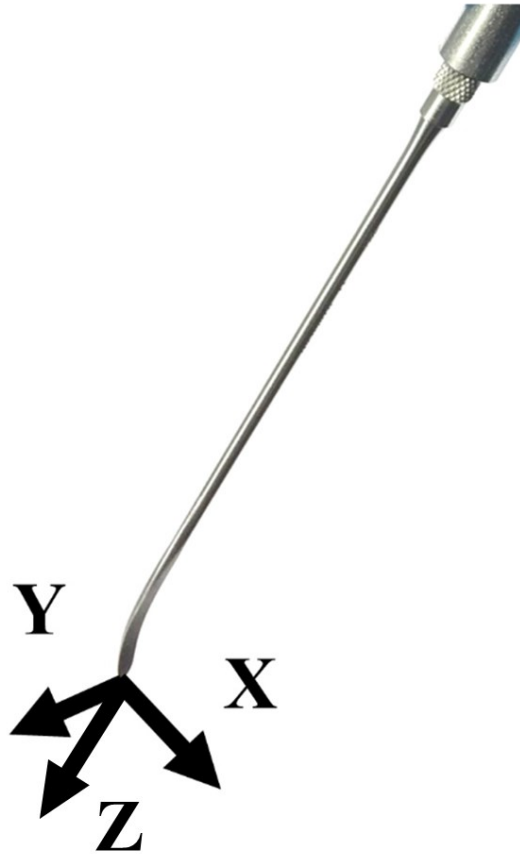


**Figure 5.3: The coordinate system for the dissecting tool.**

Seven solid models of the dissector shaft component were generated, an example of which is provided in Figure 5.4. One model represented a solid dissector shaft and the other six models incorporated beam profiles of differing thickness (2.0mm, 1.5mm, 1.2mm, 1.0mm, 0.8mm and 0.6mm, applied to both X and Y-axes). Each model was exported in Initial Graphics Exchange Specification (IGES) format and imported in to Ansys (Ansys Inc, Cecil Township, PA, USA). Two hard points were assigned to the imported geometry for each

model so as to allow Ansys to seed coincident nodes during the meshing phase. One hard point was located at the centre of the dissector tip surface and the second hard point was located at the centre of the proximal beam profile where one of the strain gauges would be bonded. These hard point locations are illustrated in Figure 5.4. Tetrahedral, 10-node meshes were automatically generated for each geometrical model using SOLID187 elements. The meshes were assigned to have isotropic, linear, elastic material properties of Young's modulus 193GPa and Poisson's ratio 0.3 in order to represent stainless steel. The nodes at the base of the dissector shaft mesh were constrained to have zero displacement so as to create a cantilever boundary condition. A load of 1N was applied to the hard point node at the instrument tip in the X-direction. An example mesh and boundary conditions is provided in Figure 5.4. For each simulation, the peak displacement sum ( $\delta$ ) and the strain in the Z-direction at the hard point node where the strain gauges are located ( $\varepsilon_z$ ) were recorded.

Figure 5.5 and Figure 5.6 are illustrative contour plots that denote the sum deflection and the Z-axis strain of the 1.2mm beam profile model in response to a 1N lateral load. Figure 5.7 plots the relationship between the elastic strain at the strain gauge bonding location with respect to the FE models of different beam thickness. Figure 5.8 plots the corresponding relationship between the peak deflection of the instrument shaft with respect to the FE models of different beam thickness. Table 5.1 reports these results and includes the peak deflection and strain for the equivalent un-profiled dissector shaft for comparison.

For each model, ratios were calculated to assess the rigidity-sensitivity trade-off. The strain ratio ( $\varepsilon_R$ ) compares the  $\varepsilon_z$  strain of the dissector shaft for each beam profile with respect to the un-profiled shaft strain ( $\varepsilon_{z1}$ ) so that:

$$\varepsilon_R = \frac{\varepsilon_z}{\varepsilon_{z1}} \quad 5.1$$

Similarly, the deflection ratio ( $\delta_R$ ) is the sum peak deflection for each profiled beam case ( $\delta$ ) with respect to the un-profiled dissector shaft sum peak deflection ( $\delta_1$ ), which is defined:

$$\delta_R = \frac{\delta}{\delta_1} \quad 5.2$$

These ratios  $\varepsilon_R$  and  $\delta_R$  can be combined to form another non-dimensional parameter that was used to assess the optimum point between sensor sensitivity and instrument rigidity:

$$\frac{\varepsilon_R}{\delta_R} \tag{5.3}$$

Figure 5.9 plots the ratio of relative strain increase and relative deflection increase as a function of profile beam thickness.

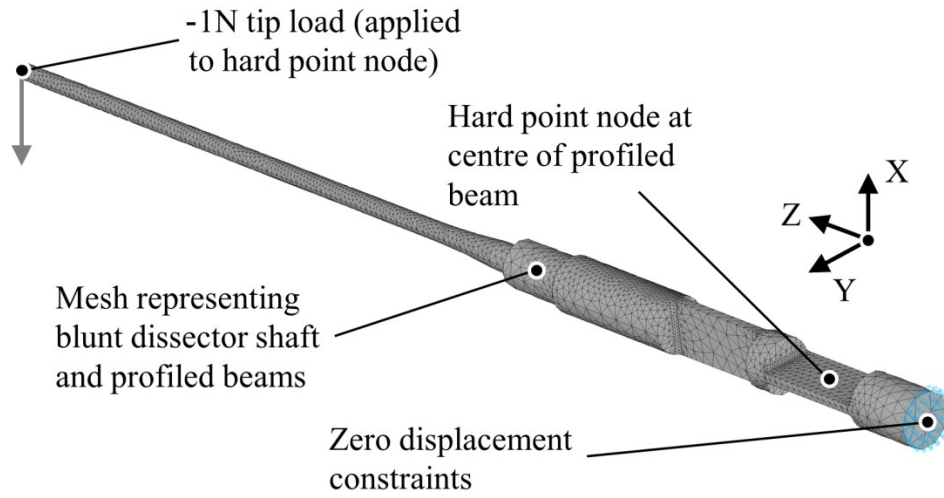


Figure 5.4: FE model denoting the loading and boundary conditions.

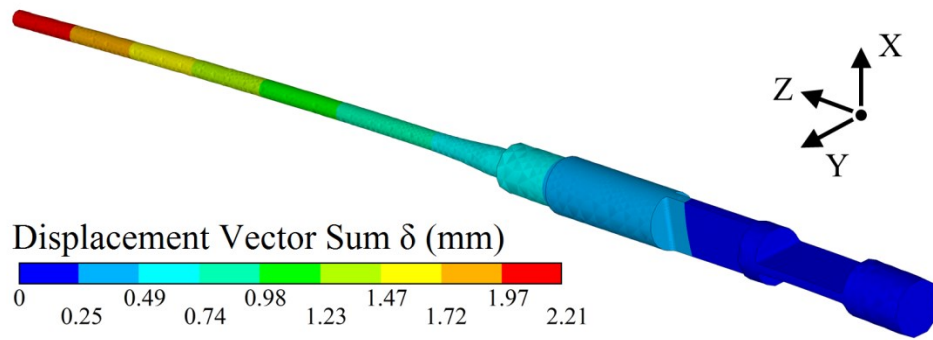


Figure 5.5: FE model result denoting displacement vector sum of the haptic dissector to a -1N lateral load in the X-direction.

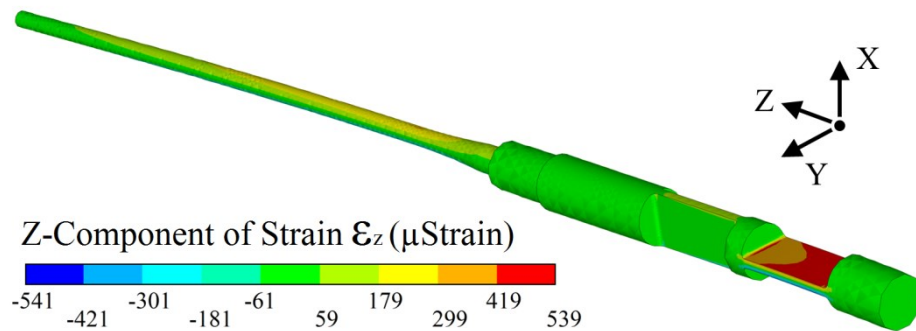


Figure 5.6: FE model result denoting the strain distribution  $\epsilon_z$  in the haptic dissector as a result of a -1N lateral tip load in the X-direction.

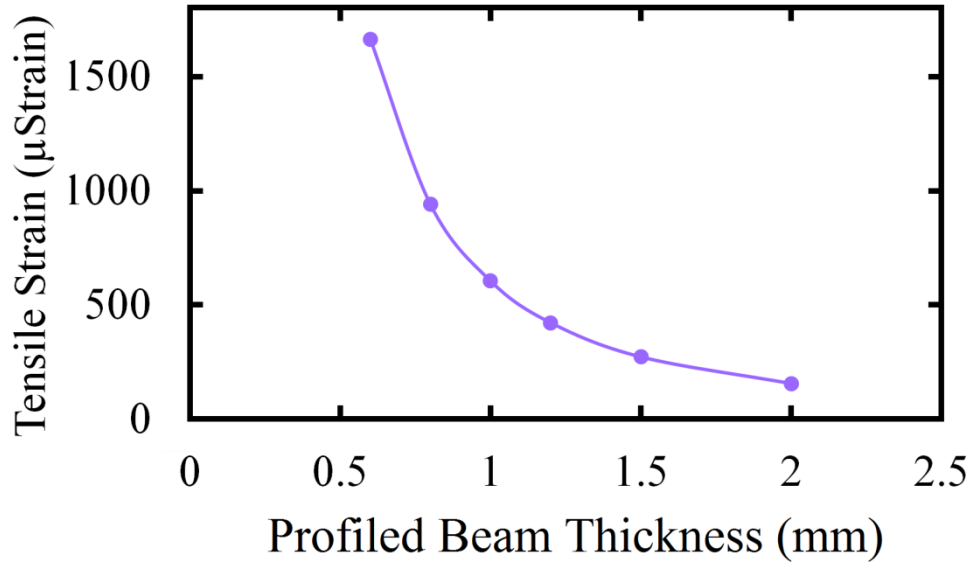


Figure 5.7: Plot showing the tensile strain  $\varepsilon_z$  at the gauge location point as a function of varying beam profile thickness.

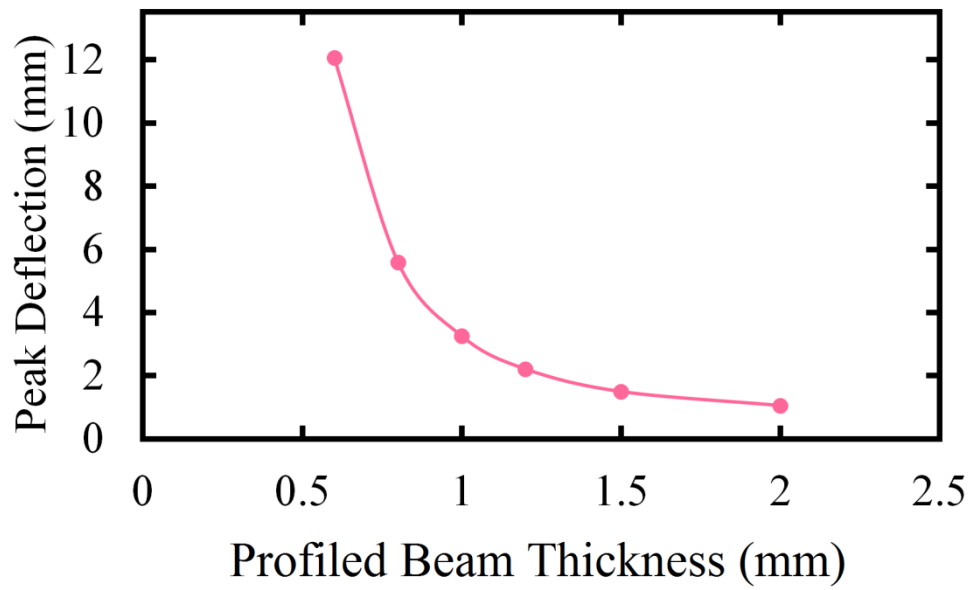


Figure 5.8: Plot showing the peak deflection (displacement vector sum magnitude) of the haptic dissector sensing assembly as a function of varying beam profile thickness.

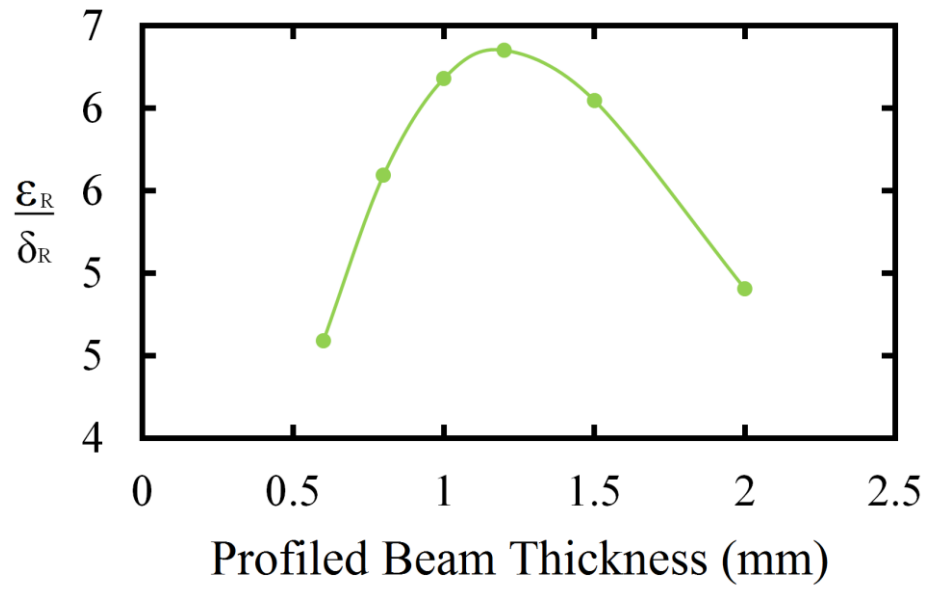


**Figure 5.9: Plot showing the ratio  $\epsilon_R/\delta_R$  with respect to profile beam thickness.**

Figure 5.7 and Figure 5.8 show the non-linear relationships of both  $Ez$  and  $\delta$  with respect to beam profile thickness. Whilst beam strain and thus sensor sensitivity is greatly increased with lowering beam profile thickness, the instrument deflection also becomes excessively high: for example, a 0.6mm thickness beam yields a 12mm instrument deflection. In considering the relative increase in sensor sensitivity against the relative decrease in rigidity, the optimum point was found to be a beam profile thickness of 1.2mm. This represents a twenty-fold increase in strain/sensor sensitivity and a three-fold reduction in instrument rigidity with respect to an un-profiled dissector shaft.

**Table 5.1: The finite element model simulation results that were used to optimise the smart dissecting device. \*This value is  $\delta_I$ . \*\*This value is  $\varepsilon_{zI}$ .**

| Beam Thickness<br>(mm) | Number of<br>Elements in<br>Model | Strain $\varepsilon_z$<br>( $\mu$ Strain) | Peak Deflection $\delta$<br>(mm) | $\frac{\varepsilon_z}{\varepsilon_{z1}}$ | $\frac{\delta}{\delta_1}$ | $\frac{\varepsilon_R}{\delta_R}$ |
|------------------------|-----------------------------------|-------------------------------------------|----------------------------------|------------------------------------------|---------------------------|----------------------------------|
| Un-Profiled            | 18496                             | 21**                                      | 0.71*                            | 1.00                                     | 1.00                      | 1.00                             |
| 2                      | 48315                             | 156                                       | 1.06                             | 7.25                                     | 1.48                      | 4.91                             |
| 1.5                    | 46597                             | 272                                       | 1.50                             | 12.68                                    | 2.10                      | 6.05                             |
| 1.2                    | 44376                             | 422                                       | 2.21                             | 19.65                                    | 3.09                      | 6.35                             |
| 1                      | 43890                             | 605                                       | 3.26                             | 28.16                                    | 4.56                      | 6.18                             |
| 0.8                    | 41675                             | 940                                       | 5.59                             | 43.79                                    | 7.83                      | 5.59                             |
| 0.6                    | 40319                             | 1664                                      | 12.07                            | 77.50                                    | 16.88                     | 4.59                             |

### **5.1.2 Mechatronics Implementation**

The strain gauges were configured in a half-bridge arrangement, ensuring good sensitivity and temperature compensation. The strain measurements signals are amplified through strain gauge amplifiers (VS412 Strain Gauge Amplifier, RS Components) before being acquired by a real-time controller (CompactRIO, cRIO-9022, National Instruments Corp) through a FPGA interface running at 30kHz. The controller converts the digitised sensor voltages in to Newtons. The total lateral load exerted by the haptic dissector is computed by taking the modulus of the two orthogonal lateral loads. The total lateral force is compared to a pre-set force threshold. If the total lateral force exerted by the dissector exceeded the pre-set limit, the controller switches on the on-board vibrotactile feedback. This vibrotactile feedback is generated through a voice coil actuator (VCA) (LVCM-010-013-1, MotiCont) and a spring-damper assembly which is integrated within the hand-held haptic dissector. A 250Hz sinusoidal signal is inputted to a linear amplifier (LSC 30/2 linear 4Q Servoamplifier, Maxon) which supplied power to the VCA. A vibration frequency of 250Hz was chosen as it has previously been identified as an excitation frequency that provides maximum stimulation [112].

## **5.2 Bench Tests**

### **5.2.1 Sensor Calibration**

Calibration of the force sensing system was performed by loading the haptic dissector shaft against a calibrated force-torque sensor (F/T) sensor (ATI Nano17, ATI Industrial Automation, Inc.) and correlating the amplified sensor output voltage with the measured lateral force. The F/T sensor was mounted on to a translatable stage and the haptic dissector was affixed to an optical table. The F/T sensor was brought in to contact with the haptic dissector shaft tip and was transversely loaded at a constant velocity up to a peak load of 1.0N. The haptic dissector shaft was then de-loaded back to 0.0N. This was performed in four 90° increments to allow bi-directional force calibration in the plane of the haptic dissector's curved tip (longitudinal X-axis) and in the plane perpendicular to the dissector's curved tip (lateral Y-axis). These coordinate definitions are consistent with those adopted in the FE simulations which were defined in Figure 5.3.

The sensor output voltages for loading in the longitudinal X-axis are presented in Figure 5.10 and the sensor output voltages for loading in the lateral Y-axis are presented in Figure 5.11. Linear regression models using the least squares method were calculated for the X and Y axis sensing assemblies for loading in the X and Y directions respectively. The force-voltage relationships were found to have regression ratios of 2.78V/N ( $R^2=1.00$ ) for the X-axis and 2.58V/N ( $R^2=1.00$ ) for the Y-axis.

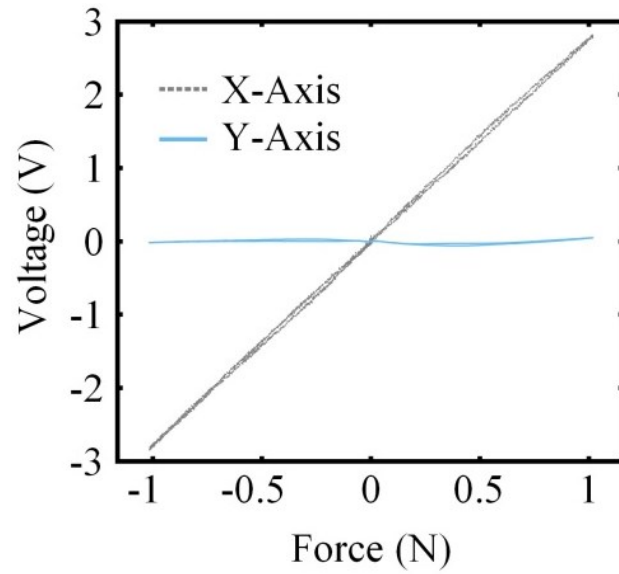


**Figure 5.10: Strain gauge calibration for loading about the X-axis.**

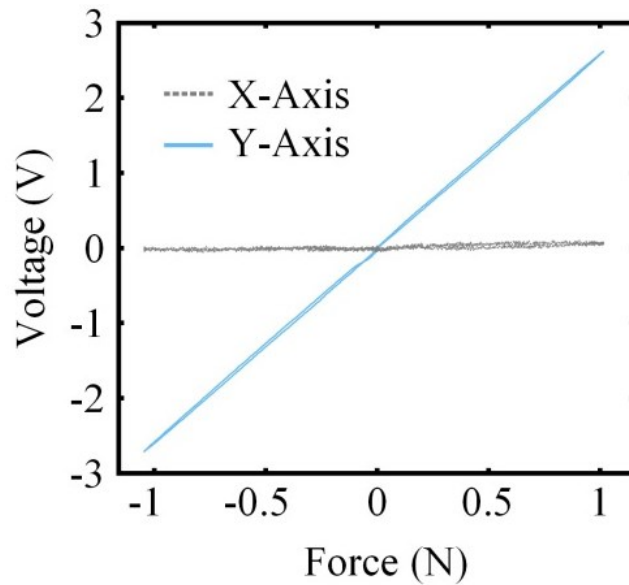


**Figure 5.11: Strain gauge calibration for loading about the Y-axis.**

### 5.2.2 Vibration Transmission Testing

An experiment was conducted in order to confirm that the vibrotactile feedback does not generate significant forces that can be transmitted from the tool tip to the surrounding tissue. The haptic dissector was lightly clamped to an optical table with a rubber mount so as to allow the haptic dissector to vibrate freely. A calibrated F/T sensor (ATI Nano17, ATI Industrial Automation, Inc.) was orientated to be co-axial with the haptic dissector shaft and translated to a static force of 0.3N. This orientation was chosen to capture the peak vibrations generated by the VCA and spring-damper assembly which generate vibrations that are axial to the haptic dissector shaft. The VCA was excited with a sinusoidal input voltage of amplitude 0-6V and at a frequency of 250Hz. The axial force signal from the F/T sensor was then recorded at 40kHz. The peak-to-peak axial force measured in the vibration bench test was measured to be 0.03N; this is shown in Figure 5.12

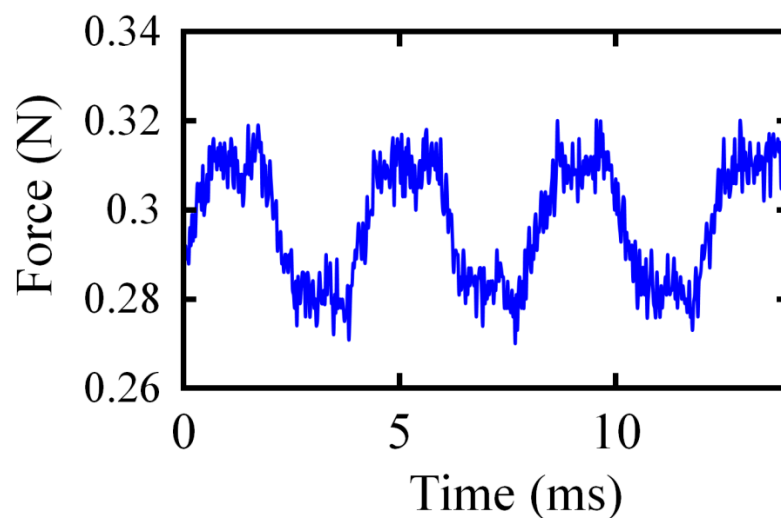


Figure 5.12: The force profile obtained from the vibration transmission study.

### 5.3 Retraction Force Study

Neurosurgical procedures commonly involve bimanual tasks in which the surgeon must simultaneously retract tissue and manipulate the exposed tissue effectively and delicately. Ideally, the surgeon will perform effective tissue retraction whilst exerting the minimum amount of force. Effective, consistent tissue retraction is challenging because of erroneous physiological hand motion, and in particular the drift component [25]. It was hypothesised

that human subjects would exert lower retraction forces with the assistance of vibrotactile force-threshold feedback. To test this hypothesis, a bimanual task was devised to represent tissue retraction and manipulation. All users recruited for this study gave their informed consent prior to the study commencing. Users were required to exert a minimum, consistent force with their left (non-dominant) hand using the haptic dissector to represent tissue retraction whilst performing a peg transfer task with their right hand to represent a surgical task. The study measures the forces exerted by the haptic dissector for the cases with and without force-threshold vibrotactile feedback. The experimental setup is shown in Figure 5.13.

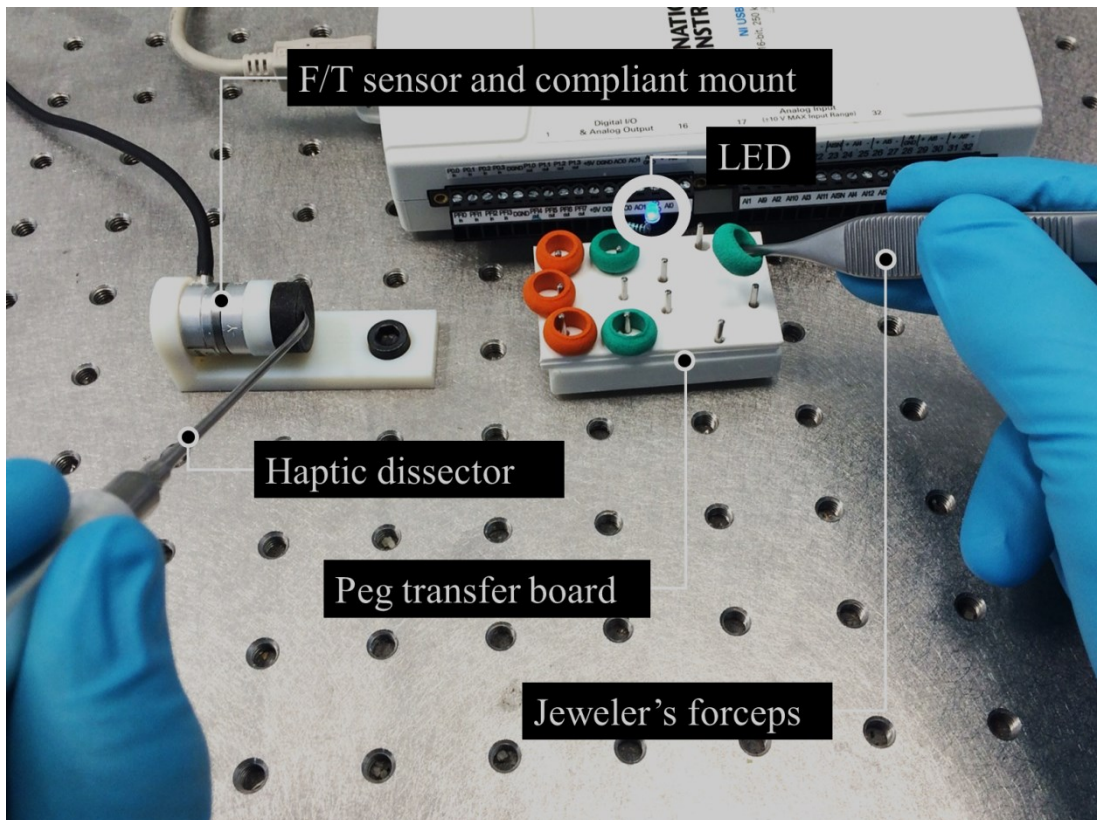


**Figure 5.13: Experimental setup for the retraction force user study.**

Users were required to rest their forearms whilst performing the bimanual task which is representative of the surgical scenario and furthermore, studies have shown that tremor is exacerbated when an armrest is not used in microsurgery [163]. A F/T sensor (ATI Nano17, ATI Industrial Automation, Inc.) mounted to an optical table was used to measure the forces

exerted in each study. Only the forces in the retraction direction were measured in this study. The force data was sampled at 40Hz. A compliant, rubber-like material was mounted on the force sensor to represent tissue and prevent slippage of the instrument tip. In order to prevent users applying too little force, they were requested to exert a minimum force which was indicated by a light emitting diode (LED). A minimum force threshold of 0.2N was used to represent an upper level of tissue retraction in neurosurgery [159]. The peg-transfer task was performed simultaneously; the task aims to prevent the users from concentrating solely on maintaining a constant retraction force which is not realistic of the surgical scenario and would bias the results. The peg-transfer task required users to sequentially transfer six rings using surgical forceps. Prior to commencing the peg-transfer task, users were asked to gradually apply force so as to prevent overshooting the minimum force threshold. Once the minimum threshold was reached, the users were requested to perform the peg transfer task repeatedly for 60 seconds whilst attempting to keep the LED lit by applying over 0.2N of force. Each user performed this procedure a total of six times, with and without force-threshold feedback (3+3) in a randomised order to prevent biasing of any potential learning effects. The F/T sensor and haptic dissector sensors were all zeroed prior to each trial. For the cases in which force-threshold feedback was provided, a force limit of 0.3N was implemented, based on the data presented by Marcus *et al* [159]. Vibrotactile feedback was provided in the form of a 250Hz vibration (0-6V amplitude). Users wore latex surgical gloves which invariably attenuate haptic user perception but is representative of the real world surgical scenario. The Mann-Whitney U test was performed to compare force and time metrics for the cases with and without force-threshold feedback, with a value of  $p < 0.05$  considered statistically significant.

Ten users were recruited for this study (4 female, 6 male) with no reported health conditions. They were aged between 21 and 33, with a median age of 26. The users all described themselves as being right-handed. Box plots of the retraction force user study are presented in Figure 5.14 shows the total distribution of forces across the entire study with and without force-threshold vibrotactile feedback cases (10 users  $\times$  3 runs  $\times$  60 seconds  $\times$  40Hz sampling frequency). A second box plot presents the distribution of maximum forces reported for each individual trial across the entire study (max values recorded for 10 users  $\times$  3 runs). The cumulative time period for when users exceeded the 0.3N threshold and when they exerted less than 0.2N were recorded for each run. Two box plots in Figure 5.15 indicate these

cumulative time period distributions across the entire study (cumulative time period for 10 users  $\times$  3 runs). Table 5.2 summarises the key findings of the retraction force user study.

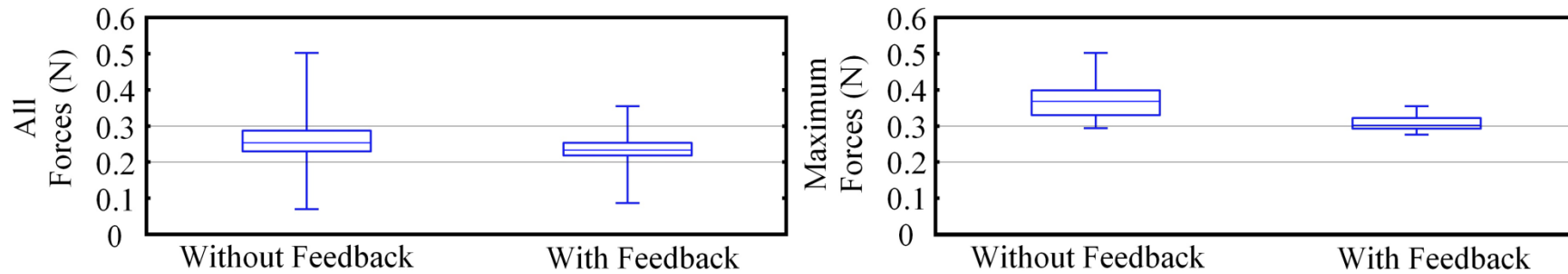


Figure 5.14: Box plots for the retraction force study comparing with and without force-threshold feedback cases, showing the total distribution of forces (left), the distribution of maximum reported forces (right). Boxes indicate the lower quartile, upper quartile and median, whiskers represent the minimum and maximum values.

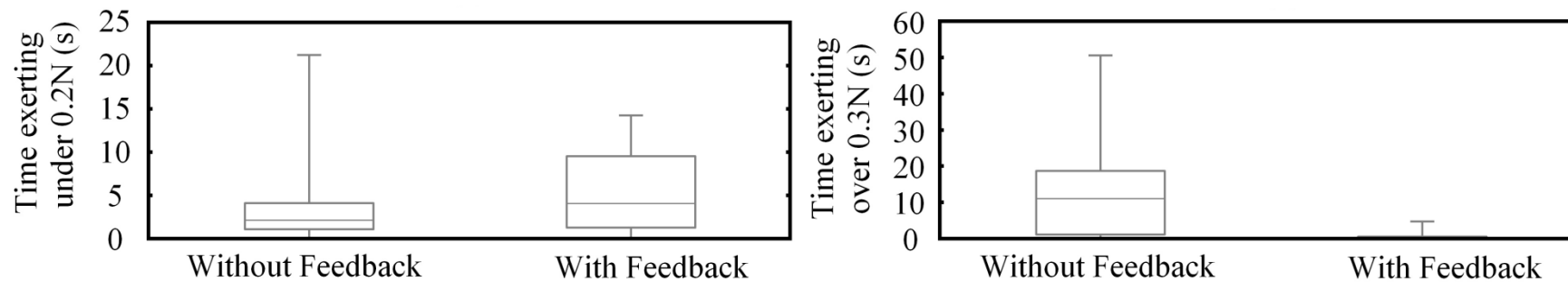


Figure 5.15: Box plots for the retraction force study comparing with and without force-threshold feedback cases, showing the distribution of time users spent exerting less than 0.2N (left) and the distribution of time users spent exerting greater than 0.3N (right). Boxes indicate the lower quartile, upper quartile and median, whiskers represent the minimum and maximum values.

**Table 5.2: Key statistics for the force retraction user study.**

|                                                            | Without Force-Threshold<br>Feedback |        | With Force-Threshold<br>Feedback |
|------------------------------------------------------------|-------------------------------------|--------|----------------------------------|
| Max force across entire study (N)                          | 0.505                               |        | 0.357                            |
| Median force across entire study (N)                       | 0.254                               |        | 0.235                            |
| Interquartile force range across entire study (N)          | 0.057                               |        | 0.035                            |
| p-value comparing total force distribution                 |                                     | <0.001 |                                  |
| Median of maximum forces per run (N)                       | 0.370                               |        | 0.303                            |
| p-value comparing maximum forces per run                   |                                     | <0.001 |                                  |
| Median time users exerted greater than 0.3N per run (secs) | 4.69                                |        | 0.05                             |
| p-value comparing time users exerted greater than 0.3N     |                                     | <0.001 |                                  |
| Median time users exerted less than 0.2N per run (secs)    | 2.18                                |        | 4.18                             |
| p-value comparing time users exerted less than 0.2N        |                                     | 0.085  |                                  |

## 5.4 Intervening Force User Study

Psychophysical studies have demonstrated that humans have a good ability to replicate force application levels with respect to a reference force after a short time delay in force-matching tasks [164]. Park *et al* showed that if users are required to generate an intervening force during a force-matching task, they would tend to overestimate or underestimate the reference force if the magnitude of the intervening force was doubled or halved, respectively [165]. It was hypothesised that vibrotactile feedback would prevent users overshooting the reference force in a force-matching task with an intervening force of double the reference force.

The experimental protocol is derived from Park *et al* [165]. All users recruited for this study gave their informed consent prior to the study commencing. Users were requested to exert a lateral 0.3N force on to a F/T sensor (ATI Nano17, ATI Industrial Automation, Inc.) for a period of six seconds in order to allow sufficient time to react and reach the desired force level. Visual feedback was provided in the form of an indicator bar which indicated the force level, allowing user to match the force they exerted to the reference force. Users were then required to exert 0.6N for a six second period following a three second interval; the same visual feedback indicator bar was provided during the second force-matching phase. In the final phase of the experiment, users were requested to exert the original 0.3N reference force for a six second period following a three second interval, without any visual feedback. Each user performed this procedure a total of six times, with and without force-threshold feedback (3+3) in a randomised order. Data was recorded throughout, but only the final three seconds of the final force matching phase was considered in the analysis to eliminate the transient, user reaction period of the force matching task. The F/T sensor and haptic dissector sensors were all zeroed prior to each trial. The force-feedback threshold was set to 0.3N and users were informed that they should reduce the amount of force exerted when they experienced vibrotactile feedback. The F/T data was sampled at 40Hz. A compliant, rubber-like material was mounted on the force sensor and users wore latex surgical gloves. A 250Hz vibration (0-6V amplitude) was used for the trials with vibrotactile feedback. The Mann-Whitney U test was performed to compare the forces exerted for the cases with and without vibrotactile feedback, with a value of  $p < 0.05$  considered statistically significant.

Ten users were recruited for this study (4 female, 6 male) with no reported health conditions. They were aged between 24 and 33, with a median age of 27.5. The users all described themselves as being right-handed. The results of the intervening force study are presented in Figure 5.16 as a box plot which indicates the total distribution of forces across the entire study (10 users  $\times$  3 runs  $\times$  3 seconds  $\times$  40Hz sampling frequency). Of note, the interquartile range with force threshold-feedback was considerably less than without force-threshold feedback (0.031N versus 0.150N respectively). Figure 5.17 an example force matching procedure for trials with and without force-threshold feedback. The key findings of the study are summarised in Table 5.3.

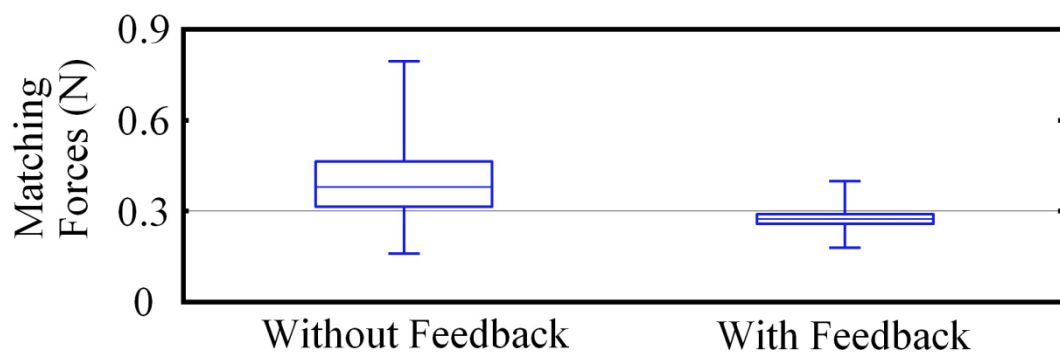


**Figure 5.16:** Box plots for the intervening force experiment showing the distribution of forces applied in the final three seconds of the final force matching phase, with and without force-threshold feedback. Boxes indicate the lower quartile, upper quartile and median, whiskers represent the minimum and maximum values.

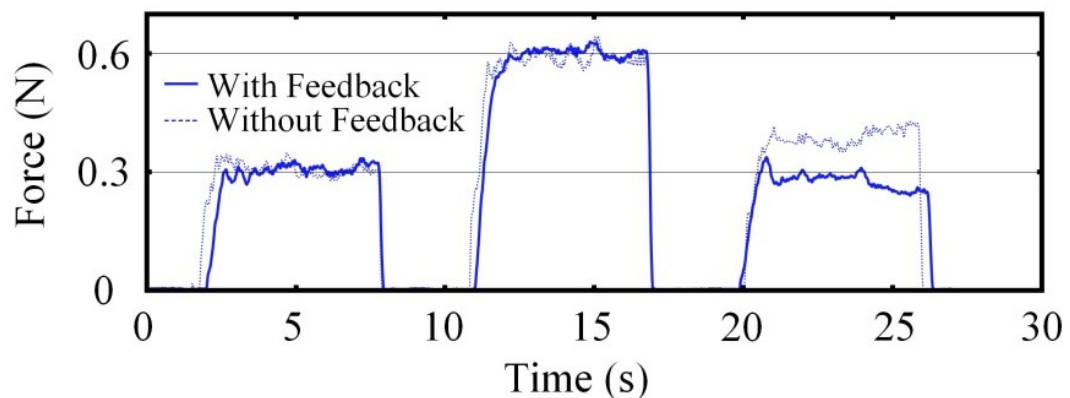


**Figure 5.17:** An example plot showing the forces exerted in a typical force matching procedure, with and without force-threshold feedback.

**Table 5.3: Key statistics for the intervening force user study.**

|                                                   | Without Force-<br>Threshold<br>Feedback | With Force-<br>Threshold<br>Feedback |
|---------------------------------------------------|-----------------------------------------|--------------------------------------|
| Max force across entire study (N)                 | 0.799                                   | 0.403                                |
| Median force across entire study (N)              | 0.380                                   | 0.274                                |
| Interquartile force range across entire study (N) | 0.150                                   | 0.031                                |
| p-value comparing total force distribution        | <0.001                                  |                                      |

## 5.5 Cadaveric User Study

In order to better evaluate the effectiveness and safety of the haptic dissector during the complex manoeuvres that characterise microneurosurgery, a cadaver study was performed. Ethical approval was obtained from the Proportionate Review Sub-committee of the National Research Ethics Service (NRES) Committee East Midlands. A formalin fixed cadaver head was procured from the Department of Anatomy, Imperial College London. A surgical resident performed the procedure. The surgical resident was gowned and gloved (Ansell Healthcare EMEA, Brussels, Belgium), and the head draped, in the usual manner. The head was secured in a clamp. A curvilinear incision was made from the root of the zygoma to the anterior midline, and a single myocutaneous flap turned. A high-speed drill (B. Braun. Melsungen, Germany) was used to fashion a pterional craniotomy, approximately 70mm in diameter. A durotomy was then performed and the dural flap retracted basally. An Albert Wetzlar operating microscope (Albert Wetzlar GmbH, Wetzlar, Germany) and a rigid endoscope (VSIII Endoscope, VisionSense, Petach Tikva, Israel) were used for visualisation. Microsurgical dissection of the Sylvian fissure was performed using straight microscissors during the initial sharp arachnoid dissection. Jeweler's forceps were then used with the smart haptic dissector to perform blunt arachnoid dissection, with the superficial Sylvian veins retracted to the temporal lobe side, and bridging veins preserved where possible. Dissection proceeded in a stepwise manner, alternating at two minute intervals between the use of the haptic dissector with vibrotactile feedback and without vibrotactile feedback. A 0.3N force threshold was implemented when vibrotactile feedback was activated. The force sensing data was sampled at 100Hz for all cases. A video of the microsurgical dissection was also

recorded. A total of twelve minutes of dissection was performed, comprising six minutes with vibrotactile feedback, and six minutes without force-threshold feedback. The Mann-Whitney U test was performed to compare the forces exerted for the cases with and without vibrotactile feedback, with a value of  $p < 0.05$  considered statistically significant.

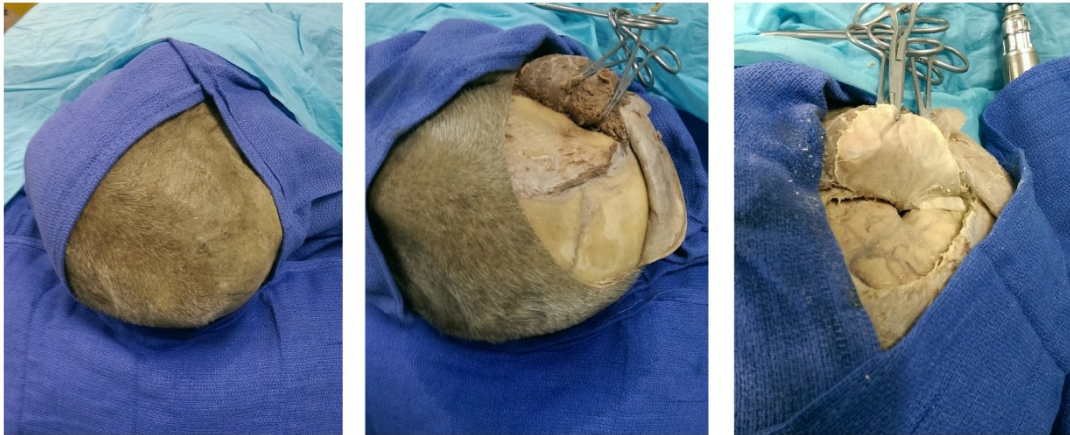


**Figure 5.18: Preparation of the cadaver head prior to the pre-clinical study: (Left) Cadaver head draped (Centre) Standard left curvilinear incision and myocutaneous flap lifted (Right) Pterional craniotomy and durotomy exposing the left Sylvian fissure.**

Example plots of dissection trials with and without force-threshold feedback are presented in Figure 5.19. A still image from the video feed shows the haptic dissector being used on the cadaver in Figure 5.20, and a box plot also indicates the total distribution of forces exerted across the entire study. Table 5.4 also summarizes the key findings of the cadaveric study.

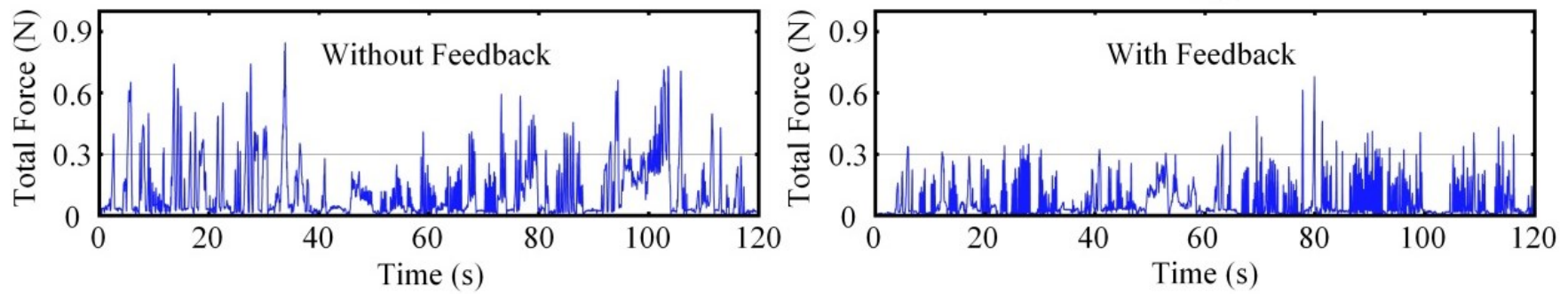


Figure 5.19: Example runs of the cadaveric microdissection without force-threshold feedback (left) with force-threshold feedback (right).

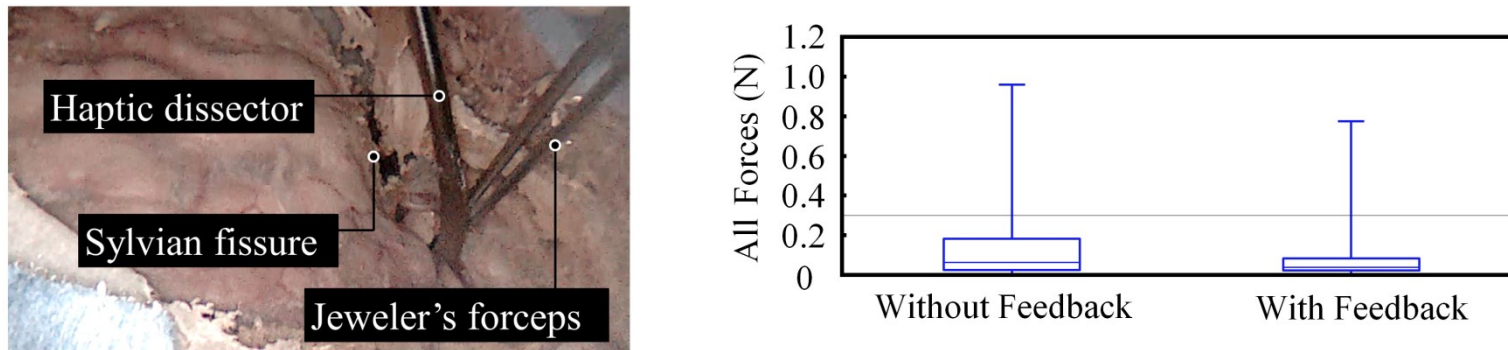


Figure 5.20: Video capture of cadaveric study (left) and box plot showing the total distribution of force exerted over the entire cadaveric study (right). Boxes indicate the lower quartile, upper quartile and median, whiskers represent the minimum and maximum values.

**Table 5.4: Key statistics for the cadaveric study.**

|                                                        | Without<br>Force-<br>Threshold<br>Feedback | With<br>Force-<br>Threshold<br>Feedback |
|--------------------------------------------------------|--------------------------------------------|-----------------------------------------|
| Maximum force across entire study (N)                  | 0.963                                      | 0.779                                   |
| Median force across entire study (N)                   | 0.063                                      | 0.038                                   |
| Interquartile force range across entire study (N)      | 0.159                                      | 0.062                                   |
| p-value comparing total force distribution             | <0.001                                     |                                         |
| Time user exerted greater than 0.3N (secs)             | 41.49                                      | 8.31                                    |
| Proportion of study user exerted greater than 0.3N (%) | 11.53                                      | 2.31                                    |

## 5.6 Discussion and Conclusions

The calibration tests confirmed that the device could effectively measure force in the range required for microneurosurgery. Furthermore, the peak vibration generated by the vibrotactile feedback system (0.03N) was comparatively low relative to the forces exerted during microneurosurgery. The psychophysical user studies demonstrate the ability of the haptic dissector to significantly improve perception of a pre-determined force limit. The retraction force study illustrated that users drift away from applying an optimum magnitude of force when distracted but that force threshold feedback improves this. For the runs when force-threshold feedback was activated, there was a 29% reduction in the maximum reported force. Furthermore, there was an 18% reduction in the median of maximum forces applied for each run. Users also spent significantly more time exerting greater than the 0.3N threshold without feedback ( $p < 0.001$ ), which would also increase the risk of causing iatrogenic injury in a surgical scenario. A 38.6% reduction of the interquartile range also demonstrates that the addition of vibrotactile feedback improves the consistency with which users performed the simulated retraction. Users did spend a greater period of the study applying less than 0.2N of force when vibrotactile feedback was present. This was not shown to be significant ( $p = 0.085$ ), although it is unsurprising that users behaved more conservatively with force-threshold vibrotactile feedback. Erroneous physiological motion will have also played a role in increasing the time users spent exerting less than 0.2N since

the vibrotactile feedback force limit constrains the user to exert the retraction force within a 0.1N band.

The results of the intervening force study are also in accordance with the findings that were reported by Park *et al* [165]. Users overestimated the 0.3N reference force by a median value of 21% after the intervening force of 0.6N without vibrotactile feedback. When force-threshold feedback was present, users exerted a median force of 0.27N when attempting to match the reference force. This underestimate is a result of users applying less force once they had felt the vibrotactile feedback at the 0.3N limit. The distribution of matching forces was significantly different between the cases with and without force threshold feedback ( $<0.001$ ). Users also exerted force with considerably greater consistency; the interquartile range was reduced by 79% to 0.031N when force-threshold feedback was activated. This reduction in force variability supports the idea that vibrotactile feedback does not itself significantly interfere with tool-tissue interactions.

The example plot in Figure 5.17 shows a user overshooting the 0.3N limit before reducing their force level to below the 0.3N limit upon detecting the force-threshold feedback which was typical of user behaviour. The study highlights that force perception varies according to prior force application which is of interest to the microsurgical community. In microneurosurgery, there are many procedures in which the operating surgeon must exert a multitude of force magnitudes. For example, lumbar microdissection requires high force application during retraction of the paraspinal muscles during the early stages of the operation, followed by delicate microsurgical manipulation during the later stages. The implementation of force thresholds could therefore benefit this type of procedure by allowing the operating surgeon to recalibrate how much force they exert.

The cadaveric study demonstrates that the haptic dissector operates successfully in a clinical setting. The haptic dissector demonstrated a significant reduction in force magnitude and time users exerted greater than the 0.3N limit during the microdissection procedure. There are instances when the resident exerted forces greater than 0.3N, the peak force exerted with force-threshold feedback was 0.779N compared to 0.963N without. This demonstrates that in

some instances, the resident has chosen to ignore the vibrotactile feedback and relied on their own judgement as to how much force should be exerted. It is important to consider that higher force does not always result in tissue injury. The nature of the tissue being manipulated and the instrument that is used are key factors in the surgeon's decision making process. Ultimately, the device provides a feedback system and the choice has to how much force exerted still remains with the surgeon.

It should be acknowledged that these studies had several weaknesses. The smart haptic dissector only has force sensing in 2 DoF and so there is no data for the forces exerted in the axial direction, although the psychophysical studies and blunt dissecting procedures were predominantly comprised of lateral force application. Clinically-inexperienced users were recruited for the psychophysical experiments and only a single resident performed the procedure on a single cadaver head. It is expected that the device would be of less use to experts who would better exploit visual cues, such as tissue deformation in order to gauge how much force should be exerted. This study can be significantly improved through the addition of intermediate and expert users in order to provide evidence of construct validity in the tasks performed. In addition, force was the only metric used to quantify surgical performance in the cadaveric study. Further metrics, such as tool motion data, and Objective Structured Assessment of Technical Skill (OSATS) scores [166], would also benefit such a study.

This chapter has presented a hand-held microneurosurgical instrument that uses vibrotactile-based feedback to indicate when a force threshold has been exceeded. The results from this chapter demonstrate that force-threshold haptic feedback could improve the safety of microneurosurgical interventions. Another finding from this chapter is that force-threshold based haptic feedback is also useful during interventions that require manipulation in the supra-threshold haptic range. Psychophysical user experiments demonstrate that this approach can limit the amount of force that is applied when users are distracted or when their force perception has been deliberately altered. The psychophysical user experiment methodologies themselves are also of potential use for assessing the efficacy of force-threshold based haptic feedback. The developed device is potentially of benefit to neurosurgical trainees who at present may only recognise force thresholds are exceeded once

tissue injury has already occurred. In practice, force levels might be exceeded under the correct conditions according to the judgement of the operating surgeon. The haptic feedback strategy presented in this chapter has shown to be effective in minimising the amount of force applied during surgery but the proposed device does not have the capability to avoid critical anatomical regions altogether. The next chapter will thus examine the case in which haptic feedback of a predefined “active constraint” to help a surgeon guide their interventional tools during delicate surgery.

# Chapter 6: Ungrounded Haptic Feedback Incorporating Active Constraints<sup>†</sup>

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<sup>†</sup> Content from this chapter was published as:

**An Ungrounded Hand-Held Surgical Device Incorporating Active Constraints with Force-Feedback.**  
Payne C.J., Kwok K. and Yang G-Z. IEEE Intelligent Robots and Systems, Tokyo, Japan, 2013. pp2559-2565.  
*ICROS Best Application Award Finalist.*

This thesis has thus far examined haptic feedback methodologies for augmenting tool-tissue interactions and for informing the operating surgeon if they begin to exert excessive force. This chapter examines the case in which tool-tissue contact is to be avoided altogether. “Active constraints” [6] also known as “virtual fixtures” [106] were introduced in Chapter 2. Active constraints exploit synergistic human-robot control to guide a surgical tool to or away from pre-defined anatomical regions [167][108]. This principle has been adopted extensively in orthopaedic applications for the precise milling of bone for implant preparation [105]. Active constraints have mainly been applied to grounded, cooperatively-controlled robotic systems, which permit haptic feedback to the surgeon. Examples include the original Acrobot™ (Active Constraint Robot, see Figure 6.1), the subsequent Sculptor RGA (Stanmore Implants, Elstree, UK.), RIO® MAKOpasty® (Stryker Corp., MI, USA) and the Steady-Hand robot for Otorhinolaryngology applications [108].



**Figure 6.1:** The Acrobot™ surgical system developed at Imperial College (©2002 reproduced from Jakopc *et al* with permission from Springer) [105].

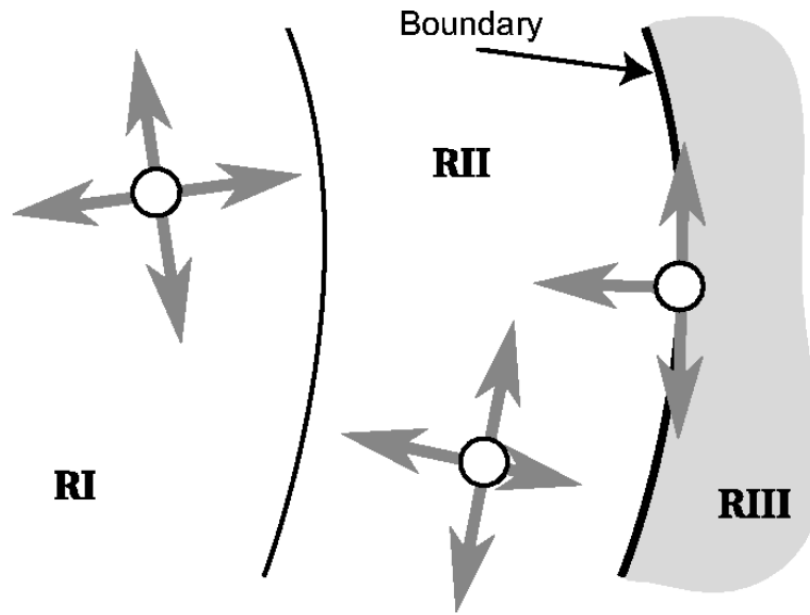


**Figure 6.2: The active constraint principle for orthopaedic surgery from Jakopcic *et al* ©2003 IEEE [107].**

These grounded robotic platforms use force control to guide the operator along or away from the active constraint. This principle, developed by Davies *et al* [107] is highlighted in Figure 6.2. RI, RII and RIII refer to three separate regions. RI is a safe region in which the robot does not impede the motion of the surgical tool. RII is a region of impedance; the robot actively resists motion in the direction normal to the pre-defined boundary. The final region, RIII, is the “no-go” boundary at which point the robot becomes very stiff to prevent penetration of the surgical instrument tip in to an unsafe region of anatomy. The application of active constraints to ungrounded, hand-held instruments requires an alternative approach.

Hand-held surgical devices with active constraints must implement position-based control on the tool tip rather than force-based constraints because of the absence of a grounding frame. The Navio PFS (BlueBelt Technologies, Plymouth, MN, USA) and the device developed by Brisson *et al* [51] are examples of hand-held tools that have been devised for precise milling of pre-defined 3D geometries in knee replacement surgery. These devices are optically-tracked and their rotating burrs are retracted when the constraint is encountered to allow intelligent sculpting of a bone surface. Active constraints have also been implemented on

Micron [48][49]. In this work, the instrument tip's position is constrained so as to not deviate from a pre-programmed trajectory. This research also features 'soft' positional constraints that effectively scale the tool tip position relative to the hand motion of the surgeon. The application of active constraints significantly improves tool tip accuracy in micromanipulation tasks; however there is no force-feedback to the operator that indicates tool interaction with the active constraint.

The novel contribution of this chapter is the fusion of position-based active constraints and force-feedback in to an ungrounded, hand-held device. The motion compensation scheme provides a position constraint on the surgical tool whilst the force-feedback allows for tactile interaction with the active constraint. This technique could allow for surgical tools to be guided through safe channels with less cognitive burdening of the operator. It is envisaged that such a device has potential applications in neurosurgery, otorhinolaryngology and paediatric surgery, in which surgeons must operate around delicate anatomical areas within tight spatial constraints and with restricted vision. This chapter includes bench tests that verify the motion compensation and force-feedback aspects of the proposed methodology. User studies also demonstrate the combined impact of motion compensation and force-feedback in allowing human perception of an active constraint with an ungrounded device.

## **6.1 Motion Compensation Device with Force-Feedback**

The device is designed to be held as a stylus by the operator with a surgical tool protruding from the tip as shown in Figure 6.3. The surgical tool can be actuated along its axis so as to retract into the stylus as it encounters an active constraint. A force-feedback display exerts forces on to the operator's fingertip when the active constraint is penetrated. An Optotrak Certus® (Northern Digital Inc, Ontario, Canada) optical tracking system is used in order to locate the device's position and orientation with respect to the active constraint. The optical tracking system is capable of achieving sub-millimetre resolution and operates at acquisition speeds in excess of 1kHz.

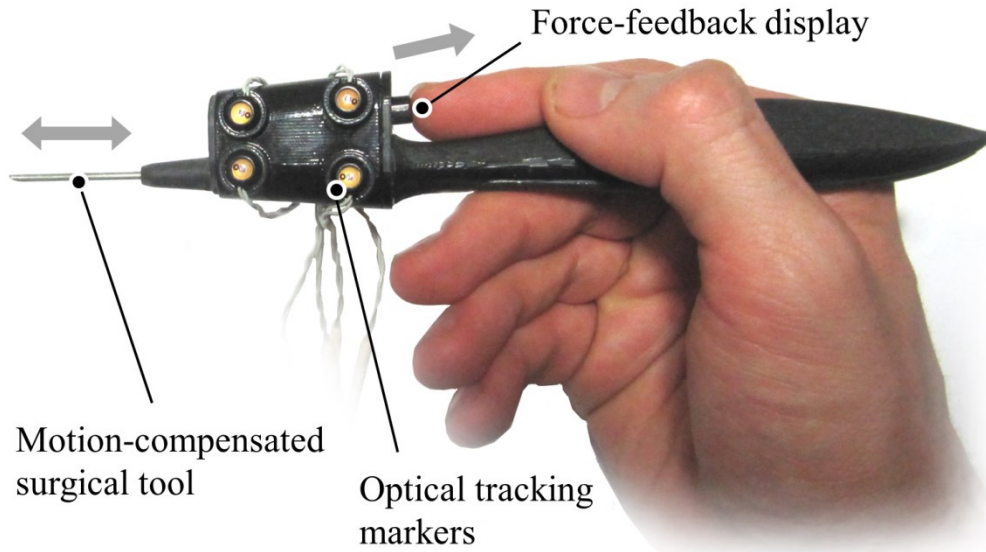


**Figure 6.3** A photograph of the hand-held device illustrating its key features of motion compensation and force-feedback.

### 6.1.1 Instrument Tracking and Active Constraints

Four optical markers are attached to the casing of the device and are tracked by the optical tracking system. A rigid body coordinate system is defined at the origin of the optical markers ( $F_{markers}$ ) which is known with respect to a world coordinate system ( $F_{world}$ ) defined by the Optotrak Certus<sup>®</sup>. Rigid transformations are computed in order to obtain coordinate systems at the base of the surgical tool ( $F_{base}$ ) and at the uncompensated tool tip ( $F_{tip}$ ). These transformation computations are as follows:

- ${}^W_M T$  is the transformation from the  $F_{world}$  to  $F_{markers}$  which is provided by the optical tracking system as the tool is manipulated by the user.
- ${}^M_B T$  is the transformation from the  $F_{markers}$  to  $F_{base}$  which is fixed by the design of the device.
- ${}^B_T T$  is the transformation from the  $F_{base}$  to  $F_{tip}$  which is fixed by the design of the device at the state of the tool position.

- ${}^W_T T$  is the transformation from the  $F_{world}$  to  $F_{tip}$  which can be obtained by multiplying the above transformations together:

$${}^W_T T = {}^W_M T {}^M_B T {}^B_T T \quad 6.1$$

The tracking coordinate systems are illustrated in Figure 6.4. The origin of  $F_{tip}$  is the uncompensated surgical tool position  $P_u$ . This is the tool tip position when the tool is fully advanced from the stylus and is always fixed relative to  $F_{base}$ . The vector between coordinate systems  $F_{base}$  and  $F_{tip}$  represents the axis of the surgical tool. An active constraint surface is also defined in the world coordinate system so that when the uncompensated tool tip  $P_u$  moves beyond the active constraint surface, a theoretical penetration distance is computed and defined as  $\chi_u$ . This is the distance between  $P_u$  and the point of intersection between the active constraint surface and the axis of the tool, defined as  $P_i$ . The actuator then retracts the tool in proportion to the penetration distance  $\chi_u$  by a factor of  $K_s$ . The tool can be retracted by the same distance as  $\chi_u$  so that there is no penetration of the active constraint, thus providing an infinitely rigid constraint ( $K_s = 1$ ). Alternatively, the tool can be retracted by a fraction of  $\chi_u$  so that the tip motion is scaled, forming a ‘soft constraint’ (where  $0 < K_s < 1$ ). For the proposed device, the operator’s index finger is placed onto a translatable slider that exerts a force of magnitude  $F_f$  that is directly proportional to  $\chi_u$  by a factor  $k$  so as to provide a spring-like force-feedback response. This translatable slider-based configuration proposed is based on the design presented in Chapter 3. The stiffness parameter  $k$  can be varied so that the operator can feel the active constraint to have different stiffness properties.

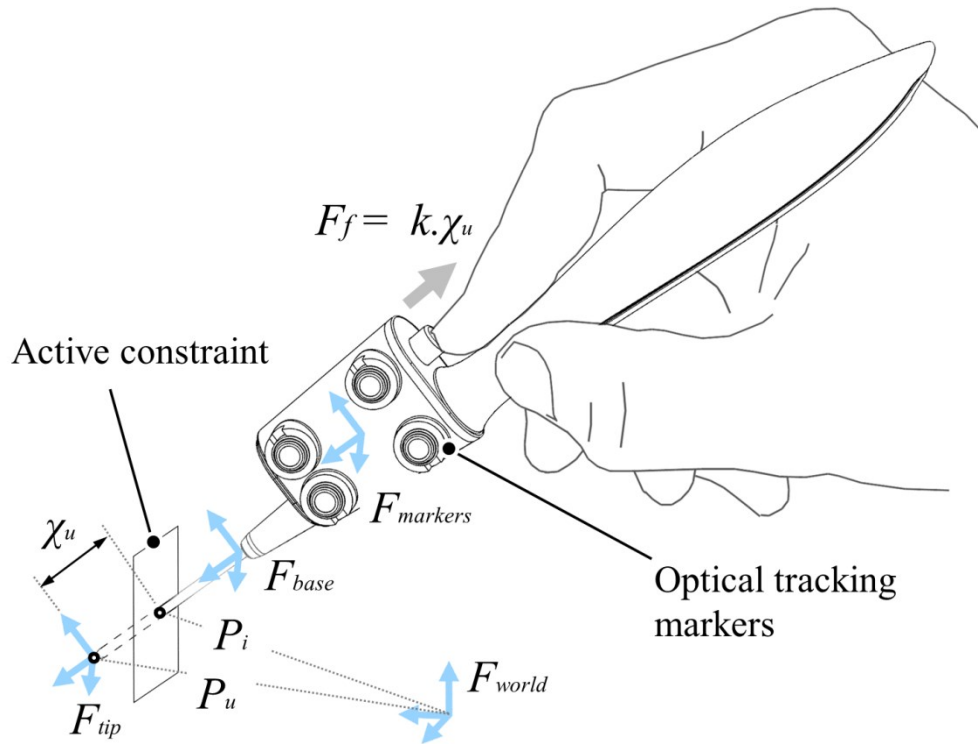


**Figure 6.4:** Shows the transformations from the world coordinate system to the uncompensated tool tip position and how the penetration distance is defined for the motion compensation. Force-feedback to the user is proportional to the penetration distance.

### 6.1.2 Mechanical Design

A linear motor (LM) (LM0830-015-01, Faulhaber) was used to actuate the surgical tool, thus providing the motion compensation. This motor was chosen for its compact size, response time, integrated position sensing and appropriate stroke length of 15mm. The magnetic shaft of the motor is co-axially mounted to the surgical tool using a coupling mechanism. An anti-rotation keyway feature is mounted onto the opposite end of the magnetic shaft. It is seated in a groove and prevents axial rotation of the surgical tool. A voice coil actuator (VCA) (LVCM-013-013-02, Moticont) was used to actuate the force-feedback display. It was chosen for its compact size relative to the peak force it can generate and its simple controllability. It has a total stroke length of 6.4mm, which is acceptable for applying loads to the fingertip of the operator. A force sensor (8438 5005, Burster) is mounted between the VCA and the force-feedback display and is used to provide closed-loop force control. These features are illustrated in Figure 6.5.

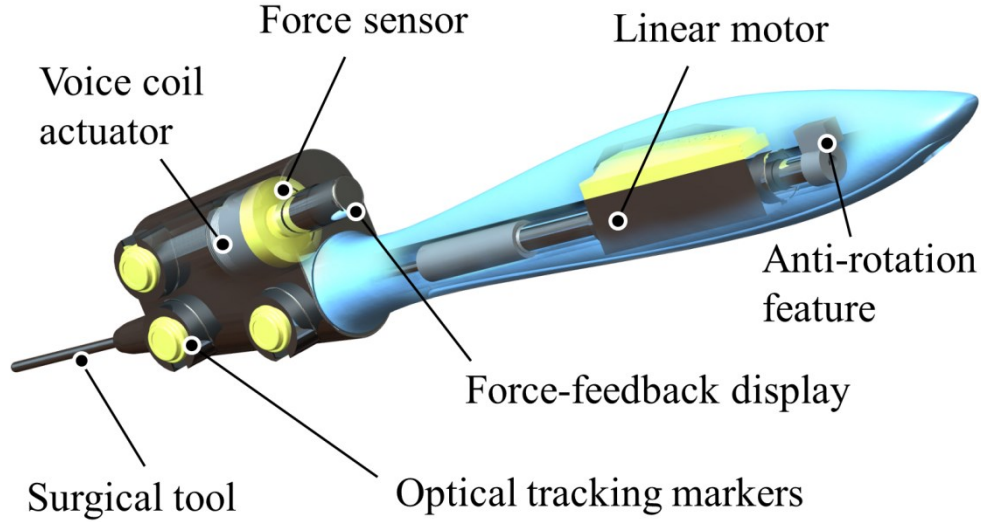


**Figure 6.5:** A ghosted CAD rendering of the hand-held the device showing the various engineering features.

### 6.1.3 Control Implementation

The coordinates of the optical markers are passed to the Optotrak Certus<sup>®</sup> acquisition hardware. A standard PC with a quad-core Intel<sup>®</sup> processor is used to receive and process the coordinate data. The active constraint is simulated in a C/C++ programming environment in order to calculate the value of the uncompensated penetration distance  $\chi_u$ . This value is then passed to a multi-threaded real-time controller (CompactRIO, cRIO-9014, National Instruments Corp.) running at 1kHz through a User Datagram Protocol (UDP) connection. The uncompensated penetration distance is then multiplied by the motion scaling factor  $K_s$ . The CompactRIO controller then outputs an analogue voltage to the LM controller (MCLM 3006S, Faulhaber) which controls the position of the LM using its own low-level position control loop. The uncompensated tool tip penetration distance  $\chi_u$  is also fed in to a force control algorithm which controls the magnitude of force that the VCA exerts on to the operator's fingertip. The distance  $\chi_u$  is multiplied by the force-feedback stiffness parameter  $k$  in order to generate a desired force setpoint. The force sensor mounted between the VCA and the operator's fingertip measures the actual force being exerted. The error between the desired force input and the measured force output is fed in to a Proportional-Derivative (PD) controller implemented on the FPGA of the CompactRIO. The output from the PD controller

is fed in to an amplifier (LSC 30/2 linear 4Q Servoamplifier, Maxon) that controls the power provided to the VCA. These control loops are shown in Figure 6.6.

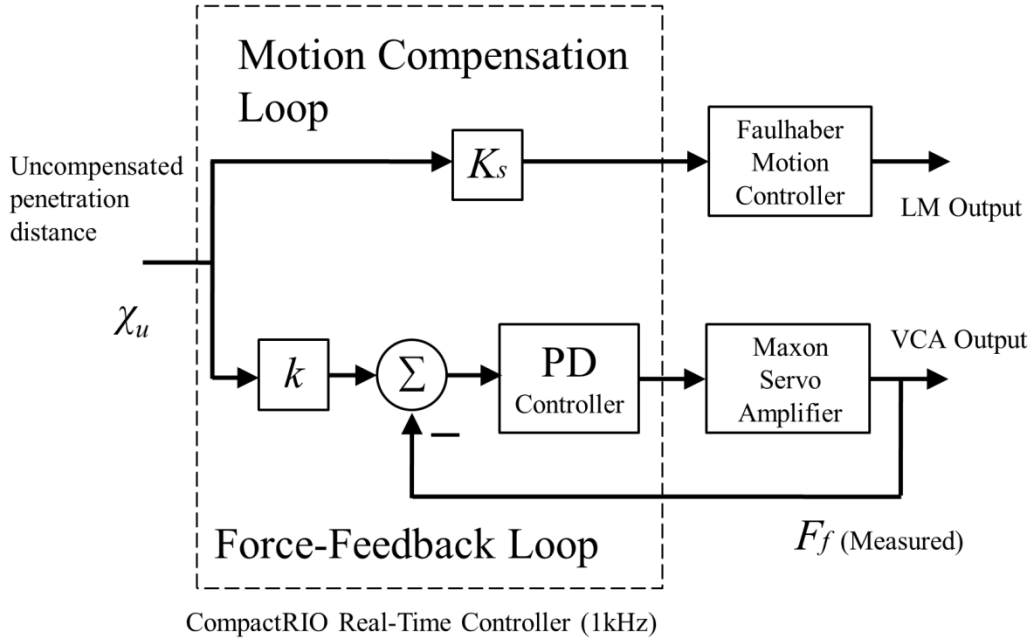


**Figure 6.6:** Shows the parallel control loops implemented on the CompactRIO responsible for the motion compensation of the tool tip and the force-feedback to the operator.

## 6.2 Bench Tests

### 6.2.1 Motion Compensation Studies

The device was evaluated in a series of bench tests in order to quantify its motion compensation and force-feedback abilities. A second rigid body containing four optical markers was attached to the surgical tool shaft so as to allow a calculation of the actual tool tip position  $P_a$  using a rigid transformation from the origin of the four additional optical markers. An initial calibration was performed in order to co-align points  $P_a$  with  $P_u$  in the world coordinate system before the experiments took place. This was done by placing the device in a jig with known geometry and zeroing the tool position when it was fully advanced. The measured distance that the tool tip had actually penetrated the active constraint  $\chi_a$  was calculated as being the distance between  $P_a$  and active constraint intersection point  $P_i$ .

The device was rigidly clamped to a test rig that excited the device linearly along the axis of the surgical tool. The device was excited with a sinusoidal input at 0.05Hz through a peak-to-peak amplitude of 16.6mm using a LM (LM1247-040-01, Faulhaber). An active constraint was positioned 2.5 mm in front of the tool tip and the measurements of  $\chi_a$  were made over time, this is shown in Figure 6.7. The device was programmed to retract the instrument tool tip for 10 scaling factors of  $\chi_u$  at 0.1 intervals between 0-1. Additionally, the device was excited with the tool rigidly fixed so as to show the trajectory of the uncompensated instrument (equivalent to the trajectory of  $\chi_u$ ) for comparison. The hand-held device was excited to an amplitude within the maximum stroke of the LM, which ensures there were no discontinuities in the motion compensation as a result of the actuation range of the device.

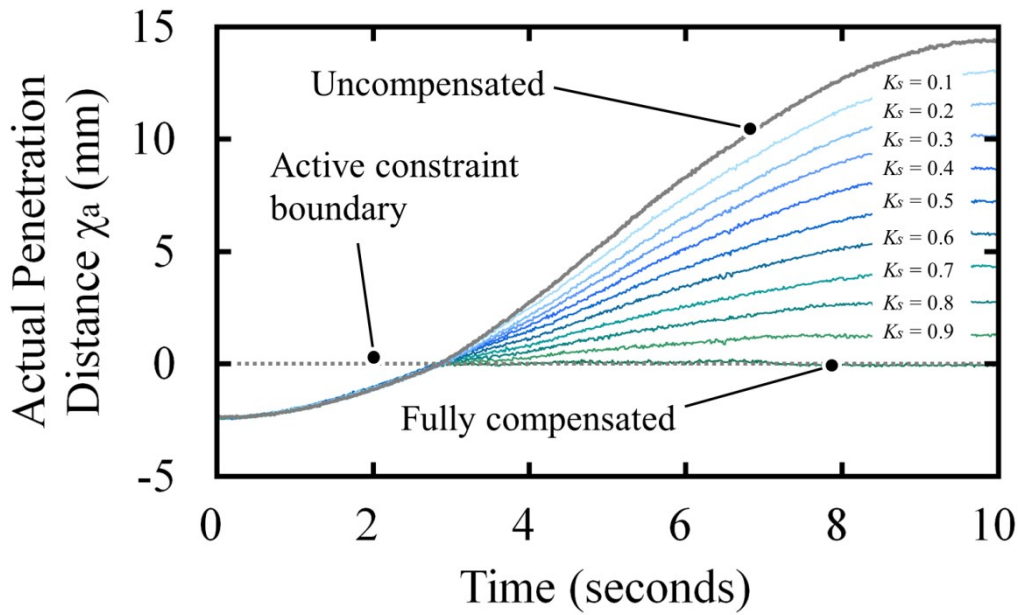


**Figure 6.7:** Plot showing actual penetration distance of the tool tip versus time for the fully compensated, uncompensated and 9 motion-scaled cases.

For each case, the trajectory starts at 2.5mm from the active constraint; the penetration distance is defined as negative when the surgical tool is approaching the active constraint. When the surgical tool penetrates the active constraint, a discontinuity in its trajectory is observed for all but the uncompensated cases. When fully compensated, the position of the

tip is held constant at the active constraint boundary threshold. A maximum penetration error of 0.13mm was observed for this case. In the motion-scaled cases, the trajectories of the tool tip can be seen as transformed sine profiles following penetration of the active constraint.

### 6.2.2 Force-Feedback Bench Tests

In a second experiment, the device was translated into an active constraint and measurements were made of the forces generated by the force-feedback display with respect to the theoretical distance  $\chi_u$ . The device was mounted onto the excitation rig as used in the previous experiment and the force-feedback display was pre-loaded with a spring to simulate the operator's fingertip. The force sensor was zeroed before each experiment and calibrated against precision weights. Figure 6.8 illustrates the force-displacement profiles for 7 settings of  $k$ . Linear regression models using the least squares method are fitted to the observed measurements in Figure 6.8 and in Table 6.1.

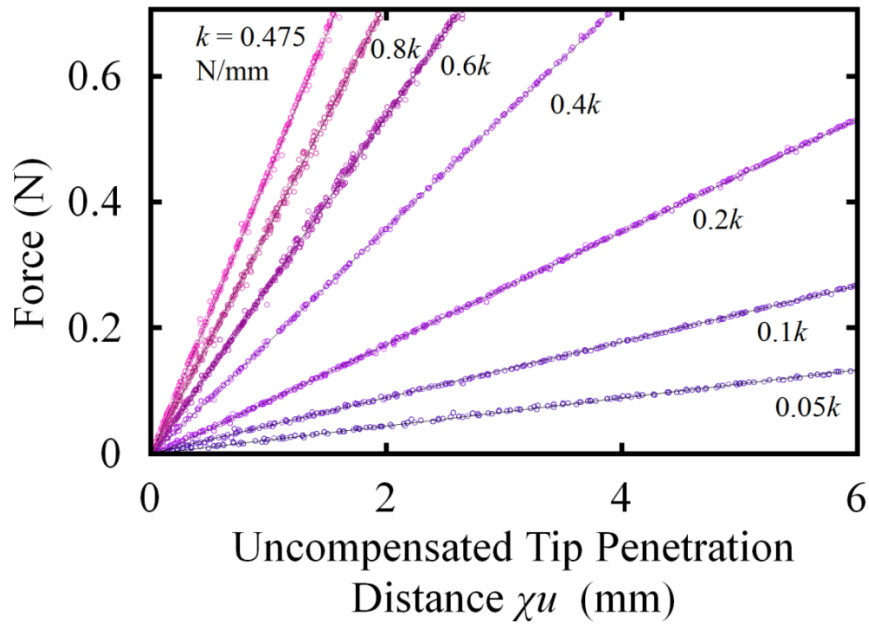


**Figure 6.8:** Plot showing force versus the uncompensated tool tip penetration distance for 7 different stiffness factors.

**Table 6.1: Linear regression models for the measured stiffness coefficients using the least squares method.**

| Desired $k$ (N/mm) | Regression Ratio (N/mm) | Residual Force (N) |
|--------------------|-------------------------|--------------------|
| 0.024              | 0.022                   | 0.000              |
| 0.048              | 0.045                   | 0.000              |
| 0.095              | 0.090                   | -0.005             |
| 0.190              | 0.180                   | -0.003             |
| 0.285              | 0.270                   | -0.003             |
| 0.380              | 0.358                   | 0.000              |
| 0.475              | 0.449                   | -0.002             |

It has been shown that the device was capable of accurately reproducing a range of stiffness factors which could be used to create rigid and soft constraints as the operator collides the device with an active constraint. A linear relationship between force and uncompensated tip penetration distance is observed. This is true up until the point that the VCA saturates, after which point a discontinuity is introduced into the force-feedback response. At higher stiffness factors, the motors saturate after only a short penetration distance. The current limits set in this experiment meant that the VCA saturated at around 0.8N. The linear regression models fitted to the experimental data showed that the observed values of  $k$  were less than the pre-set value by an average of 5.7%. This can be attributed to the absence of integral control in the force-feedback control scheme.

### 6.2.3 Combined Active Constraints and Force-Feedback Studies

To illustrate the combined motion compensation and force-feedback abilities of the device, a freehand palpation of a spherical active constraint was performed with a single user. Figure 6.9 shows an operator interacting with both a rigid (fully motion compensated) and Figure 6.10 shows an interaction with a soft (motion scaled, where  $K_s = 0.5$ ) active constraint. A force-feedback stiffness factor of 0.19N/mm was adopted for these interactions.

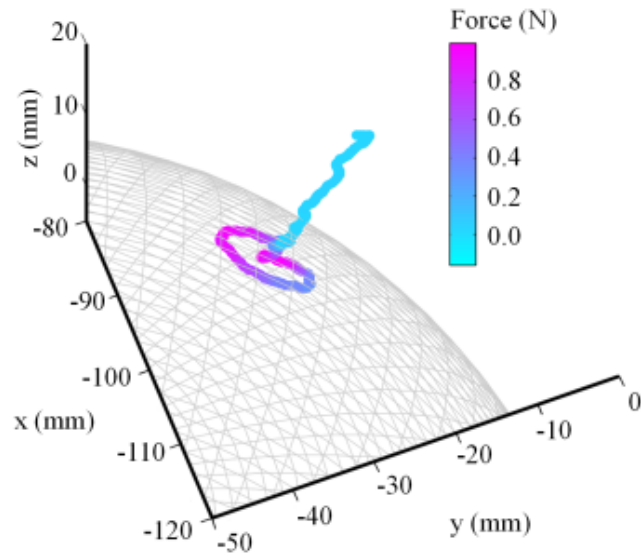


**Figure 6.9:** Plot showing the actual tool tip position  $P_a$  in a free hand palpation of a spherical active constraint for a rigid active constraint in which the tool tip motion is fully compensated and scaled respectively.

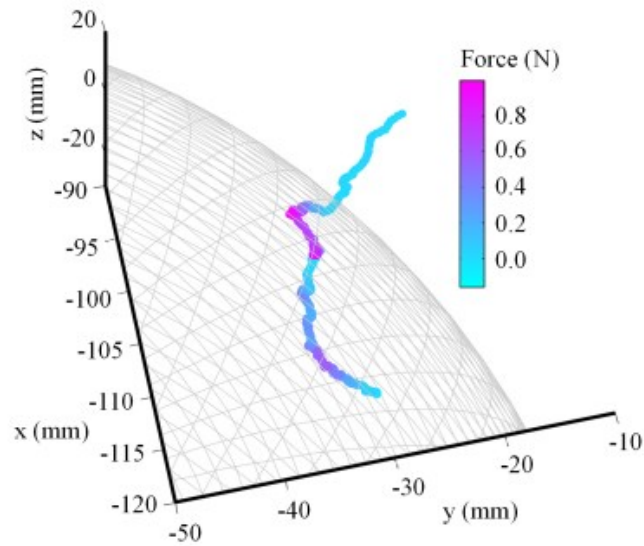


**Figure 6.10:** Plot showing the actual tool tip position  $P_a$  in a free hand palpation of a spherical active constraint for a soft active constraint.

The colour overlay represents the force being exerted on the operator as the active constraint is palpated. In both cases, the tool tip approaches the active constraint with no force being exerted on to the operator. As the uncompensated tool tip position fluctuates, so does the force being exerted onto the operator. When the rigid positional constraint has been implemented, the tool tip motion is constrained to the surface of the spherical active constraint. In the motion-scaled case, it can be seen that the tool trajectory penetrates beyond the constraint boundary in proportion to the motion scaling factor. Additionally, the force-feedback is proportional to the distance penetrated by the surgical tool, so that the operator experiences a spring-like response from the active constraint.

### **6.3 Active Constraint Boundary Detection User Study**

A user study was conducted in order to quantify the performance of the combined motion compensation and force-feedback abilities of the hand-held device with respect to user performance. 8 right-handed users (4 female and 4 male) of ages 24 to 35 with no clinical experience were recruited for this study. Each user gave their prior consent to take part in this study. The users were asked to hold the hand-held device in a stationary position whilst an active constraint was translated on to the tool tip so as to cause a virtual collision and generate force-feedback on to the user's fingertip. The penetration depth threshold at which users could perceive the active constraint was measured when the device was operated with and without the tool tip motion compensation. A fully compensated (rigid) position constraint was imposed for the compensated case in this study. The same hand-held device as used in the bench tests that incorporated the additional optical markers that could track the tool directly was adopted in this user study. An additional rigid body containing optical markers was mounted on to a LM (LM1247-040-01, Faulhaber) and translated towards the hand-held device at a constant velocity of 9mm/s towards the tool tip. The LM shaft was constrained so as to only allow linear translation of the rigid body containing the optical markers. A constant velocity profile was chosen so that all virtual collisions would occur at the same velocity across the user study and prevent biasing of the results. This would not have been the case if users had instead been requested to translate the device towards a stationary active constraint. Additionally, this would potentially cause users to learn the position of the active constraint through their proprioception sense. The device was positioned to make contact with the translating active constraint at approximately halfway through its stroke. The experimental setup is shown in Figure 6.11.

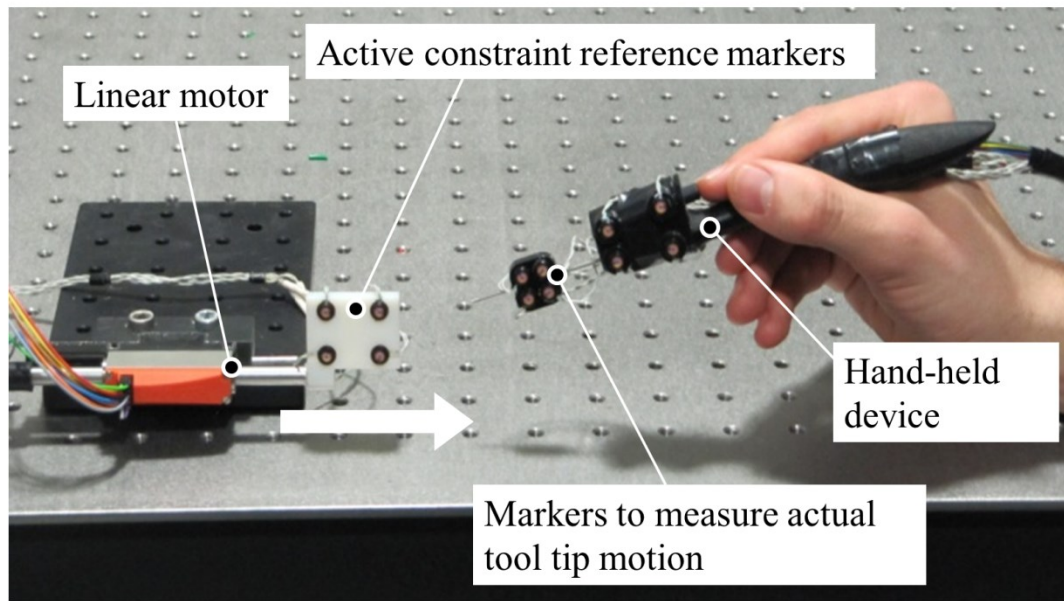


**Figure 6.11:** Shows the experimental setup. The hand-held device is held stationary by the operator whilst the LM translates the active constraint on to the tip of the device so as to cause a collision. The optical markers are tracked using the Optotrak Certus<sup>®</sup> system which is not shown.

A stiffness factor of  $k = 0.475\text{N/mm}$  was implemented in the force-feedback control algorithm for both cases. Since the velocity of the active constraint was translated at  $9\text{mm/s}$ , neglecting physiological tremor, on average the VCA would reach its peak force  $0.19$  seconds after the tool tip penetrates the active constraint. This ensured a near instantaneous sensation to the user, which reduced the time for them to perceive the active constraint. Use of a lower stiffness factor would bias the study in favour of the motion-compensated case. The users were required to depress a push button immediately after they felt the force-feedback from the device. For each collision detection, the instantaneous actual penetration depth  $\chi_a$  was acquired through a DAQ card (NI-6221, National Instruments Corp., USA) and then passed to the multi-threaded real-time controller (cRIO-9014, National Instruments Corp.) which returned the data to a host PC. The data was logged from the host PC at a rate of  $100\text{Hz}$ . The users were required to perform five runs for both the compensated and uncompensated cases. The motion compensation was switched on and off randomly between each run without the user's knowledge so as to prevent biasing of the results. Users were requested to not observe the experiment area so as to be wholly dependent on their sense of

touch in perceiving the active constraint. The results of the user study are presented as a box plot in Figure 6.12 and Table 6.2

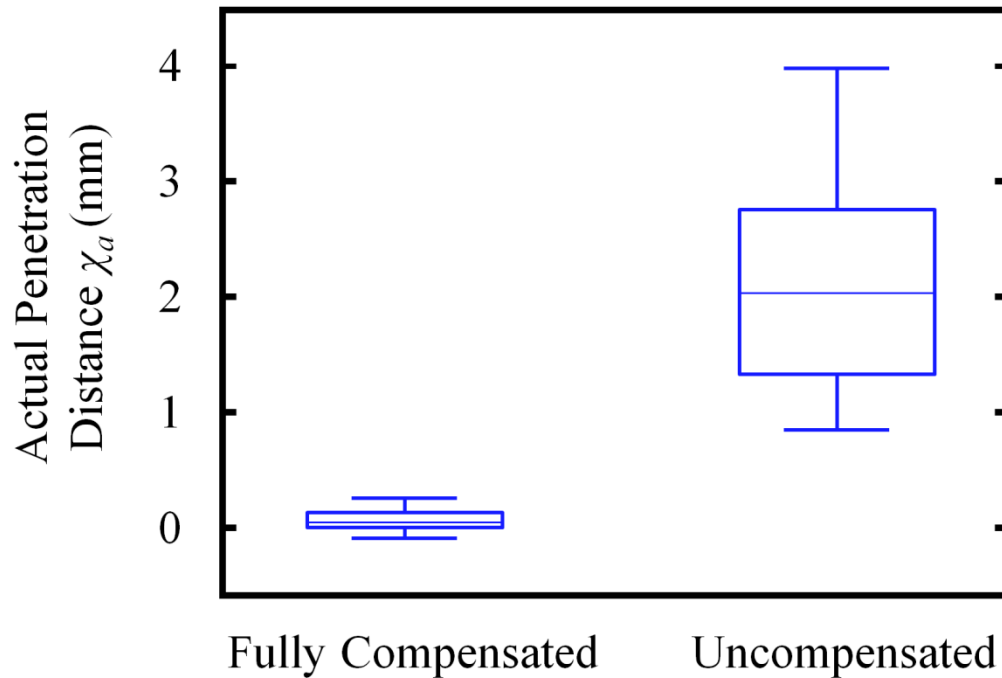


Figure 6.12: A box plot showing thresholds for the actual distance penetrated by the tool tip following the user's indication that they had perceived the active constraint for both fully compensated and uncompensated cases. Boxes indicate the lower quartile, upper quartile and median, whiskers represent the minimum and maximum values.

Table 6.2: Key statistics for the active constraint boundary detection user study.

|                                   | Fully Compensated | Uncompensated |
|-----------------------------------|-------------------|---------------|
| Minimum $X_a$ (mm)                | -0.11             | 0.83          |
| Maximum $X_a$ (mm)                | 0.27              | 3.99          |
| Median $X_a$ (mm)                 | 0.05              | 2.04          |
| Interquartile range of $X_a$ (mm) | 0.13              | 1.43          |

## 6.4 Discussion and Conclusions

It has been shown that the use of motion compensation reduced the actual penetration distance of the tool tip significantly. The median penetration distance for the compensated case was 0.05mm and for the uncompensated case, 2.04mm. The distribution of penetration distances from these 2 cases was shown to be significant ( $p < 0.001$ ) through a statistical comparison test (Mann-Whitney U). The peak penetration distance measured for the compensated case (0.27 mm) is lower than the lowest measured penetration distance for the uncompensated case (0.83mm). No trends were observed to suggest an improvement in performance over time by the users. Another salient point of discussion that can be observed from Figure 6.12 is the considerable differences in the variability of the measured tool penetration distances, with the motion compensation providing greater consistency. The interquartile range of the penetration compensated case (0.13 mm) and uncompensated case (1.43 mm) highlight this point. These results indicate the ability of the hand-held device to respond faster than its human operator. The variability in the compensated cases can be largely attributed to latency in control of the instrument tool tip. The large variability observed in the uncompensated case can be attributed to a number of human factors, although most significantly, the reaction time in perceiving, cognitively processing the active constraint and mechanically depressing the push button. For this reason, the results observed for the uncompensated case represent an overestimate for the actual penetration depth thresholds that were perceived through the force-feedback display. This is because the active constraint was always advancing towards the tool tip, whereas in the compensated case, the relative motion between active constraint and tool tip could be advancing or receding from the active constraint boundary. For the same reasons, quantisation errors in the measurements will also increase the measured penetration distances in the uncompensated case.

In addition to variations in the intrinsic reaction times between users, other factors explain the large variations in the perceived active constraint penetration distance thresholds. Any pre-loading of the force-feedback display will have reduced the fidelity of the tactile sensation felt when the tool tip penetrated the active constraint. The reaction time is also dependent on the concentration of the operator in the instant in which they perceived the active constraint. Differences in the extent of erroneous physiological motion between users may also explain the inter-user variability in both the compensated and uncompensated

cases. High amplitude physiological tremor or other high frequency motions cannot be fully compensated for by the control scheme implemented in this study.

This chapter has proposed a novel ungrounded hand-held device for providing active constraints with force-feedback. The hand-held device has been quantified with respect to its motion compensation and force-feedback abilities in bench tests. It has also been validated in terms of its ability to indicate interaction with an active constraint to a user through a controlled user study. It is expected that this work can be extended to demonstrate tool guidance of an ungrounded, hand-held instrument through pre-defined active constraints. Whilst implementation of active constraints with an ungrounded hand-held device has previously been demonstrated [51][48], this work demonstrates that force-feedback allows intuitive perception of the active constraint by the user. The force-feedback could thus provide a prompt for the user to indicate when they are in close proximity to an anatomically-critical area. Additionally, knowledge of the active constraint boundary is important in preventing the operator exceeding the range of the motion compensation.

# **Chapter 7: Conclusions and Future Research Directions**

## **7.1 Contribution of this Thesis**

This thesis has specifically examined haptic feedback strategies for ungrounded hand-held surgical devices. These strategies have been manifested in a series of haptic hand-held device prototypes that have leveraged different techniques to improve human haptic perception of surgical interaction forces and pre-defined “active constraints”. The development and evaluation of these devices has resulted in a number of technical, clinical and scientific contributions:

- Demonstration of improved human haptic perception through a novel, ungrounded force-amplifying device that provides haptic feedback directly to the surgeon's fingertip.
- Development of a device capable of improving grasping force perception in microsurgical forceps through force amplification. Demonstration of improved human haptic perception and detailed bench tests that examine the force sensing, amplification and dynamic response.
- Implementation of force-limiting using vibrotactile feedback in a hand-held device. Systematic analysis of human perception to vibrotactile force-threshold feedback in reducing force application under different distracting conditions. Demonstration of force-limiting for microsurgical dissection in a pre-clinical human cadaveric study on a human brain.
- Development of a system that can simultaneously provide haptic and position-based active constraints to prevent undesirable tool-tissue collisions. Demonstration that this proposed methodology can improve human haptic perception of a virtual boundary.

There is significant evidence to suggest that improving a surgeon's haptic perception of mechanical tool-patient interaction can improve the efficacy [97][101][99] or safety of an intervention [18][168]. Hand-held surgical tools allow the surgeon to directly perceive these important tool-patient manipulation forces. In very delicate surgical procedures, particularly microsurgical interventions, the tool-patient forces can be beyond the range of human haptic acuity and thus cannot be felt by the surgeon. Chapter 3 proposed a force-amplifying device that was devised to augment tool-tissue interactions for micromanipulation tasks in delicate interventions. The proposed device could display the amplified tool-tissue forces on to the operator's fingertip, so as to avoid the need for a bracing mechanism [144]. Bench tests verified the intrinsic amplifying ability of the device; these tests demonstrated that the device could scale force by up to a factor of 15. A user study was performed to illustrate that the force-amplifying strategy could improve human haptic perception. The study performed in this chapter highlighted that forces other than those exerted on to the tool tip influence user haptic perception. This point alludes to the psychophysical model known as Weber's law

which implies that a heavier hand-held device would diminish a surgeon's haptic perception of the forces applied to the tool tip.

The ungrounded, hand-held force-amplifying strategy proposed in Chapter 3 was further evolved in Chapter 4. This chapter examined force amplification for a case in which human haptic perception is distorted by an additional force, unrelated to the tool-tip force. The proposed strategy explicitly examined the case of microsurgical forceps that operate using an elastic spring return which is orders-of-magnitude greater than the forces exerted by the forceps tips in manipulating delicate tissue. Two iterations of the microsurgical forceps designs were presented. These designs sought to address some of the technical limitations of the force-amplifying device proposed in Chapter 3. One such limitation was the poor force sensing resolution at the low force levels, below the human threshold for touch, where force amplification has the greatest potential. The microsurgical forceps designs made use of a silicon strain gauge sensor that was integrated in to the proposed designs. Simulations were performed to assist the design of these devices and calibration techniques were employed to improve the intrinsic force-amplifying capabilities of the developed prototypes. Bench tests demonstrated that the improved force sensing significantly enhanced the force-amplifying performance, with up to scaling factors of 80 being reported. This chapter also examined the bandwidth of the actuation system which must be capable of rendering the forces within a frequency range of surgical interventions involving hand-held devices. A user study was performed to demonstrate the efficacy of the haptic feedback strategy in augmenting human perception of tool tip force in the presence of a contaminating force. The results demonstrated a significant improvement in user perception and it was proposed that such a haptic feedback strategy might have potential applications in neurosurgery and vascular microanastomosis.

The greatest potential applications for the strategies proposed in Chapters 3 and 4 are in delicate microsurgery, where the surgical manipulation forces are below a threshold that human subjects can perceive. The proposed strategies relay the tool tip forces directly to the surgeon's fingertip so as to highlight the important surgical interactions above other contaminating forces that diminish human haptic acuity. Thus, if Chapters 3 and 4 were concerned with *what* force, Chapter 5 addressed the question of *how much* force. Whilst the

studies of Chapters 3 and 4 demonstrate that such haptic augmentation strategies could allow the surgeon to apply less force, they cannot inform the surgeon as to when tissue injury is imminent. Chapter 5 investigated a force-threshold haptic feedback strategy for a hand-held device. A dissecting instrument was proposed that could sense lateral load and inform the user as to when a force limit had been breached using vibrotactile feedback. Whilst the strategies postulated in the prior chapters were discussed in abstract terms, Chapter 5 introduced the clinical exemplar of microneurosurgery for which such a haptic feedback strategy might prove effective. This clinical application was chosen because the consequences of exerting excessive force in neurosurgery is profound and the capacity for regeneration of neurovascular tissue is limited [159]. The force sensing elements of the device were simulated and calibrated to meet the requirements of neurosurgery. Psychophysical user studies demonstrated the efficacy of the device in consistently limiting the forces applied by users when performing tasks with distracting factors representative of different surgical scenarios. A cadaveric study demonstrated the potential clinical value of such an approach. All of the studies demonstrated that the device could aid human subjects limit the force and the proposed methodologies are of potential use to researchers investigating novel haptic feedback methods and/or clinical applications. The studies in this chapter demonstrated that the force-threshold feedback strategy is also effective for clinical scenarios that involve supra-threshold forces.

The strategies investigated in Chapters 3, 4 and 5 considered surgical scenarios in which tool-tissue contact was augmented by some means. The haptic feedback strategy examined in Chapter 6 investigated the use of active constraints to prevent tool-tissue contact altogether. This principle has most commonly been investigated in the context of grounded robots that provide haptic interaction between the operator and the surgical tool so as to constrain the tool effector to, or away from pre-defined regions. An ungrounded mechatronic device was proposed that could actively reposition the tool effector to achieve the same means of virtual contact avoidance. The novel contribution of this chapter was in providing ungrounded force-feedback to the operator's fingertip to alert them as to when the active constraint had been encountered. The force-feedback mechanism itself was an evolution of the designs presented in Chapters 3 and 4. The device was evaluated with respect to its motion compensation and force-feedback abilities. A user study demonstrated the ability of the device to successfully and consistently alert the operator that they had encountered an active constraint before the tool tip could penetrate the boundary.

## 7.2 Future Research Directions

As is the inherent nature of research, for all of the questions that have been addressed in this thesis there are as many questions that remain. These can be broadly stratified in to technical challenges, new experimental protocols and clinical translation. There are also a number of recurring technical considerations common to all of the devices presented in this thesis. One is that of miniaturisation and weight minimisation. It has been acknowledged that increasing the device weight diminishes haptic perception, but it also potentially exacerbates tremor through muscle fatigue [34]. Therefore, future device implementations need to focus on minimising weight and systematic ergonomics studies should be performed in the optimisation of mechatronic hand-held device design. Experiments that quantify the effects of increased size and weight of hand-held instruments on surgical performance are an important first step. For some specific cases, it may be that ergonomic, lightweight instrument design can outweigh the enhancements of mechatronic augmentation.

Another limitation of the devices proposed in this thesis is the deficiencies in actuation and sensing DoFs which have largely been constrained to single or two DoF systems. In the case of the device presented in Chapter 3, Stetten *et al* argue that the axial force-sensing DoF is most significant because lateral loads are effectively magnified through the mechanical lever effect. This is true for open interventions, such as cardiac surgery, otorhinolaryngological surgery, neurosurgery and corneal surgery. However, in retinal microsurgery, the incision point in the sclera serves to contaminate proximal force sensing techniques which has prompted the design of optical-fibre based sensing methods that are integrated in to the tool shaft so as to capture the true tool tip forces. The force-amplifying device presented in Chapter 3 has already been evolved to incorporate lateral sensing of the tool tip as well as axial force sensing. This proposed device is akin to the force-sensing hand-held microsurgical tool presented in [137], it is shown in Figure 7.1. The forceps designs presented in Chapter 4 focused on grasping force; however the lateral forces that are applied are of potentially equal importance. Indeed, microvascular surgeons apply lateral forces to accurately tension sutures and drive micro needles through fragile vasculature wall. The design proposed in Chapter 5 incorporated lateral force sensing but lacked axial sensing. Whilst it is hypothesised that the blunt dissection task that was examined involves mainly lateral force application, this hypothesis has not yet been confirmed. The technique used with hooked sharp dissection is considerably likely to involve axial force application. The

ungrounded active constraint system proposed in Chapter 6 could also benefit from incorporating additional DoFs in to the motion compensation manipulator. The single DoF system presented is simple and exploits the existing dexterity of the surgeon's hand. However, the single DoF mechanism cannot compensate for lateral or highly oblique active constraints which are likely to feature prominently in clinical applications such as otorhinolaryngology. A solution akin to the latest implementation of the tremor suppression device Micron, that uses a 6 DoF Stewart-Gough architecture could precisely navigate such constraints. However, a significant challenge with multi DoF force sensing and actuation systems is the increased cost and complexity that hand-held mechatronic solutions seek to avoid. Emerging manufacturing techniques such as pop-up MEMS are one potential solution in fabrication of robust, low cost, mass-producible sensing and actuation mechanisms [169][170]. Such miniaturisation technologies may also prove fruitful in facilitating weight reduction of hand-held mechatronic elements.

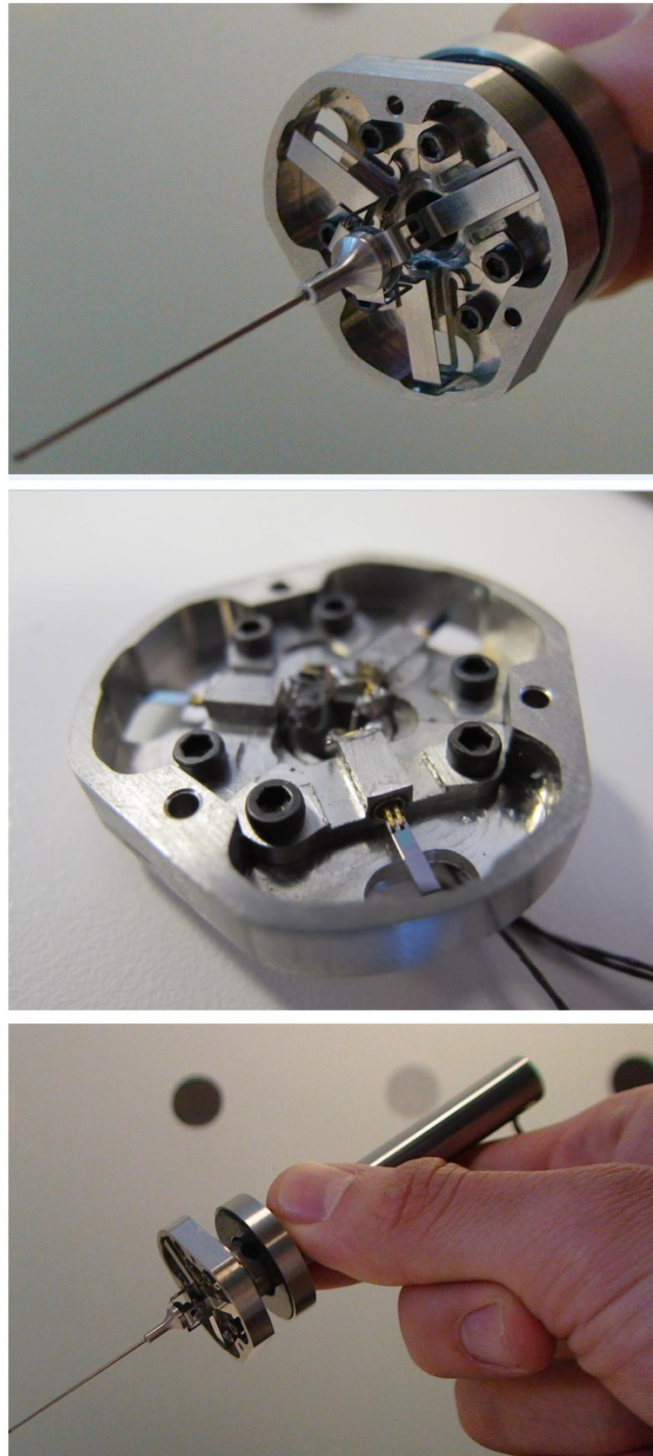


**Figure 7.1:** Photographs showing a future, 3 DoF embodiment of the hand-held force-amplifying device presented in Chapter 3.

A technical challenge of note that has not been addressed in this thesis is the impact of motion artefacts that arise as a result of integrating electromechanical actuators in to hand-held surgical instruments. The force amplification devices in particular, will recoil as a result of the reaction forces being transmitted through the surgeon's grip of the devices being manipulated. The use of smaller actuators is one way to mitigate such a problem; this was a primary motivation for the miniaturisation efforts presented in Chapter 4. Nonetheless, the motion artefacts generated by on-board actuators in mechatronic hand-held surgical devices should be formally analysed in future work. A related technical issue that has also not been addressed thus far is the weight and inertial artefacts that are sensed by the surgical effectors as the hand-held instrumentation is rotated and translated by the operator. One simple solution is to filter out high frequency tremor artefacts electronically. Gravitational effects and other inertial loads could be compensated using inertial measurement units (IMUs). This problem has been investigated by Gilbertson and Anthony in the context of ultrasound probe force control [78], but not in the context of microsurgical mechatronic instrumentation. With respect to Chapter 5, the means by which force-threshold feedback is provided could also be further investigated. For example, the vibrotactile feedback method could be adapted to display stratified force-threshold by modulating the vibration frequency for multiple force-thresholds. There is also scope to investigate other means of sensory substitution, for example, ongoing work in the Hamlyn Centre is investigating the use of augmented reality visual feedback [171] in conjunction with neurosurgical dissecting instrumentation.

In addition to technical improvements, the experimental protocols adopted in this thesis, whilst having demonstrated the effectiveness of the equipment developed, could be improved to provide more accurate quantification of intrinsic human perception. The user studies of Chapters 3 and 5 represented overestimates of the minimum force threshold that could be perceived because of the time delay factors. The method proposed by Stetten *et al* [144] is akin to the method-of-adjustment [172]: the user study is designed such that each user adjusts the force applied to the instrument tip through a magnetic levitation system. This method provides a solution to the challenges faced in this thesis but it is not free of biases altogether. The method-of-constant-stimuli [172] may prove a more reliable technique for assessing the absolute performance of hand-held haptic augmentation device. This protocol would involve providing a force stimuli of constant magnitude to the hand-held device and

asking the subject if they could perceive the force for each instant. The method is more statistically rigorous and would provide a better evaluation of force threshold perception.

Further experiments would also allow other psychophysical parameters to be evaluated. For example, Wu *et al* have recently demonstrated the potential of a force-magnifying hand-held device to aid the discrimination of different supra-threshold forces [173]. The same authors also demonstrated improved stiffness perception using a force-magnifying hand-held device. A further finding was that the perceptual effectiveness of such a force-feedback approach diminishes with stimulus intensity [173]. How these findings relate to practical clinical scenarios is yet to be determined. Klatzky *et al* have also recently postulated that microsurgical tasks that involve surface puncture may benefit from haptic augmentation but post-puncture control is also a significant challenge [174]. This alludes to a more general point regarding haptic feedback hand-held devices. The provision of haptic feedback to the clinician is only useful if it is within their power to control the surgical device in question. If the tool-tissue forces that result from the surgeon's tremor approach the forces at which iatrogenic injury occur, then haptic feedback of any kind is of limited use. The sub-field of hand-held tremor suppression devices is one such solution and their integration with micro-scale force sensing/feedback systems is an ongoing research trend. The work of Gonenc *et al* considers the use of active tremor suppression in conjunction with force-threshold feedback which is one potential technical solution [140].

The extent to which visual cues aid haptic perception in microsurgical tasks is another point of investigation that should be considered in future experimental work. Wagner *et al*, for example, demonstrated that robotic telemanipulation of lower stiffness objects resulted in improved haptic perception compared to higher stiffness objects that deform less [99]. The successful clinical uptake of the da Vinci™ robot is further evidence that visual feedback of the surgical view is a sufficient surrogate for intrinsic haptic perception. This same principle is likely to be applicable to hand-held surgical devices, which potentially negates for haptic augmentation mechanisms under certain conditions. Nonetheless, there are a number of scenarios that should be investigated. Haptic augmentation may for example, improve the surgical performance for less experienced clinicians. It may also be the case that haptic augmentation systems are better suited to scenarios where visual feedback is impaired. Lee

*et al* postulate that hand-held haptic augmenting devices may be of greater use in cardiac surgery, where blood obscures the surgeon's view of the operating scene and thus increases their reliance on haptic perception [145]. Whilst visual feedback can indicate tool-tissue contact and approximately indicate to the surgeon how safe and/or effective they are performing, the information that is obtained by the surgeon is still qualitative. This gives credence to the adoption of force-threshold haptic feedback over force amplification, even when visual feedback is not necessarily diminished. The work of Gonenc *et al* has gone some way to demonstrate this point in studies that have compared freehand retinal peeling tasks with and without auditory force-threshold feedback, where manipulation forces are sub-threshold and visual feedback is provided to the users [140]. Nonetheless, further studies could assess the more fundamental psychophysical aspects of combined visual and haptic feedback.

With respect to the user study in Chapter 6, future work will investigate the use of force-feedback to aid navigation of an active constraint path. For this to work, it is likely that additional degrees-of-freedom would need to be rendered back to the operator to provide a direction to the tactile cue. Whilst there is merit in adopting a simple, single degree-of-freedom motion compensation system, additional degrees-of-freedom of the surgical tool would enhance performance, for example, when the tool approaches an oblique surface. Whilst force-feedback has been provided in this study, there is also the potential for sensory substitution such as auditory feedback to indicate proximity to an active constraint. This work could also be extended to incorporate dynamic active constraints [175][176] when physiological motion needs to be compensated for. This would allow autonomous tracking of the tool tip and permit high-level guidance of the instrument through the use of force-feedback.

Technical engineering advancements and psychophysical experimentation undoubtedly hold scientific value; however, the goal of clinical translation is perhaps the most significant future direction for the concepts proposed in this thesis. One of the major insights was the demonstration that force-threshold limits can effectively assist the surgeon operating a hand-held microsurgical device, even at supra-threshold force application levels. An interesting future direction for this strategy would be the investigation of other clinical applications in

which distinct force-thresholds associated with iatrogenic injury exist. Even clinical applications that require manipulation forces one or two orders of magnitude greater than the microneurosurgery exemplar could potentially benefit from such a haptic feedback strategy. In cases in which the surgical manipulation forces are well perceived, the surgeon can nonetheless never absolutely know how much force is being applied. There are numerous clinical scenarios in which force-threshold feedback might prove useful, including endovascular stent deployment, laparoscopic surgery, cochlear implant insertion and orthopaedic drilling. These potential clinical applications and the manipulation forces involved are presented in Figure 7.2 and Table 7.1. For many such interventions, the force sensing elements do not need to be located at the instrument tip, which would circumvent many of the sterilisation and cost issues which make clinical translation more challenging. The latter two applications mentioned have additional potential on the basis that haptic cues from visual feedback are limited because either the surgical view is obscured, or the anatomy is rigid.

Whilst the advantages and efficacy of force-threshold feedback has been demonstrated and discussed, it is evident that the principle relies on a force-threshold to be pre-determined. The work presented in Chapter 5 relies heavily on the data that was acquired by Marcus *et al* in defining the iatrogenic neurovascular tissue injury thresholds [159]. Future research should therefore focus on obtaining clinical data that correlates the extent and duration of force application with iatrogenic injury for a range of surgical interventions. One study did assess histological damage of colon tissue undergoing controlled stresses [177] and researchers have measured the tear threshold for retinal peeling [18] but basic science research into tissue injury remains sparse.

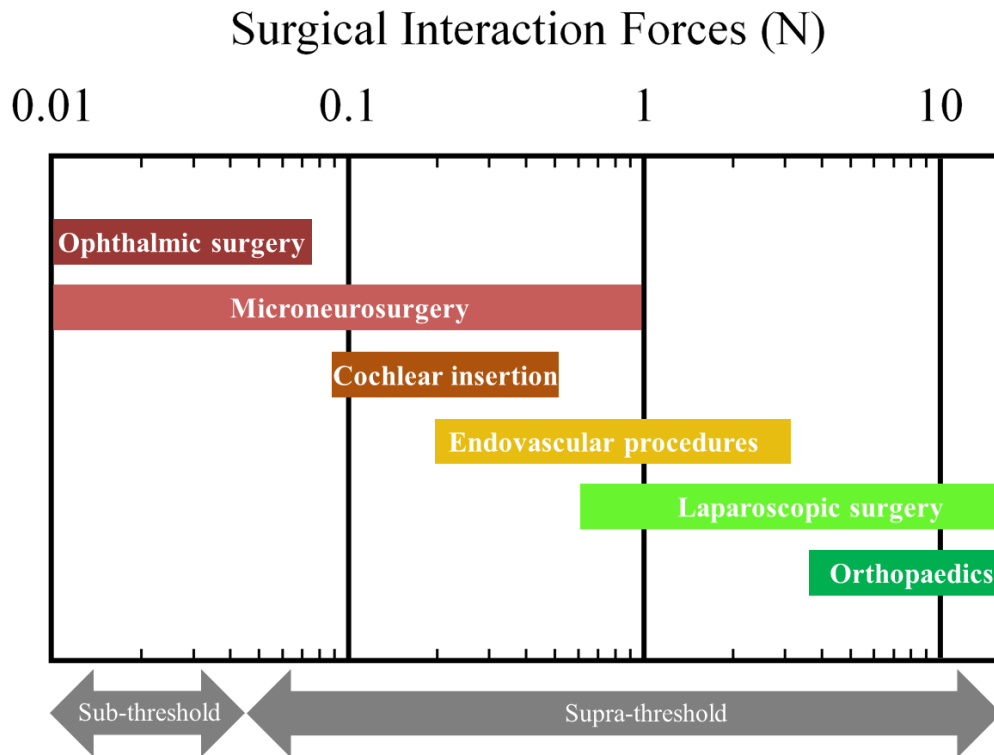


**Figure 7.2:** A schematic demonstrating the differing force levels exerted for different interventions.

**Table 7.1:** Force ranges for different interventions that may benefit from hand-held haptic feedback.

| Medical Intervention       | Approximate Force Range | References                                                                          |
|----------------------------|-------------------------|-------------------------------------------------------------------------------------|
| Ophthalmic surgery         | 0 – 7.5mN               | Gupta <i>et al</i> [90],<br>Uneri <i>et al</i> [18]                                 |
| Microneurosurgery          | 10 – 1000mN             | Marcus <i>et al</i> [159]                                                           |
| Cochlear implant insertion | 90 – 420mN              | Kontorinis <i>et al</i> [178]                                                       |
| Endovascular intervention  | 210 – 3030mN            | Rafii-Tari <i>et al</i> [179]                                                       |
| Laparoscopic surgery       | 0.61 – 18.3N            | Hwang <i>et al</i> [158],<br>Cundy <i>et al</i> [180],<br>Fakhry <i>et al</i> [130] |
| Orthopedic bone drilling   | 3.5 – 12.3N             | Lee <i>et al</i> [181]                                                              |

Another method by which a force-threshold could be determined is through skill assessment. For example, Cundy *et al* correlated traction force application levels in laparoscopic surgery with operator skill [180]. By setting the force-threshold to the levels that experts typically apply, surgical trainees could potentially learn to perform surgical interventions more delicately and more effectively. A natural extension of this concept would be to devise a means by which such a force-threshold might be modulated according to the surgical context. Intelligent *in situ* sensing of the interventional environment, such as in the techniques described by Brett *et al* [72][74], may provide one avenue for real-time modulation of the haptic feedback limits.

The very starting point for hand-held haptic feedback surgical devices is the development of force-sensing surgical tools that can be deployed *in vivo*. Knowledge of the magnitude and nature of surgical intervention forces is essential in order to ascertain *when* and *what* haptic feedback might prove effective. This is not a trivial task; the practicalities of integrating sensors in to instruments that can be trialled clinically require careful consideration of ergonomics, failure analysis and sterilisation. Nonetheless, there is precedent for surgical force sensing in clinical use: in endovascular intervention [7], in cochleostomy [73] and in laparoscopic surgery [130], to name a few. With the collation of clinical force data, the development of effective hand-held haptic feedback surgical devices may soon become a reality.

# Appendices

## I. AE801 Silicon Strain Gauge Sensor Specifications

The technical specifications for the strain gauge sensor used in Chapter 4 (AE801, Kronex, Oakland, USA) are provided in Table A1 and a schematic is provided in Figure A1.

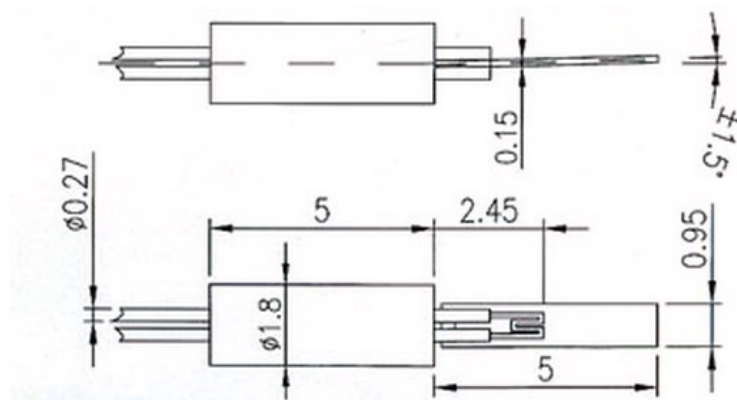


**Figure A1: Schematic of the AE801 showing the key dimensions. Image and data provided, with permission from Kronex Technologies Corporation (Oakland, USA) [www.kronex.com](http://www.kronex.com).**

**Table A1: Key parameters for the AE801 force sensor. All data provided by Kronex Technologies Corporation (Oakland, USA) [www.kronex.com](http://www.kronex.com).**

---

|                                                 |                                                |
|-------------------------------------------------|------------------------------------------------|
| <b>Input</b>                                    |                                                |
| Voltage excitation                              | 1-10V DC or AC                                 |
| Resistors Value                                 | $1000\Omega \pm 25\%$                          |
| Resistance matching                             | $\pm 10\%$                                     |
| Deflection of the beam at full scale (FS)       | $\sim 70\mu\text{m}$                           |
| Load at tip of the beam at FS                   | 0.12N (12.2gram)                               |
| Safe overload                                   | 120% of FS                                     |
| Frequency response                              | DC - 5kHz                                      |
| <b>Output</b>                                   |                                                |
| Full scale output (FSO)                         | $30\text{mV/V} \pm 30\%$                       |
| Gauge factor of individual resistors            | 55 - 70                                        |
| Non-linearity (typical)                         | $\pm 0.25\%$ FSO                               |
| Hysteresis (typical)                            | $\pm 0.1\%$ FSO                                |
| Breakdown voltage between resistors             | min. 15V                                       |
| <b>Environmental</b>                            |                                                |
| Temperature range (operating and storage)       | -55 °C to + 125 °C                             |
| Temperature coefficient of individual resistors | $0.8 \times 10^{-3} / ^\circ\text{C} \pm 25\%$ |
| Thermal zero shift                              | $\pm 0.02\%$ FSO/ °C                           |
| Thermal sensitivity shift                       | $-0.17 \pm 0.05\%$ of reading / °C             |
| Natural resonance frequency (in air)            | $\sim 12\text{kHz}$                            |
| <b>Mechanical</b>                               |                                                |
| Modules of elasticity                           | $1.6 \times 10^5 \text{ N/mm}^2$               |
| Spring constant at full length (typical)        | 2 N/mm                                         |
| Weight approximate                              | 120 mg                                         |
| Total sensor length                             | 10mm                                           |
| Sensor element length                           | 5mm                                            |
| Electrical connection wires (length)            | 30mm                                           |
| Diameter of header:                             | 1.8mm                                          |

---

## II. Determination of AE801 Orthotropic Elastic Stiffness Matrix

The AE801 is manufactured from a monocrystalline silicon wafer (Semiconductor Equipment Materials International (SEMI) specification: SEMI M1.5-89, see Table A2). Monocrystalline silicon is a material with a continuum structure, free of grain boundaries. These material attributes are ideal for force sensing elements because they yield excellent linearity and repeatability to mechanical loading. Nonetheless, monocrystalline silicon exhibits anisotropic stiffness properties: The elastic mechanical response of monocrystalline silicon varies depending upon the direction from which the load is applied. For any mechanical FE analyses to be performed, the elastic stiffness matrix must be correctly orientated. The work of Hopcroft *et al*, “*What is the Young’s Modulus of Silicon?*” [152] was used in conjunction with specifications provided by the AE801 manufacturer Kronex Technologies Corporation for determining the orientation and magnitude of the elastic stiffness matrix in the FE analyses performed in Chapter 4. Crystallographic planes and directions can be described by Miller indices which can also be references to a Cartesian coordinate system which are described by Hopcroft *et al* [152]. With reference to this notation, the AE801 sensors are cut from a standard n-type (100) wafer (see Table A2). The (100) plane of the crystalline silicon is therefore co-aligned with the top surface of the wafer (see Figure A2). The AE801 sensor elements are thin cuboid cantilevers of dimensions  $5\text{mm} \times 0.95\text{mm} \times 0.15\text{mm}$  (see Figure A1). The sensor elements are cut as 2D profiles from the wafer so that the  $0.15\text{mm}$  edge is oriented to be co-aligned with the Z-axis and perpendicular to the (100) plane. In general, silicon wafers have a flat feature which is used to denote the remaining crystal orientation constraint. The wafer that the AE801 is cut from is oriented so that the  $\{110\}$  direction is parallel with this flat (see Figure A3). The sensor elements are manufactured from the silicon wafer so that the  $0.95\text{mm}$  edge is oriented to be parallel with the silicon wafer flat (see Figure A4), therefore the Y-axis was attributed to be parallel with the longest ( $5\text{mm}$ ) length of the sensor element. This coordinate system is presented in Figure 4.9.

**Table A2: Specifications of the silicon wafer from which the AE801 sensor elements are manufactured from. All data provided by Kronex Technologies Corporation (Oakland, USA) [www.kronex.com](http://www.kronex.com).**

|                         |                   |
|-------------------------|-------------------|
| Standard                | SEMI M1.5-89      |
| Crystal Orientation     | (100)             |
| Conductivity type       | N                 |
| Dopant                  | Phosphorus        |
| Diameter                | 100.0 $\pm$ 0.5mm |
| Primary flat length     | 32.5mm            |
| Primary flat location   | {110}             |
| Secondary flat length   | 18mm              |
| Secondary flat location | 180°              |

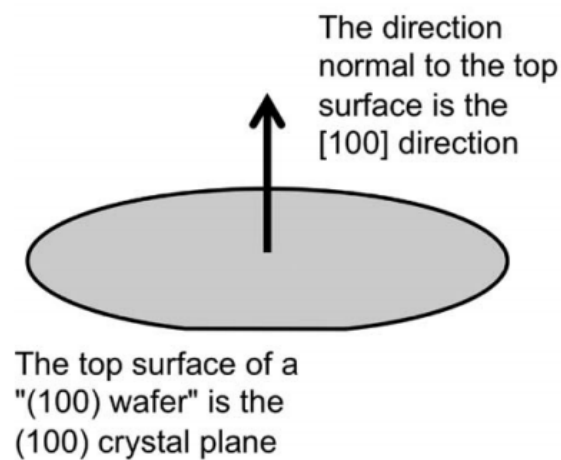


**Figure A2: Illustration from Hopcroft *et al* denoting the [100] direction for a "(100) silicon wafer" equivalent to that which the AE801 is manufactured from. Image from [152] ©2010 IEEE.**

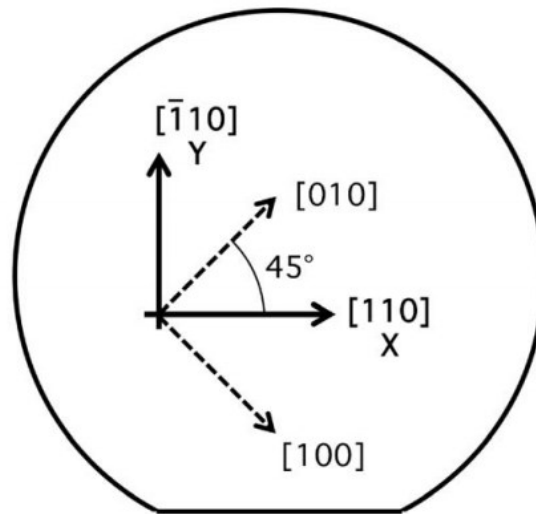


Figure A3: Illustration from Hopcroft *et al* showing the crystal orientation with respect to the flat of a “(100) silicon wafer” in which the X and Y-axes are in the  $[110]$  directions. Image from [152] ©2010 IEEE.



Figure A4: The orientation in which the AE801 sensing elements are profiled with respect to the “(100) silicon wafer” flat. Data and image provided with permission from Kronex Technologies Corporation (Oakland, USA) [www.kronex.com](http://www.kronex.com).

### III. Empirical Characterisation of VCA Force Constant

A characterisation of the VCA (LVCM-010-013-01, Moticont) was performed in order to obtain the force constant ( $k_{VCA}$ ). The force generated by a VCA is directly proportional to the current flowing through its coil and the force constant relates these quantities:

$$F_{VCA} = k_{VCA} \times I_{VCA} \quad A.1$$

The VCA was coupled to a F/T (Nano17, ATI Industrial Automation, Inc., USA) and affixed so as to be in its mid-stroke position. Voltages from 0-3V at 0.3V increments were supplied to the VCA through a power supply with integrated ammeter (E3631A, Agilent Technologies, Santa Clara, USA). A simultaneous measurement of the current and force was made for each voltage increment. The relationship is plotted graphically in Figure A5 and is presented in tabulated form in Table A3. Table A4 provides the derived parameters including the force constant  $k_{VCA}$  and the coil resistance.

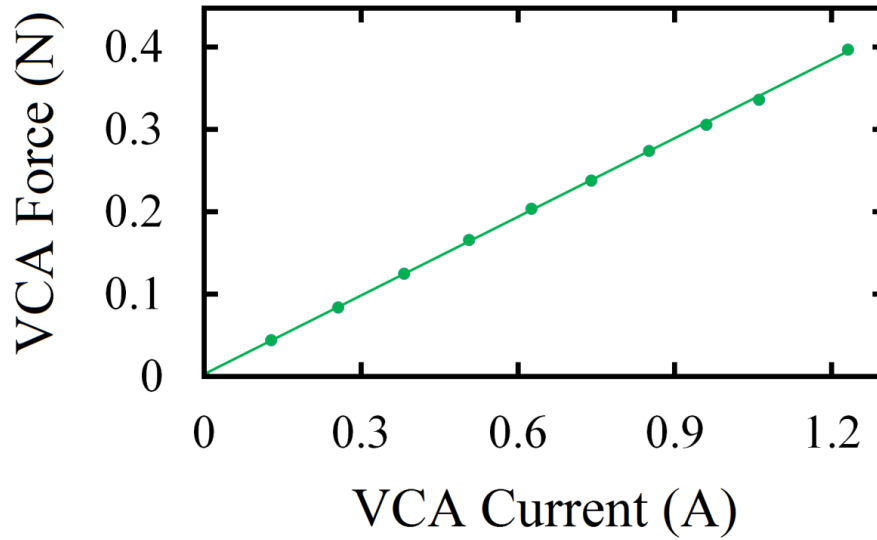


Figure A5: The empirically-characterised linear relationship between  $F_{VCA}$  and  $I_{VCA}$  with a line of best fit using the least squares method, the gradient of which is equivalent to  $k_{VC}$ .

**Table A3: Tabulated voltage, current and force values for the VCA characterisation experiment.**

| Voltage (V) | Current (A) | Force (N) |
|-------------|-------------|-----------|
| 0.00        | 0.00        | 0.00      |
| 0.30        | 0.13        | 0.04      |
| 0.60        | 0.26        | 0.08      |
| 0.90        | 0.38        | 0.13      |
| 1.20        | 0.51        | 0.17      |
| 1.50        | 0.63        | 0.20      |
| 1.80        | 0.74        | 0.24      |
| 2.10        | 0.85        | 0.27      |
| 2.40        | 0.96        | 0.31      |
| 2.70        | 1.06        | 0.34      |
| 3.00        | 1.23        | 0.40      |

**Table A4: Parameters derived from the VCA characterisation and comparison to those quoted in the LVCM-010-013-01 datasheet provided by (Moticont, Van Nuys, USA) [www.moticont.com](http://www.moticont.com).**

| Parameter                                                 | Value |
|-----------------------------------------------------------|-------|
| Measured force constant $K_{VCA}$ (N/A)                   | 0.32  |
| Coefficient of determination $R^2$                        | 1.00  |
| Measured coil resistance ( $\Omega$ )                     | 2.42  |
| Force constant in LVCM-010-013-01 datasheet (N/A)         | 0.29  |
| Coil resistance in LVCM-010-013-01 datasheet ( $\Omega$ ) | 1.90  |

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# References

- [1] Y. S. Kwoh, J. Hou, E. A. Jonckheere, and S. Hayati, "A robot with improved absolute positioning accuracy for CT guided stereotactic brain surgery.," *IEEE Trans. Biomed. Eng.*, vol. 35, no. 2, pp. 153–60, Feb. 1988.
- [2] P. Kazanzides, B. D. Mittelstadt, B. L. Musits, W. L. Bargar, J. F. Zuhars, B. Williamson, P. W. Cain, and E. J. Carbone, "An integrated system for cementless hip replacement," *IEEE Eng. Med. Biol. Mag.*, vol. 14, no. 3, pp. 307–313, 1995.
- [3] S. J. Harris, F. Arambula-Cosio, Q. Mei, R. D. Hibberd, B. L. Davies, J. E. Wickham, M. S. Nathan, and B. Kundu, "The Probot--an active robot for prostate resection.," *Proc. Inst. Mech. Eng. H.*, vol. 211, no. 4, pp. 317–25, Jan. 1997.
- [4] B. Davies, "A review of robotics in surgery," *Proc. Inst. Mech. Eng. Part H J. Eng. Med.*, vol. 214, no. 1, pp. 129–140, Jan. 2000.
- [5] R. D. Howe and Y. Matsuoka, "Robotics for surgery.," *Annu. Rev. Biomed. Eng.*, vol. 1, pp. 211–40, Jan. 1999.
- [6] B. L. Davies, S. J. Harris, W. J. Lin, R. D. Hibberd, R. Middleton, and J. C. Cobb, "Active compliance in robotic surgery--the use of force control as a dynamic constraint.," *Proc. Inst. Mech. Eng. H.*, vol. 211, no. 4, pp. 285–92, Jan. 1997.
- [7] H. Rafii-Tari, C. J. Payne, and G.-Z. Yang, "Current and Emerging Robot-Assisted Endovascular Catheterization Technologies: A Review.," *Ann. Biomed. Eng.*, Nov. 2013.
- [8] G. R. Sutherland, P. B. McBeth, and D. F. Louw, "NeuroArm: an MR compatible robot for microsurgery," *Int. Congr. Ser.*, vol. 1256, pp. 504–508, Jun. 2003.
- [9] G. S. Guthart and J. Salisbury Jr., "The Intuitive telesurgery system: overview and application," in *Robotics and Automation, 2000. Proceedings. ICRA '00. IEEE International Conference on*, 2000, vol. 1, pp. 618–621 vol.1.
- [10] H. Marcus, D. Nandi, A. Darzi, and G.-Z. Yang, "Surgical robotics through a keyhole: from today's translational barriers to tomorrow's 'disappearing' robots.," *IEEE Trans. Biomed. Eng.*, vol. 60, no. 3, pp. 674–81, Mar. 2013.
- [11] B. J. Park and R. M. Flores, "Cost comparison of robotic, video-assisted thoracic surgery and thoracotomy approaches to pulmonary lobectomy.," *Thorac. Surg. Clin.*, vol. 18, no. 3, pp. 297–300, vii, Aug. 2008.
- [12] S. Paul, P. McCulloch, and A. Sedrakyan, "Robotic surgery: revisiting 'no innovation without evaluation'.," *BMJ*, vol. 346, p. f1573, Jan. 2013.

- [13] J. Heemskerk, R. van Dam, W. G. van Gemert, G. L. Beets, J. W. M. Greve, M. J. H. M. Jacobs, and N. D. Bouvy, "First results after introduction of the four-armed da Vinci Surgical System in fully robotic laparoscopic cholecystectomy.," *Dig. Surg.*, vol. 22, no. 6, pp. 426–31, Jan. 2005.
- [14] C. Bergeles and G.-Z. Yang, "From passive tool holders to microsurgeons: safer, smaller, smarter surgical robots.," *IEEE Trans. Biomed. Eng.*, vol. 61, no. 5, pp. 1565–76, May 2014.
- [15] P. Dario, B. Hannaford, and A. Menciassi, "Smart surgical tools and augmenting devices," *IEEE Trans. Robot. Autom.*, vol. 19, no. 5, pp. 782–792, Oct. 2003.
- [16] R. H. Taylor, B. D. Mittelstadt, H. A. Paul, W. Hanson, P. Kazanzides, J. F. Zuhars, B. Williamson, B. L. Musits, E. Glassman, and W. L. Bargar, "An image-directed robotic system for precise orthopaedic surgery," *IEEE Trans. Robot. Autom.*, vol. 10, no. 3, pp. 261–275, Jun. 1994.
- [17] R. Taylor, "A Steady-Hand Robotic System for Microsurgical Augmentation," *Int. J. Rob. Res.*, vol. 18, no. 12, pp. 1201–1210, Dec. 1999.
- [18] A. Uneri, M. A. Balicki, J. Handa, P. Gehlbach, R. H. Taylor, and I. Iordachita, "New Steady-Hand Eye Robot with Micro-Force Sensing for Vitreoretinal Surgery.," *Proc. IEEE. RAS. EMBS. Int. Conf. Biomed. Robot. Biomechatron.*, vol. 2010, no. 26–29, pp. 814–819, Sep. 2010.
- [19] B. M. Wolfe, "Endoscopic Cholecystectomy," *Arch. Surg.*, vol. 126, no. 10, p. 1192, Oct. 1991.
- [20] S. E. Salcudean, S. Ku, and G. Bell, "Performance measurement in scaled teleoperation for microsurgery," in *CVRMed-MRCAS'97 SE - 86*, vol. 1205, J. Troccaz, E. Grimson, and R. Mösges, Eds. Springer Berlin Heidelberg, 1997, pp. 789–798.
- [21] S. Charles, H. Das, T. Ohm, C. Boswell, G. Rodriguez, R. Steele, and D. Istrate, "Dexterity-enhanced telerobotic microsurgery," in *1997 8th International Conference on Advanced Robotics. Proceedings. ICAR'97*, pp. 5–10.
- [22] P. D. Le Roux, H. Das, S. Esquenazi, and P. J. Kelly, "Robot-assisted microsurgery: a feasibility study in the rat," *Neurosurgery*, vol. 48, no. 3, pp. 584–589, 2001.
- [23] A. Morita, S. Sora, M. Mitsuishi, S. Warisawa, K. Suruman, D. Asai, J. Arata, S. Baba, H. Takahashi, R. Mochizuki, and T. Kirino, "Microsurgical robotic system for the deep surgical field: development of a prototype and feasibility studies in animal and cadaveric models," *J. Neurosurg.*, vol. 103, no. 2, pp. 320–327, 2005.
- [24] S. Yang, R. A. MacLachlan, and C. N. Riviere, "Manipulator Design and Operation for a Six-Degree-of-Freedom Handheld Tremor-Canceling Microsurgical Instrument.," *IEEE ASME Trans. Mechatron.*, vol. 20, no. 2, pp. 761–772, Apr. 2015.

- [25] L. F. Hotrathinyo and C. N. Riviere, "Three-dimensional accuracy assessment of eye surgeons," in *2001 Conference Proceedings of the 23rd Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, vol. 4, pp. 3458–3461.
- [26] W. T. Ang, C. N. Riviere, and P. K. Khosla, "An active hand-held instrument for enhanced microsurgical accuracy," in *Proceedings of the Third International Conference on Medical Image Computing and Computer-Assisted Intervention (MICCAI), LNCS 1935, Pittsburgh, USA, 2000*, pp. 878–886.
- [27] W. T. Ang, P. K. Pradeep, and C. N. Riviere, "Active tremor compensation in microsurgery.," in *IEEE Engineering in Medicine and Biology Society.*, 2004, vol. 4, pp. 2738–2741.
- [28] B. C. Becker, C. R. Valdivieso, J. Biswas, L. A. Lobes, and C. N. Riviere, "Active guidance for laser retinal surgery with a handheld instrument.," *Conf. Proc. IEEE Eng. Med. Biol. Soc.*, vol. 2009, pp. 5587–90, Jan. 2009.
- [29] B. C. Becker, S. Voros, L. A. Lobes, J. T. Handa, G. D. Hager, and C. N. Riviere, "Retinal vessel cannulation with an image-guided handheld robot.," *Conf. Proc. IEEE Eng. Med. Biol. Soc.*, vol. 2010, pp. 5420–3, Jan. 2010.
- [30] G. Montes Grande, A. J. Knisely, B. C. Becker, S. Yang, B. E. Hirsch, and C. N. Riviere, "Handheld micromanipulator for robot-assisted stapes footplate surgery.," *Conf. Proc. IEEE Eng. Med. Biol. Soc.*, vol. 2012, pp. 1422–5, Jan. 2012.
- [31] J. Cuevas Tabares, R. A. Maclachlan, C. A. Ettensohn, and C. N. Riviere, "Cell micromanipulation with an active handheld micromanipulator.," *Conf. Proc. IEEE Eng. Med. Biol. Soc.*, vol. 2010, pp. 4363–4366, 2010.
- [32] S. Yang, M. Balicki, R. A. MacLachlan, X. Liu, J. U. Kang, R. H. Taylor, and C. N. Riviere, "Optical coherence tomography scanning with a handheld vitreoretinal micromanipulator.," *Conf. Proc. IEEE Eng. Med. Biol. Soc.*, vol. 2012, pp. 948–51, Jan. 2012.
- [33] M. Patkin, "Ergonomics applied to the practice of microsurgery.," *Aust. N. Z. J. Surg.*, vol. 47, no. 3, pp. 320–329, 1977.
- [34] R. C. Harwell and R. L. Ferguson, "Physiologic tremor and microsurgery.," *Microsurgery*, vol. 4, no. 3, pp. 187–192, 1983.
- [35] W. T. Latt, U.-X. Tan, C. Y. Shee, C. N. Riviere, and W. T. Ang, "Compact Sensing Design of a Handheld Active Tremor Compensation Instrument.," *IEEE Sens. J.*, vol. 9, no. 12, pp. 1864–1871, Dec. 2009.
- [36] W. T. Latt, U.-X. Tan, C. N. Riviere, and W. T. Ang, "Placement of Accelerometers for High Sensing Resolution in Micromanipulation.," *Sens. Actuators. A. Phys.*, vol. 167, no. 2, pp. 304–316, Jun. 2011.

- [37] U.-X. Tan, K. C. Veluvolu, W. T. Latt, C. Y. Shee, C. N. Riviere, and W. T. Ang, "Estimating Displacement of Periodic Motion With Inertial Sensors.," *IEEE Sens. J.*, vol. 8, no. 8, pp. 1385–1388, Aug. 2009.
- [38] R. A. Maclachlan and C. N. Riviere, "High-Speed Microscale Optical Tracking Using Digital Frequency-Domain Multiplexing.," *IEEE Trans. Instrum. Meas.*, vol. 58, no. 6, pp. 1991–2001, Jun. 2009.
- [39] R. A. Maclachlan, B. C. Becker, J. C. Tabarés, G. W. Podnar, L. A. Lobes, and C. N. Riviere, "Micron: an Actively Stabilized Handheld Tool for Microsurgery.," *IEEE Trans. Robot.*, vol. 28, no. 1, pp. 195–212, Feb. 2012.
- [40] B. C. Becker, S. Voros, R. A. Maclachlan, G. D. Hager, and C. N. Riviere, "Active Guidance of a Handheld Micromanipulator using Visual Servoing.," *IEEE Int. Conf. Robot. Autom.*, vol. 2009, pp. 339–344, May 2009.
- [41] W. T. Latt, U.-X. Tan, C. Y. Shee, and W. T. Ang, "A compact hand-held active physiological tremor compensation instrument," in *2009 IEEE/ASME International Conference on Advanced Intelligent Mechatronics*, 2009, pp. 711–716.
- [42] U.-X. Tan, W. T. Latt, C. Y. Shee, and W. T. Ang, "A Low-Cost Flexure-Based Handheld Mechanism for Micromanipulation," *IEEE/ASME Trans. Mechatronics*, vol. 16, no. 4, pp. 773–778, Aug. 2011.
- [43] A. Saxena and R. V. Patel, "An active handheld device for compensation of physiological tremor using an ionic polymer metallic composite actuator," in *2013 IEEE/RSJ International Conference on Intelligent Robots and Systems*, 2013, pp. 4275–4280.
- [44] D. Chang, G. M. Gu, and J. Kim, "Design of a novel tremor suppression device using a linear delta manipulator for micromanipulation," in *2013 IEEE/RSJ International Conference on Intelligent Robots and Systems*, 2013, pp. 413–418.
- [45] S. Yang, R. A. MacLachlan, and C. N. Riviere, "Design and analysis of 6 DOF handheld micromanipulator," in *2012 IEEE International Conference on Robotics and Automation*, 2012, pp. 1946–1951.
- [46] B. Gonenc, E. Feldman, P. Gehlbach, J. Handa, R. H. Taylor, and I. Iordachita, "Towards Robot-Assisted Vitreoretinal Surgery: Force-Sensing Micro-Forceps Integrated with a Handheld Micromanipulator.," *IEEE Int. Conf. Robot. Autom.*, vol. 2014, pp. 1399–1404, Jan. 2014.
- [47] B. Gonenc, P. Gehlbach, J. Handa, R. H. Taylor, and I. Iordachita, "Motorized Force-Sensing Micro-Forceps with Tremor Cancelling and Controlled Micro-Vibrations for Easier Membrane Peeling.," *Proc. IEEE. RAS. EMBS. Int. Conf. Biomed. Robot. Biomechatron.*, vol. 2014, pp. 244–251, Aug. 2014.
- [48] B. C. Becker, R. A. MacLachlan, L. A. Lobes, G. D. Hager, and C. N. Riviere, "Vision-Based Control of a Handheld Surgical Micromanipulator With Virtual Fixtures," *IEEE Trans. Robot.*, vol. 29, no. 3, pp. 674–683, Jun. 2013.

- [49] B. C. Becker, R. A. Maclachlan, G. D. Hager, and C. N. Riviere, "Handheld Micromanipulation with Vision-Based Virtual Fixtures.," *IEEE Int. Conf. Robot. Autom.*, vol. 2011, pp. 4127–4132, Aug. 2011.
- [50] C. Song, D. Y. Park, P. L. Gehlbach, S. J. Park, and J. U. Kang, "Fiber-optic OCT sensor guided 'SMART' micro-forceps for microsurgery.," *Biomed. Opt. Express*, vol. 4, no. 7, pp. 1045–50, Jul. 2013.
- [51] G. Brisson, T. Kanade, A. M. DiGioia III, and B. Jaramaz, "Precision Freehand Sculpting of Bone," in *7th International Conference Medical Image Computing Computer-Assisted Intervention*, 2004, vol. 11, pp. 105–112.
- [52] A. Follmann, A. Korff, T. Fuertjes, S. C. Kunze, K. Schmieder, and K. Radermacher, "A novel concept for smart trepanation.," *J. Craniofac. Surg.*, vol. 23, no. 1, pp. 309–14, Jan. 2012.
- [53] G. Kane, G. Eggers, R. Boesecke, J. Raczowsky, H. Wörn, R. Marmulla, and J. Mühling, "System design of a hand-held mobile robot for craniotomy.," *Med. Image Comput. Comput. Assist. Interv.*, vol. 12, no. Pt 1, pp. 402–9, Jan. 2009.
- [54] A. Hackethal, M. Koppan, K. Eskef, and H.-R. Tinneberg, "Handheld articulating laparoscopic instruments driven by robotic technology. First clinical experience in gynecological surgery," *Gynecol. Surg.*, vol. 9, no. 2, pp. 203–206, Oct. 2011.
- [55] N. Matsuhira, M. Jinno, T. Miyagawa, T. Sunaoshi, T. Hato, Y. Morikawa, T. Furukawa, S. Ozawa, M. Kitajima, and K. Nakazawa, "Development of a functional model for a master–slave combined manipulator for laparoscopic surgery," *Adv. Robot.*, vol. 17, no. 6, pp. 523–539, Jan. 2003.
- [56] A. H. Zahraee, J. K. Paik, J. Szewczyk, and G. Morel, "Toward the Development of a Hand-Held Surgical Robot for Laparoscopy," *IEEE/ASME Trans. Mechatronics*, Dec. 2010.
- [57] A. Hassan-Zahraee, B. Herman, and J. Szewczyk, "Mechatronic design of a hand-held instrument with active trocar for laparoscopy," in *2011 IEEE International Conference on Robotics and Automation*, 2011, pp. 1890–1895.
- [58] F. Focacci, M. Piccigallo, O. Tonet, G. Megali, A. Pietrabissa, and P. Dario, "Lightweight Hand-held Robot for Laparoscopic Surgery," in *Proceedings 2007 IEEE International Conference on Robotics and Automation*, 2007, pp. 599–604.
- [59] M. Piccigallo, F. Focacci, O. Tonet, G. Megali, C. Quaglia, and P. Dario, "Hand-held robotic instrument for dextrous laparoscopic interventions.," *Int. J. Med. Robot.*, vol. 4, no. 4, pp. 331–8, Dec. 2008.
- [60] J. Shang, D. P. Noonan, C. Payne, J. Clark, M. H. Sodergren, A. Darzi, and G.-Z. Yang, "An articulated universal joint based flexible access robot for minimally invasive surgery," in *2011 IEEE International Conference on Robotics and Automation*, 2011, pp. 1147–1152.

- [61] H. Yamashita, N. Hata, M. Hashizume, and T. Dohi, "Handheld Laparoscopic Forceps Manipulator Using Multi-slider Linkage Mechanisms," in *Medical Image Computing and Computer-Assisted Intervention – MICCAI 2004 SE - 16*, vol. 3217, C. Barillot, D. Haynor, and P. Hellier, Eds. Springer Berlin Heidelberg, 2004, pp. 121–128.
- [62] S. Okazawa, R. Ebrahimi, J. Chuang, S. E. Salcudean, and R. Rohling, "Hand-Held Steerable Needle Device," *IEEE/ASME Trans. Mechatronics*, vol. 10, no. 3, pp. 285–296, Jun. 2005.
- [63] P. Dario, M. C. Carrozza, M. Marcacci, S. D’Attanasio, B. Magnani, O. Tonet, and G. Megali, "A novel mechatronic tool for computer-assisted arthroscopy," *IEEE Trans. Inf. Technol. Biomed.*, vol. 4, no. 1, pp. 15–29, Mar. 2000.
- [64] S. D’Attanasio, O. Tonet, G. Megali, M. C. Carrozza, and P. Dario, "A semi-automatic handheld mechatronic endoscope with collision-avoidance capabilities," in *Proceedings 2000 ICRA. Millennium Conference. IEEE International Conference on Robotics and Automation. Symposia Proceedings (Cat. No.00CH37065)*, vol. 2, pp. 1586–1591.
- [65] S. G. Yuen, M. C. Yip, N. V Vasilyev, D. P. Perrin, P. J. Del Nido, and R. D. Howe, "Robotic Force Stabilization for Beating Heart Intracardiac Surgery.," *Med. Image Comput. Comput. Assist. Interv.*, vol. 5761, no. 2009, pp. 26–33, Oct. 2009.
- [66] S. G. Yuen, D. P. Perrin, N. V. Vasilyev, P. J. del Nido, and R. D. Howe, "Force Tracking With Feed-Forward Motion Estimation for Beating Heart Surgery," *IEEE Trans. Robot.*, vol. 26, no. 5, pp. 888–896, Oct. 2010.
- [67] O. Bebek and M. C. Cavusoglu, "Intelligent Control Algorithms for Robotic-Assisted Beating Heart Surgery," *IEEE Trans. Robot.*, vol. 23, no. 3, pp. 468–480, Jun. 2007.
- [68] J. M. Florez, D. Bellot, and G. Morel, "LWPR-model based predictive force control for serial comanipulation in beating heart surgery," in *2011 IEEE/ASME International Conference on Advanced Intelligent Mechatronics (AIM)*, 2011, pp. 320–326.
- [69] J. M. Florez, J. Szewczyk, and G. Morel, "An impedance control strategy for a hand-held instrument to compensate for physiological motion," in *2012 IEEE International Conference on Robotics and Automation*, 2012, pp. 1952–1957.
- [70] S. G. Yuen, K.-A. Dubec, and R. D. Howe, "Haptic noise cancellation: Restoring force perception in robotically-assisted beating heart surgery," in *2010 IEEE Haptics Symposium*, 2010, pp. 387–392.
- [71] P. Brett, X. Du, M. Zoka Assadi, D. Proops, A. Reid, and C. Coulson, "Mechatronic hand-held surgical robots," in *Mechatronics and Machine Vision in Practice (M2VIP), 2012 19th International Conference*, 2012, pp. 447–449.
- [72] P. N. Brett, D. A. Baker, L. Reyes, and J. Blanshard, "An automatic technique for micro- drilling a stapedotomy in the flexible stapes footplate," *Arch. Proc. Inst.*

- Mech. Eng. Part H J. Eng. Med. 1989-1996 (vols 203-210)*, vol. 209, no. 48, pp. 255–262, Jun. 1995.
- [73] P. N. Brett, R. P. Taylor, D. Proops, C. Coulson, A. Reid, and M. V Griffiths, “A surgical robot for cochleostomy,” *Conf. Proc. ... Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. IEEE Eng. Med. Biol. Soc. Annu. Conf.*, vol. 2007, pp. 1229–32, Jan. 2007.
  - [74] R. Taylor, X. Du, D. Proops, A. Reid, C. Coulson, and P. N. Brett, “A sensory-guided surgical micro-drill,” *Proc. Inst. Mech. Eng. Part C J. Mech. Eng. Sci.*, vol. 224, no. 7, pp. 1531–1537, Jan. 2010.
  - [75] W. T. Latt, R. C. Newton, M. Visentini-Scarzanella, C. J. Payne, D. P. Noonan, J. Shang, and G.-Z. Yang, “A Hand-held Instrument to Maintain Steady Tissue Contact during Probe-Based Confocal Laser Endomicroscopy,” *IEEE Trans. Biomed. Eng.*, vol. 58, no. 9, pp. 2694–2703, 2011.
  - [76] W. Tun Latt, T. P. Chang, A. Di Marco, P. Pratt, K.-W. Kwok, J. Clark, and G.-Z. Yang, “A hand-held instrument for in vivo probe-based confocal laser endomicroscopy during Minimally Invasive Surgery,” in *2012 IEEE/RSJ International Conference on Intelligent Robots and Systems*, 2012, pp. 1982–1987.
  - [77] R. C. Newton, D. Noonan, C. Payne, J. Andreyev, A. Di Marco, M. V. Scarzanella, A. Darzi, and G.-Z. Yang, “Probe tip contact force and bowel distension affect crypt morphology during confocal endomicroscopy,” *Gut*, vol. 60, no. Suppl 1, pp. A12–A13, Mar. 2011.
  - [78] M. W. Gilbertson and B. W. Anthony, “Ergonomic control strategies for a handheld force-controlled ultrasound probe,” in *2012 IEEE/RSJ International Conference on Intelligent Robots and Systems*, 2012, pp. 1284–1291.
  - [79] T. S. Wells, R. A. MacLachlan, and C. N. Riviere, “Toward hybrid position/force control for an active handheld micromanipulator,” in *2014 IEEE International Conference on Robotics and Automation (ICRA)*, 2014, pp. 772–777.
  - [80] J. J. Gibson, *The senses considered as perceptual systems*. Houghton Mifflin, 1966.
  - [81] S. J. Lederman and R. L. Klatzky, “Haptic perception: a tutorial,” *Atten. Percept. Psychophys.*, vol. 71, no. 7, pp. 1439–59, Oct. 2009.
  - [82] J. M. Wolfe, *Sensation & perception*. Sinauer Associates, 2006.
  - [83] J. J. Gibson, “Observations on active touch,” *Psychol. Rev.*, vol. 69, no. 6, pp. 477–491, 1962.
  - [84] S. Weinstein, “Intensive and extensive aspects of tactile sensitivity as a function of body part, sex and laterality,” in *the First Int’l symp. on the Skin Senses, 1968*, 1968.
  - [85] J. C. Stevens and M. Q. Patterson, “Dimensions of spatial acuity in the touch sense: changes over the life span,” *Somatosens. Mot. Res.*, vol. 12, no. 1, pp. 29–47, Jan. 1995.

- [86] G. A. Gescheider, "Effects of signal probability on vibrotactile signal recognition.," *Percept. Mot. Skills*, vol. 38, no. 1, pp. 15–23, Feb. 1974.
- [87] E. H. Weber, H. E. Ross, and D. J. Murray, *E.H. Weber on the Tactile Senses*. Erlbaum (UK) Taylor & Francis, 1996.
- [88] E. P. Westebring-van der Putten, R. H. M. Goossens, J. J. Jakimowicz, and J. Dankelman, "Haptics in minimally invasive surgery--a review.," *Minim. Invasive Ther. Allied Technol.*, vol. 17, no. 1, pp. 3–16, Jan. 2008.
- [89] O. S. Bholat, R. S. Haluck, W. B. Murray, P. J. Gorman, and T. M. Krummel, "Tactile feedback is present during minimally invasive surgery.," *J. Am. Coll. Surg.*, vol. 189, no. 4, pp. 349–55, Oct. 1999.
- [90] P. Gupta, P. Jensen, and E. Juan Jr., "Surgical Forces and Tactile Perception During Retinal Microsurgery," in *Medical Image Computing and Computer-Assisted Intervention – MICCAI'99 SE - 132*, vol. 1679, C. Taylor and A. Colchester, Eds. Springer Berlin Heidelberg, 1999, pp. 1218–1225.
- [91] K. Harada, Y. Minakawa, Y. Baek, Y. Kozuka, S. Sora, A. Morita, N. Sugita, and M. Mitsuishi, "Microsurgical skill assessment: toward skill-based surgical robotic control.," *Conf. Proc. ... Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. IEEE Eng. Med. Biol. Soc. Annu. Conf.*, vol. 2011, pp. 6700–3, Jan. 2011.
- [92] F. W. Mohr, V. Falk, A. Diegeler, T. Walther, J. F. Gummert, J. Bucerius, S. Jacobs, and R. Autschbach, "Computer-enhanced 'robotic' cardiac surgery: experience in 148 patients.," *J. Thorac. Cardiovasc. Surg.*, vol. 121, no. 5, pp. 842–53, May 2001.
- [93] R. L. Klatzky, S. J. Lederman, and C. Reed, "There's more to touch than meets the eye: The salience of object attributes for haptics with and without vision.," *J. Exp. Psychol. Gen.*, vol. 116, no. 4, p. 356, 1987.
- [94] M. O. Ernst and M. S. Banks, "Humans integrate visual and haptic information in a statistically optimal fashion.," *Nature*, vol. 415, no. 6870, pp. 429–33, Jan. 2002.
- [95] C. G. L. Cao, J. L. Webster, J. O. Perreault, S. Schwaitzberg, and G. Rogers, "Visually perceived force feedback in simulated robotic surgery," in *Proceedings of the Human Factors and Ergonomics Society Annual Meeting*, 2003, vol. 47, no. 12, pp. 1466–1470.
- [96] M. E. Hagen, J. J. Meehan, I. Inan, and P. Morel, "Visual clues act as a substitute for haptic feedback in robotic surgery.," *Surg. Endosc.*, vol. 22, no. 6, pp. 1505–8, Jun. 2008.
- [97] G. Tholey, J. P. Desai, and A. E. Castellanos, "Force feedback plays a significant role in minimally invasive surgery: results and analysis.," *Ann. Surg.*, vol. 241, no. 1, pp. 102–9, Jan. 2005.
- [98] C. R. Wagner, N. Stylopoulos, and R. D. Howe, "The role of force feedback in surgery: analysis of blunt dissection," in *Proceedings 10th Symposium on Haptic*

*Interfaces for Virtual Environment and Teleoperator Systems. HAPTICS 2002*, pp. 68–74.

- [99] C. R. Wagner, D. P. Perrin, R. D. Howe, N. Vasilyev, and P. J. del Nido, “Force Feedback in a Three-Dimensional Ultrasound-Guided Surgical Task,” in *2006 14th Symposium on Haptic Interfaces for Virtual Environment and Teleoperator Systems*, pp. 43–48.
- [100] L. A. Jones, “Visual and haptic feedback in the control of force.,” *Exp. brain Res.*, vol. 130, no. 2, pp. 269–72, Jan. 2000.
- [101] M. Kitagawa, D. Dokko, A. M. Okamura, and D. D. Yuh, “Effect of sensory substitution on suture-manipulation forces for robotic surgical systems.,” *J. Thorac. Cardiovasc. Surg.*, vol. 129, no. 1, pp. 151–8, Jan. 2005.
- [102] C. E. Reiley, T. Akinbiyi, D. Burschka, D. C. Chang, A. M. Okamura, and D. D. Yuh, “Effects of visual force feedback on robot-assisted surgical task performance.,” *J. Thorac. Cardiovasc. Surg.*, vol. 135, no. 1, pp. 196–202, Jan. 2008.
- [103] K. J. Kuchenbecker, J. Gewirtz, W. McMahan, D. Standish, P. Martin, J. Bohren, P. J. Mendoza, and D. I. Lee, “VerroTouch: high-frequency acceleration feedback for telerobotic surgery,” in *Haptics: Generating and perceiving tangible sensations*, Springer, 2010, pp. 189–196.
- [104] K. Bark, W. McMahan, A. Remington, J. Gewirtz, A. Wedmid, D. I. Lee, and K. J. Kuchenbecker, “In vivo validation of a system for haptic feedback of tool vibrations in robotic surgery.,” *Surg. Endosc.*, vol. 27, no. 2, pp. 656–64, Feb. 2013.
- [105] M. Jakopc, S. J. Harris, F. R. y Baena, P. Gomes, J. Cobb, and B. L. Davies, “Preliminary results of an early clinical experience with the acrobot™ system for total knee replacement surgery,” in *Medical Image Computing and Computer-Assisted Intervention—MICCAI 2002*, Springer, 2002, pp. 256–263.
- [106] L. B. Rosenberg, “Virtual fixtures: Perceptual tools for telerobotic manipulation,” in *Proceedings of IEEE Virtual Reality Annual International Symposium*, pp. 76–82.
- [107] M. Jakopc, F. Rodriguez Baena, S. J. Harris, P. Gomes, J. Cobb, and B. L. Davies, “The hands-on orthopaedic robot ‘acrobot’: early clinical trials of total knee replacement surgery,” *IEEE Trans. Robot. Autom.*, vol. 19, no. 5, pp. 902–911, Oct. 2003.
- [108] M. Li, M. Ishii, and R. H. Taylor, “Spatial Motion Constraints Using Virtual Fixtures Generated by Anatomy,” *IEEE Trans. Robot.*, vol. 23, no. 1, pp. 4–19, Feb. 2007.
- [109] A. M. Okamura, “Haptic feedback in robot-assisted minimally invasive surgery,” *Curr. Opin. Urol.*, vol. 19, no. 1, pp. 102–107, 2009.
- [110] T. H. Massie and J. K. Salisbury, “The phantom haptic interface: A device for probing virtual objects,” in *Proceedings of the ASME winter annual meeting*,

*symposium on haptic interfaces for virtual environment and teleoperator systems.*, 1994, pp. 295–300.

- [111] V. Hayward, O. R. Astley, M. Cruz-Hernandez, D. Grant, and G. Robles-De-La-Torre, “Haptic interfaces and devices,” *Sens. Rev.*, vol. 24, no. 1, pp. 16–29, 2004.
- [112] V. Hayward and K. Maclean, “Do it yourself haptics: part I,” *IEEE Robot. Autom. Mag.*, vol. 14, no. 4, pp. 88–104, Dec. 2007.
- [113] J. Luk, J. Pasquero, S. Little, K. MacLean, V. Levesque, and V. Hayward, “A role for haptics in mobile interaction,” in *Proceedings of the SIGCHI conference on Human Factors in computing systems - CHI '06*, 2006, p. 171.
- [114] A. Arasan, C. Basdogan, and T. M. Sezgin, “Haptic stylus with inertial and vibrotactile feedback,” in *2013 World Haptics Conference (WHC)*, 2013, pp. 425–430.
- [115] C. Swindells, A. Unden, and T. Sang, “TorqueBAR: an ungrounded haptic feedback device,” in *Proceedings of the 5th international conference on Multimodal interfaces - ICMI '03*, 2003, pp. 52–59.
- [116] Y. Tanaka, S. Masataka, K. Yuka, Y. Fukui, J. Yamashita, and N. Nakamura, “Mobile torque display and haptic characteristics of human palm,” in *in Proc. ICAT*, 2001, pp. 115–120.
- [117] H. Yano, M. Yoshie, and H. Iwata, “Development of a non-grounded haptic interface using the gyro effect,” in *11th Symposium on Haptic Interfaces for Virtual Environment and Teleoperator Systems, 2003. HAPTICS 2003. Proceedings.*, pp. 32–39.
- [118] K. N. Winfree, J. Gewirtz, T. Mather, J. Fiene, and K. J. Kuchenbecker, “A high fidelity ungrounded torque feedback device: The iTorqU 2.0,” in *World Haptics 2009 - Third Joint EuroHaptics conference and Symposium on Haptic Interfaces for Virtual Environment and Teleoperator Systems*, 2009, pp. 261–266.
- [119] K. N. Winfree, J. M. Romano, J. Gewirtz, and K. J. Kuchenbecker, “Control of a high fidelity ungrounded torque feedback device: The iTorqU 2.1,” in *2010 IEEE International Conference on Robotics and Automation*, 2010, pp. 1347–1352.
- [120] A. Badshah, S. Gupta, D. Morris, S. Patel, and D. Tan, “GyroTab: a handheld device that provides reactive torque feedback,” in *Proceedings of the 2012 ACM annual conference on Human Factors in Computing Systems - CHI '12*, 2012, pp. 3153–3156.
- [121] H. Gurocak, S. Jayaram, B. Parrish, and U. Jayaram, “Weight Sensation in Virtual Environments Using a Haptic Device With Air Jets,” *J. Comput. Inf. Sci. Eng.*, vol. 3, no. 2, p. 130, 2003.
- [122] M. Turner, D. Gomez, M. Tremblay, and M. R. Cutkosky, “Preliminary tests of an arm-grounded haptic feedback device in telemanipulation,” in *Proc. of the ASME Dynamic Systems and Control Division.*, 1998, pp. 145–149.

- [123] M. Aiple and A. Schiele, "Pushing the limits of the CyberGrasp™ for haptic rendering," in *2013 IEEE International Conference on Robotics and Automation*, 2013, pp. 3541–3546.
- [124] S. Kamuro, K. Minamizawa, N. Kawakami, and S. Tachi, "Ungrounded kinesthetic pen for haptic interaction with virtual environments," in *RO-MAN 2009 - The 18th IEEE International Symposium on Robot and Human Interactive Communication*, 2009, pp. 436–441.
- [125] S. Kamuro, K. Minamizawa, and S. Tachi, "An ungrounded pen-shaped kinesthetic display: Device construction and applications," in *2011 IEEE World Haptics Conference*, 2011, pp. 557–562.
- [126] C. Richard and M. R. Cutkosky, "Contact force perception with an ungrounded haptic interface," in *In Proceedings of the ASME Dynamic Systems and Control Division*, 1997, pp. 181–187.
- [127] A. Bicchi, G. Canepa, D. De Rossi, P. Iaconi, and E. P. Scillingo, "A sensor-based minimally invasive surgery tool for detecting tissutal elastic properties(003) 5323219," in *Proceedings of IEEE International Conference on Robotics and Automation*, vol. 1, pp. 884–888.
- [128] T. Hu, G. Tholey, J. P. Desai, and A. E. Castellanos, "Evaluation of a laparoscopic grasper with force feedback.," *Surg. Endosc.*, vol. 18, no. 5, pp. 863–7, May 2004.
- [129] G. B. Hanna, T. Drew, G. Arnold, M. Fakhry, and A. Cuschieri, "Development of force measurement system for clinical use in minimal access surgery.," *Surg. Endosc.*, vol. 22, no. 2, pp. 467–71, Feb. 2008.
- [130] M. Fakhry, F. Bello, and G. B. Hanna, "A real-time compliance mapping system using standard endoscopic surgical forceps.," *IEEE Trans. Biomed. Eng.*, vol. 56, no. 4, pp. 1245–53, Apr. 2009.
- [131] M. V Ottermo, M. Ovstedal, T. Langø, O. Stavdahl, Y. Yavuz, T. A. Johansen, and R. Mårvik, "The role of tactile feedback in laparoscopic surgery.," *Surg. Laparosc. Endosc. Percutan. Tech.*, vol. 16, no. 6, pp. 390–400, Dec. 2006.
- [132] M. V. Ottermo, O. Stavdahl, and T. A. Johansen, "Palpation instrument for augmented minimally invasive surgery," in *2004 IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS) (IEEE Cat. No.04CH37566)*, vol. 4, pp. 3960–3964.
- [133] R. D. Howe, W. J. Peine, D. A. Kantarinis, and J. S. Son, "Remote palpation technology," *IEEE Eng. Med. Biol. Mag.*, vol. 14, no. 3, pp. 318–323, 1995.
- [134] H.-Y. Yao, V. Hayward, and R. E. Ellis, "A tactile enhancement instrument for minimally invasive surgery.," *Comput. Aided Surg.*, vol. 10, no. 4, pp. 233–9, Jul. 2005.

- [135] P. J. Berkelman, L. L. Whitcomb, R. H. Taylor, and P. Jensen, "A miniature instrument tip force sensor for robot/human cooperative microsurgical manipulation with enhanced force feedback," in *Medical Image Computing and Computer-Assisted Intervention—MICCAI 2000*, 2000, pp. 897–906.
- [136] P. J. Berkelman, L. L. Whitcomb, R. H. Taylor, and P. Jensen, "A miniature microsurgical instrument tip force sensor for enhanced force feedback during robot-assisted manipulation," *IEEE Trans. Robot. Autom.*, vol. 19, no. 5, pp. 917–922, Oct. 2003.
- [137] A. S. Jagtap and C. N. Riviere, "Applied force during vitreoretinal microsurgery with handheld instruments.," *Conf. Proc. ... Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. IEEE Eng. Med. Biol. Soc. Annu. Conf.*, vol. 4, pp. 2771–3, Jan. 2004.
- [138] I. Iordachita, Z. Sun, M. Balicki, J. U. Kang, S. J. Phee, J. Handa, P. Gehlbach, and R. Taylor, "A sub-millimetric, 0.25 mN resolution fully integrated fiber-optic force-sensing tool for retinal microsurgery.," *Int. J. Comput. Assist. Radiol. Surg.*, vol. 4, no. 4, pp. 383–90, Jun. 2009.
- [139] M. Balicki, A. Uneri, I. Iordachita, J. Handa, P. Gehlbach, and R. Taylor, "Micro-force sensing in robot assisted membrane peeling for vitreoretinal surgery.," *Med. Image Comput. Comput. Assist. Interv.*, vol. 13, no. Pt 3, pp. 303–10, Jan. 2010.
- [140] B. Gonenc, M. A. Balicki, J. Handa, P. Gehlbach, C. N. Riviere, R. H. Taylor, and I. Iordachita, "Evaluation of a Micro-Force Sensing Handheld Robot for Vitreoretinal Surgery.," *Rep. U. S.*, vol. 2012, pp. 4125–4130, Dec. 2012.
- [141] I. Kuru, B. Gonenc, M. Balicki, J. Handa, P. Gehlbach, R. H. Taylor, and I. Iordachita, "Force sensing micro-forceps for robot assisted retinal surgery.," *Conf. Proc. ... Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. IEEE Eng. Med. Biol. Soc. Annu. Conf.*, vol. 2012, pp. 1401–4, Jan. 2012.
- [142] X. He, M. A. Balicki, J. U. Kang, P. L. Gehlbach, J. T. Handa, R. H. Taylor, and I. I. Iordachita, "Force sensing micro-forceps with integrated fiber Bragg grating for vitreoretinal surgery," 2012, p. 82180W.
- [143] X. He, J. Handa, P. Gehlbach, R. Taylor, and I. Iordachita, "A submillimetric 3-DOF force sensing instrument with integrated fiber Bragg grating for retinal microsurgery.," *IEEE Trans. Biomed. Eng.*, vol. 61, no. 2, pp. 522–34, Feb. 2014.
- [144] G. Stetten, B. Wu, R. Klatzky, J. Galeotti, M. Siegel, R. Lee, F. Mah, A. Eller, J. Schuman, and R. Hollis, "Hand-Held Force Magnifier for Surgical Instruments," in *Information Processing in Computer-Assisted Interventions SE - 9*, vol. 6689, R. Taylor and G.-Z. Yang, Eds. Springer Berlin Heidelberg, 2011, pp. 90–100.
- [145] R. Lee, B. Wu, R. Klatzky, V. Shivaprabhu, J. Galeotti, S. Horvath, M. Siegel, J. Schuman, R. Hollis, and G. Stetten, "Hand-Held Force Magnifier for Surgical Instruments: Evolution toward a Clinical Device," in *Augmented Environments for Computer-Assisted Interventions SE - 9*, vol. 7815, C. Linte, E. S. Chen, M.-O.

- Berger, J. Moore, and D. Holmes III, Eds. Springer Berlin Heidelberg, 2013, pp. 77–89.
- [146] Y. Koseki, D. De Lorenzo, K. Chinzei, and A. M. Okamura, “Coaxial needle insertion assistant for epidural puncture,” in *2011 IEEE/RSJ International Conference on Intelligent Robots and Systems*, 2011, pp. 2584–2589.
  - [147] Y. Kurita, F. Sugihara, J. Ueda, and T. Ogasawara, “Piezoelectric Tweezer-Type End Effector With Force- and Displacement-Sensing Capability,” *IEEE/ASME Trans. Mechatronics*, vol. 17, no. 6, pp. 1039–1048, Dec. 2012.
  - [148] X. He, J. Handa, P. Gehlbach, R. Taylor, and I. Iordachita, “A Sub-Millimetric 3-DOF Force Sensing Instrument with Integrated Fiber Bragg Grating for Retinal Microsurgery,” *IEEE Trans. Biomed. Eng.*, Sep. 2013.
  - [149] K. Kim, X. Liu, Y. Zhang, and Y. Sun, “Nanonewton force-controlled manipulation of biological cells using a monolithic MEMS microgripper with two-axis force feedback,” *J. Micromechanics Microengineering*, vol. 18, no. 5, p. 055013, May 2008.
  - [150] M. C. Carrozza, A. Eisinger, A. Menciassi, D. Campolo, S. Micera, and P. Dario, “Towards a force-controlled microgripper for assembling biomedical microdevices,” *J. Micromechanics Microengineering*, vol. 10, no. 2, pp. 271–276, Jun. 2000.
  - [151] A. Menciassi, A. Eisinger, M. Mazzoni, and P. Dario, “A sensorized  $\mu$ electro discharge machined superelastic alloy microgripper for micromanipulation: simulation and characterization,” in *IEEE/RSJ International Conference on Intelligent Robots and System*, vol. 2, pp. 1591–1595.
  - [152] M. A. Hopcroft, W. D. Nix, and T. W. Kenny, “What is the Young’s Modulus of Silicon?,” *J. Microelectromechanical Syst.*, vol. 19, no. 2, pp. 229–238, Apr. 2010.
  - [153] S. H. Greenblatt, T. F. Dagi, and M. H. Epstein, *A history of neurosurgery : in its scientific and professional contexts*. Park Ridge, Ill.: American Association of Neurological Surgeons, 1997, p. xiv, 625 p.
  - [154] N. Ahmed, K. S. Devitt, I. Keshet, J. Spicer, K. Imrie, L. Feldman, J. Cools-Lartigue, A. Kayssi, N. Lipsman, M. Elmi, A. V Kulkarni, C. Parshuram, T. Mainprize, R. J. Warren, P. Fata, M. S. Gorman, S. Feinberg, and J. Rutka, “A systematic review of the effects of resident duty hour restrictions in surgery: impact on resident wellness, training, and patient outcomes,” *Ann. Surg.*, vol. 259, no. 6, pp. 1041–1053, 2014.
  - [155] L. Thorne, S. Burn, S. Shaw, B. Arvin, and R. Bradford, “Neurosurgical trainees operative experience before and after introduction of the New Deal for junior doctors,” *Br. J. Neurosurg.*, vol. 20, no. 1, pp. 31–35, 2006.
  - [156] H. Marcus, V. Vakharia, M. A. Kirkman, M. Murphy, and D. Nandi, “Practice makes perfect? The role of simulation-based deliberate practice and script-based mental rehearsal in the acquisition and maintenance of operative neurosurgical skills,” *Neurosurgery*, vol. 72 Suppl 1, pp. 124–130, 2013.

- [157] H. J. Marcus, A. Hughes-Hallett, T. P. Cundy, D. Nandi, G. Z. Yang, and A. Darzi, "Not everything that counts can be easily counted," *BMJ*, vol. 346, p. f2461, 2013.
- [158] H. Hwang, J. Lim, C. Kinnaird, A. G. Nagy, O. N. Panton, A. J. Hodgson, and K. A. Qayumi, "Correlating motor performance with surgical error in laparoscopic cholecystectomy," *Surg. Endosc.*, vol. 20, no. 4, pp. 651–655, 2006.
- [159] H. J. Marcus, K. Zareinia, L. S. Gan, F. W. Yang, S. Lama, G. Z. Yang, and G. R. Sutherland, "Forces exerted during microneurosurgery: a cadaver study," *Int. J. Med. Robot.*, 2014.
- [160] L. A. Jones, "Perception of force and weight: theory and research.," *Psychol. Bull.*, vol. 100, no. 1, p. 29, 1986.
- [161] S. J. Lederman, R. D. Howe, R. L. Klatzky, and C. Hamilton, "Force variability during surface contact with bare finger or rigid probe," in *12th International Symposium on Haptic Interfaces for Virtual Environment and Teleoperator Systems, 2004. HAPTICS '04. Proceedings.*, 2004, pp. 154–160.
- [162] M. G. Yasargil, *Microneurosurgery*. Georg Thieme, 1996.
- [163] B. Safwat, E. L. M. Su, R. Gassert, C. L. Teo, and E. Burdet, "The role of posture, magnification, and grip force on microscopic accuracy.," *Ann. Biomed. Eng.*, vol. 37, no. 5, pp. 997–1006, May 2009.
- [164] L. A. Jones, "Perceptual constancy and the perceived magnitude of muscle forces.," *Exp. brain Res.*, vol. 151, no. 2, pp. 197–203, Jul. 2003.
- [165] W.-H. Park and C. T. Leonard, "The effect of intervening forces on finger force perception.," *Neurosci. Lett.*, vol. 438, no. 3, pp. 286–9, Jun. 2008.
- [166] J. A. Martin, G. Regehr, R. Reznick, H. MacRae, J. Murnaghan, C. Hutchison, and M. Brown, "Objective structured assessment of technical skill (OSATS) for surgical residents.," *Br. J. Surg.*, vol. 84, no. 2, pp. 273–8, Feb. 1997.
- [167] H. C. Lin, K. Mills, P. Kazanzides, G. D. Hager, P. Marayong, A. M. Okamura, and R. Karam, "Portability and applicability of virtual fixtures across medical and manufacturing tasks," in *Proceedings 2006 IEEE International Conference on Robotics and Automation, 2006. ICRA 2006.*, pp. 225–230.
- [168] N. Cutler, M. Balicki, M. Finkelstein, J. Wang, P. Gehlbach, J. McGready, I. Iordachita, R. Taylor, and J. T. Handa, "Auditory force feedback substitution improves surgical precision during simulated ophthalmic surgery.," *Invest. Ophthalmol. Vis. Sci.*, vol. 54, no. 2, pp. 1316–24, Feb. 2013.
- [169] R. Wood and C. Walsh, "Smaller, softer, safer, smarter robots.," *Sci. Transl. Med.*, vol. 5, no. 210, p. 210ed19, Nov. 2013.

- [170] J. B. Gafford, S. B. Kesner, R. J. Wood, and C. J. Walsh, "Force-sensing surgical grasper enabled by pop-up book MEMS," in *2013 IEEE/RSJ International Conference on Intelligent Robots and Systems*, 2013, pp. 2552–2558.
- [171] M. Lerotic, A. J. Chung, G. Mylonas, and G. Z. Yang, "Pq-space based non-photorealistic rendering for augmented reality," *Med. Image Comput. Comput. Assist. Interv.*, vol. 10, no. Pt 2, pp. 102–109, 2007.
- [172] G. A. Gescheider, *Psychophysics: the fundamentals*. Psychology Press, 2013.
- [173] B. Wu, R. Klatzky, R. Lee, V. Shivaprabhu, J. Galeotti, M. Siegel, J. S. Schuman, R. Hollis, and G. Stetten, "Psychophysical Evaluation of Haptic Perception Under Augmentation by a Handheld Device," *Hum. Factors J. Hum. Factors Ergon. Soc.*, Sep. 2014.
- [174] R. L. Klatzky, P. Gershon, V. Shivaprabhu, R. Lee, B. Wu, G. Stetten, and R. H. Swendsen, "A model of motor performance during surface penetration: from physics to voluntary control.," *Exp. brain Res.*, vol. 230, no. 2, pp. 251–60, Oct. 2013.
- [175] K.-W. Kwok, G. P. Mylonas, L. W. Sun, M. Lerotic, J. Clark, T. Athanasiou, A. Darzi, and G.-Z. Yang, "Dynamic active constraints for hyper-redundant flexible robots," *Med. Image Comput. Comput. Assist. Interv.*, vol. 12, no. Pt 1, pp. 410–7, Jan. 2009.
- [176] K.-W. Kwok, K. H. Tsoi, V. Vitiello, J. Clark, G. C. T. Chow, W. Luk, and G.-Z. Yang, "Dimensionality Reduction in Controlling Articulated Snake Robot for Endoscopy Under Dynamic Active Constraints.," *IEEE Trans. Robot.*, vol. 29, no. 1, pp. 15–31, Feb. 2013.
- [177] J. Barrie, Z. Ehteshami, P. Culmer, D. G. Jayne, and A. Neville, "Relaxation and Recovery of Colon After Application of a Mechanical Stress," in *Biomedical Engineering*, 2013.
- [178] G. Kontorinis, T. Lenarz, T. Stöver, and G. Paasche, "Impact of the insertion speed of cochlear implant electrodes on the insertion forces.," *Otol. Neurotol.*, vol. 32, no. 4, pp. 565–70, Jun. 2011.
- [179] H. Rafii-Tari, C. J. Payne, C. Riga, C. Bicknell, S.-L. Lee, and G.-Z. Yang, "Assessment of navigation cues with proximal force sensing during endovascular catheterization.," *Med. Image Comput. Comput. Assist. Interv.*, vol. 15, no. Pt 2, pp. 560–7, 2012.
- [180] T. P. Cundy, E. Thangaraj, H. Rafii-Tari, C. J. Payne, G. Azzie, M. H. Sodergren, G.-Z. Yang, and A. Darzi, "Force-Sensing Enhanced Simulation Environment (ForSense) for laparoscopic surgery training and assessment.," *Surgery*, vol. 157, no. 4, pp. 723–31, Apr. 2015.
- [181] J. Lee, B. A. Gozen, and O. B. Ozdoganlar, "Modeling and experimentation of bone drilling forces.," *J. Biomech.*, vol. 45, no. 6, pp. 1076–83, Apr. 2012.